THE RESPONSE OF ELDERLY FEMALE FAST GAIT TO WHOLE BODY VIBRATION

Submitted by
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Declaration

This thesis contains no material published elsewhere or extracted in whole or in part from a thesis by which I have qualified for or been awarded another degree or diploma. No other person's work has been used without due acknowledgement in the main text of the thesis. This thesis has not been submitted for the award of any degree or diploma in any other tertiary institution. All research procedures reported in this thesis received the approval of the ACU National Research Ethics Committee.

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Christian Lorenzen Date

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Background:

Older adults walk more slowly than healthy young adults at fast and normal walking speeds. These age-associated changes in mobility impact upon daily function. A slower gait, for example, may reduce the older adult's ability to safely cross at traffic intersections due to the time restriction. Recent research has demonstrated whole body vibration (WBV) can improve the strength and power (Roelants, Delecluse & Verschueren, 2004; Russo et al., 2003; Verschueren, Roelants, Delecluse, Swinnen, Vanderschueren & Boonen, 2004) of community dwelling elderly females, and the mobility of nursing home residents (Bautmans, Van Hees, Lemper & Mets, 2005; Bruyere et al., 2005). To date, no published research has examined the impact WBV has on the gait parameters of community dwelling elderly females. The research was conducted in three phases.

Phase One – Development of a WBV Platform:

This phase outlines the development of a WBV platform (ACUWBV) that was designed and built for this research. A unique aspect of the ACUWBV was the method of adjusting WBV amplitude and therefore intensity. Current WBV technology, using tilting oscillations, requires the individual to increase their stance width. The ACUWBV allowed for the adjustment of WBV amplitude while maintaining the same stance width. The reliability and accuracy of the ACUWBV eccentric cam was measured during this phase of the research. Although an intraclass correlation coefficient of 0.4 was calculated and is considered an indication of low reliability, calculations of typical error (TE - 95% error range) for each amplitude indicated the error to be small in the

overall precision of the instrument. Specifically, at a frequency of 20 Hz, the expected WBV acceleration ranges for amplitudes of 0.5 mm and 1.0 mm were 7.58 $m.s^{-2}$ to 8.85 $m.s^{-2}$ (TE = 0.02 mm) and 16.90 $m.s^{-2}$ to 17.53 $m.s.^{-2}$ (TE = 0.01 mm), respectively.

Phase Two – Pilot Study:

This phase established the response of elderly community-dwelling female fast gait to WBV. Seven elderly female participants attended three WBV sessions per week for three weeks. Participants performed fast walks over an electronic walkway (GAITRite) at the end of each WBV session. A time-series graph displayed a linear increase in stride velocity over the three week intervention period. Conversely, stride time, stance time and double support time exhibited linear decreases. However, stride time ($p=0.04$) and stance time ($p=0.04$) were the only variables that exhibited a significant difference. It was concluded that the linear changes in stride velocity, stride time, stance time and double support time warranted further investigation with a larger sample size within a longer intervention period.

Phase Three – Major Study:

Phase three was an extension of phase two. This WBV intervention study was performed over a twelve week period. Twenty-two elderly female participants were placed in one of two groups. Group one (placebo/WBV; Group; n=12) was exposed to a placebo intervention for the first six weeks followed by a six week WBV intervention. Group two (Group WBV/placebo; n=10) was exposed to WBV for the first six weeks and a placebo intervention

for the following six weeks. Group placebo/WBV exhibited no change in stride velocity during the placebo period, but a seven per cent increase during the six week WBV period (p=0.005). The changes in stride velocity coincided with increases in stride length (p=0.017), and reductions in stride time (p=0.007), stance time (p=0.001) and double support time (p=0.001). Group WBV/Placebo demonstrated stride velocity to increase by five per cent during the WBV period. Although the time-series graphs demonstrated improvements in stride velocity to be associated with decreases in stride time, stance time, and double support time, the changes failed to reach significance. Single support time and stride length showed no change over the WBV period. The improvements shown by group WBV/placebo from the first six weeks of WBV were maintained during the six week placebo (detraining) period. In summary, WBV was an effective intervention for enhancing the walking speed of community dwelling elderly female gait. This form of exercise may have positive outcomes on the daily function of elderly females, which in turn may improve their quality of life.

1.0 Overview

The Disablement Process, of which an abridged model is presented in figure 1.1, provides a conceptual model guiding the terminology, measurement and hypotheses of disability (Verbrugge & Jette, 1994). This model was used as a framework for the subsequent literature review to describe: (1) how chronic conditions affect functioning in older adults during the basic physical action of walking; and (2) the potential benefit of whole body vibration as an intervention to retard or reverse the deficits in walking function present in community dwelling older adults.

The diminished walking velocity exhibited by older adults restricts their capacity to function effectively and safely within the community. A slower gait, for example, may reduce the older adult's ability to safely cross at traffic intersections due to the time restriction (Oxley, 2004). Recent research has demonstrated whole body vibration (WBV) can improve the strength and power of community dwelling elderly females, and the mobility of nursing home residents. To date, no published research has examined the impact this form of exercise has on the gait parameters of community dwelling elderly females.

An "older adult" was defined as an individual 65 years and older (Pikna, 2001). This population is typically subdivided by chronological age into young-old (65 to 74 yrs), middle-old (75 to 84 yrs), and old-old (85 + years) to reflect the functional changes that occur throughout these phases of life. These categories, however, may not always be appropriate for categorising

the function of individuals as they are considered to be poor predictors of biological function (Pikna, 2001).

1.1 The Disablement Process

The disablement process describes the influence chronic and acute conditions have on the physiological function of individuals, and the ability to function in the community at a level that is considered essential, normal, and personally desirable (Verbrugge & Jette, 1994). This model is applicable to the various forms and onsets of disablement, including periods of brief disability (e.g. the common cold), early onset childhood conditions (e.g. spinal cord injury) as well as the chronic conditions associated with gerontology.

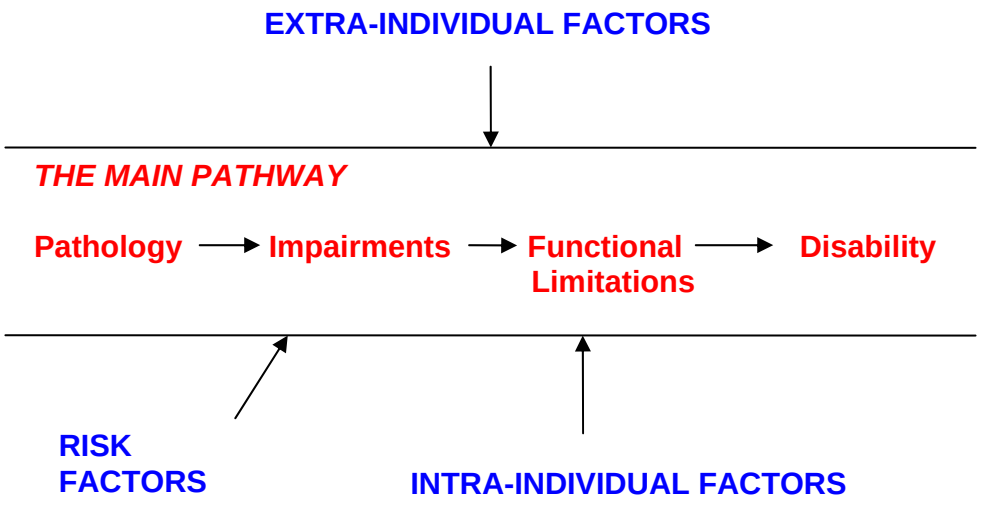


Figure 1.1: Abridged model of Disablement Process (Verbrugge & Jette, 1994). See Appendix A for the complete model.

There are four key concepts in the disablement process: (1) pathology; (2) impairments; (3) functional limitations; and (4) disability. The following sections describe these concepts as outlined by Verbrugge & Jette (1994).

1.1.1 Pathology

Pathology is considered to be the deviation from the normal biochemical and physiological condition. These are medically categorized as disease, injury or congenital/developmental conditions (Verbrugge & Jette, 1994). Examples of pathology include osteoarthritis, cataracts, Alzheimer's disease and diabetes (Verbrugge & Jette, 1994).

1.1.2 Impairments

Generally, impairments are recognised when the pathology has crossed clinically defined thresholds (Verbrugge & Jette, 1994). Impairments are dysfunctions and structural irregularities that impact upon the individual's physical, mental and/or social functioning (Verbrugge & Jette, 1994). The level of impairment is used to determine the extent of the pathology (Verbrugge & Jette, 1994).

1.1.3 Functional Limitations

Functional limitations refer to the age and gender specific limitations that the impairments have on the individual's fundamental physical and mental

function at levels that are considered 'normal' (Verbrugge & Jette, 1994). Fundamental physical functions include mobility, vision, hearing and communication. Fundamental mental functions include short-term memory, intelligible speech and orientation in time and space (Verbrugge & Jette, 1994). Functional limitations are regularly measured by performance based measures (Verbrugge & Jette, 1994). The functional mobility of older adults, for example, is often assessed by measuring their walking speed (Bendall, Bassey and Pearson, 1989; Bohannon, 1997; Elbe, Thomas, Higgins & Colliver, 1991; Hageman & Blanke, 1986; Leiper & Craik, 1991; Lord, Lloyd & Li, 1996; Menz, Latt, Tiedemann, Mun San Kwan & Lord, 2004; Öberg, Karsznia & Öberg, 1993; Ostrosky, VanSwearingen, Burdett & Gee, 1994; Winter, Patla, Frank & Walt, 1990; Woo, Ho, Lau, Chan & Yuen, 1995).

1.1.4 Disability

Disability occurs when an individual has difficulty completing activities specific to their age and gender due to any health or physical condition (Verbrugge & Jette, 1994). Activities include those necessary for hygiene, performing hobbies, work, fiscal management, household management, shopping and sleep (Verbrugge & Jette, 1994). According to the severity, the individual may be dependent (i.e. require help from someone else) or require an affordance to reduce disability (e.g. a walking frame) (Verbrugge & Jette, 1994). Disability is differentiated from functional limitations by the impact it has upon the individual in a social context, where the individual is unable to perform at a level that is expected of them by society (Verbrugge & Jette, 1994).

1.1.5 Extra-Individual, Intra-Individual and Risk Factors

External means used to ameliorate individual impairment, functional limitation and disability are termed extra-individual factors (Verbrugge & Jette, 1994). These include medical care and rehabilitation (e.g. surgery, physical therapy), medications and other therapeutic regimens (e.g. drugs, exercise), external supports (e.g. personal assistance, specialised equipment and devices) and changes to the physical and social environment (e.g. structural modifications at job/home) (Verbrugge & Jette, 1994). Intra-individual factors refer to those strategies that come from within the individual. These intrinsic factors include lifestyle and behavioural changes (e.g. changes to alter disease impact), psychosocial attributes and coping (e.g. emotional vigour, prayer), and activity accommodations (e.g. changes in procedure for doing activity) (Verbrugge & Jette, 1994). The disablement process model also recognises risk factors that may lead to impairment including the individual's lifestyle and genetic inheritance. These factors are regarded as predisposing since they existed before the onset of the disablement process (Verbrugge & Jette, 1994).

1.1.6 A Modified Model of the Disablement Process

Since several of the variables of this model are outside the scope of the present research, a modified model of the disablement process is outlined in figure 1.2. Although recognised that the reduced walking function exhibited by older adults may be due to various pathologies including, but not limited to osteoarthritis, emphysema, cataracts, Parkinson's disease, peripheral neural degeneration or cognitive conditions such as dementia; healthy older adults

were recruited for this research and therefore these pathologies are not considered in this modified model. For this research, gait analysis was used to examine the changes in gait of healthy older adults following whole body vibration.

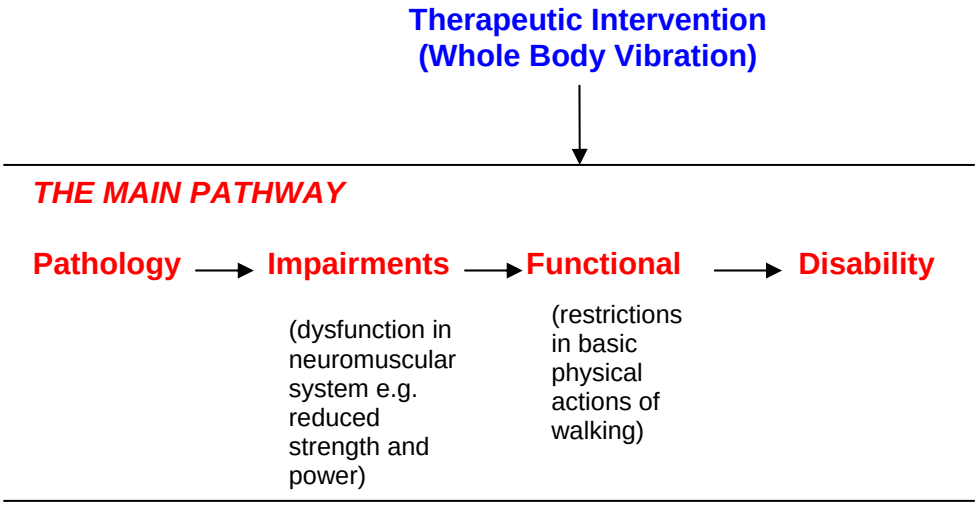


Figure 1.2: A simplified model of the Disablement Process specific to this research

1.2 Australian Aging Demographics

Australia has a growing population of older adults providing a challenge to its economy and health care system. Improvements in health technology have contributed to the extended lifespan of Australians. Over the past 20 years the elderly population in Australia has grown by 165%, compared to the 29% in total population growth over the same period (Australian Bureau of Statistics, 2003). During this time period the median age of the Australian population has increased from 30.2 years to 36.1 years (Australian Bureau of Statistics, 2003). The major contributors to this increase in median age are the current low level of human fertility and an increased human life expectancy. Two and a half million Australians are aged 65 years and over

which comprises 10.7% of the total Australian population (Australian Bureau of Statistics, 2003). It has been projected that by 2051, one-quarter of the Australian population will be 65 years and older (Australian Bureau of Statistics, 1999). The largest increases have occurred in people aged 85 years and over (Australian Bureau of Statistics, 2003). The life-expectancy of Australians is continuing to rise. The 2002 median age at death was 76.2 years for males and 82.2 years for females, representing increases of 6.2 and 5.0 years respectively (Australian Bureau of Statistics, 2002). The fact that there are twice as many women as men 85 years and older reflects the higher life-expectancy of women (Australian Bureau of Statistics, 2003).

Falls are a significant problem for both the elderly and broader community. Injuries resulting from falls include lacerations, soft tissue injuries, haematomas and fractures to the upper limb, lower limb, hip, head and trunk (Hall & Hendrie, 2003). Older women are more prone than older men to have a fall and are more likely to sustain a fracture from a fall (Hall & Hendrie, 2003). A Western Australian study reported the fall-related national cost of hospitalisation and for the three-month post-hospitalisation period after the fall to be $287 million (Hall & Hendrie, 2003). Hospitalisation accounted for approximately 80% of these costs. The federal and state/local governments respectively fund approximately 50% and 43% of these costs. The patient (6%) and insurance companies (1%) meet the final 7% of these costs (Hall & Hendrie, 2003).

Considering the economic burden of falls and the resulting personal hardships, numerous interventions have been used to address these problems (Kannus et al., 1999). The following literature review introduces whole body vibration as a potential mechanism that can be used to intervene with gait disablement. Previous research has shown that increasing the strength and power of elderly lower limbs with resistance training can improve their walking velocity (Fiatarone et al., 1990; Heiwe, Tollbäck, & Clyne, 2001; Judge, Underwood & Gennosa, 1993; Rubenstein et al., 2000; Schlicht, Camaione & Owen, 2001; Topp, Mikesky, Dayhoff & Holt, 1996). Whole body vibration has also been shown to increase the lower limb strength and power of older adults (Russo et al., 2003; Verschueren et al., 2004; Roelants, Delecluse & Verschueren, 2004) and for that reason it is hypothesized that this form of exercise will also improve the walking velocity of this age group.

The review begins at the third stage of the disablement process (functional stage), comparing the gait of younger and older adults. The first and second stages follow, outlining common pathologies and impairments that have been associated with changes in gait. Finally, the current literature for whole body vibration was reviewed and a rationale put forward for the use of whole body vibration as an intervention for gait disablement.

2.0 Human Gait Nomenclature

Repetitive events are consistently observed during normal human walking and running (Sutherland, Kaufman & Moitoza, 1994). The successive occurrence of one of these repetitive events is termed the *gait cycle* (Whittle, 2002). Both spatial and temporal parameters are used to describe the footfall patterns of the gait cycle.

The most common event to indicate the beginning and ending of a walking gait cycle is heel contact, also termed initial contact or foot strike (Whittle, 2002). However, any cyclical event that sequentially demarcates two steps, such as left toe off to left toe off or right peak knee flexion to right peak knee flexion may be used to indicate the beginning and ending of the gait cycle (Craik & Dutterer, 1995; Whittle, 2002). *Stride length* (Figure 2.1) is the distance from a single event such as heel contact, to the same contact event of the same foot (Craik & Dutterer, 1995; Whittle, 2002). A stride consists of two step lengths (Figure 2.1), that is, left and right steps and is distinguished by the same event on the same leg (e.g. heel contact to heel contact) (Whittle, 2002). The advancing limb is used as a reference for defining step length. Left step length would therefore be represented by a right heel contact succeeded by forward progression of the left leg and heel contact (Figure 2.1). It should be noted that inconsistent definitions for stride length and step length have been applied within the gait literature. Clinical gait literature typically defines stride length and step length as it is presented above (Craik & Dutterer, 1995; Sutherland et al., 1994; Whittle, 2002). Alternatively, sports biomechanics literature related to track and field events has defined a stride as the distance between two consecutive foot touchdowns (Hay & Nohara, 1990; Bradshaw & Aisbett, 2006).

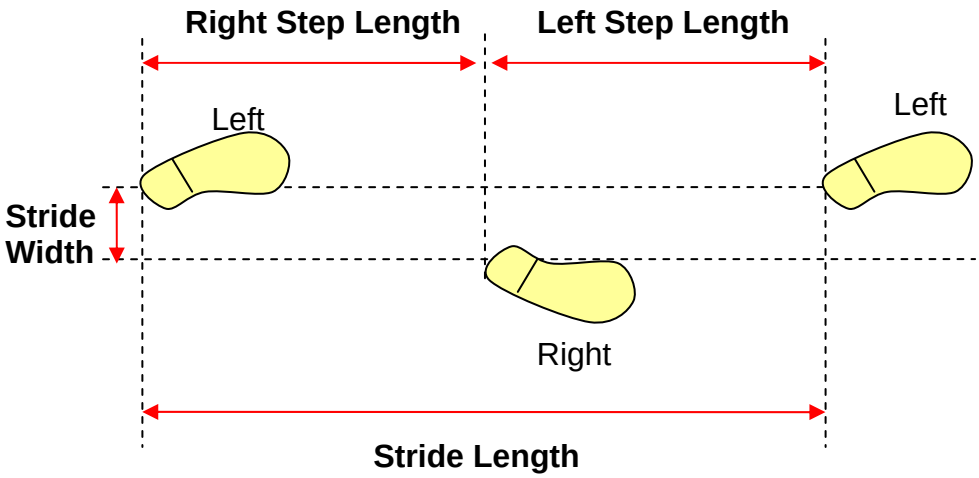


Figure 2.1: Illustration of foot placement during the gait cycle
(Adapted from Whittle, 2002).

Stride width has also been termed the base of support and walking base (Craik & Dutterer, 1995; Whittle, 2002). Various definitions for this gait feature have been applied by researchers making clear comparisons across research difficult (Craik & Dutterer, 1995). This feature has been described as the distance between the line of two feet, and is measured from the centre of the back of the heel (Figure 2.1) (Whittle, 2002). Step width, the single step component of stride width, has been measured as the distance between the outermost two borders of two consecutive footprints (Figure 2.2) (Brach, Berthold, Craik, VanSwearingen & Newman, 2001; Brach, Berlin, Van Swearingen, Newman & Studenski, 2005). The technology used to measure stride width may influence the definition applied by researchers. This is demonstrated by two different electronic walkways that have been reported in the literature. Electronic walkways have built in sensor pads that provide spatio-temporal data as the participant walks across the walkway. The GAITRite walkway calculates stride width as the vertical distance of one foot print to the line formed by the midline midpoints of the opposite foot

(Figure 2.3). In contrast, the GaitMat II™ system calculates step width as the distance between the outermost two borders of two consecutive foot prints. Therefore if stride width is to be compared across studies, the researcher must examine the method of measurement used by each study.

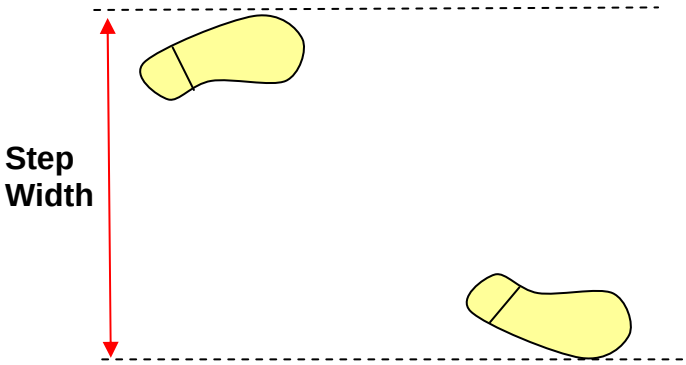


Figure 2.2: The measurement of step width by the GaitMat II™ system (Brach et al., 2001)

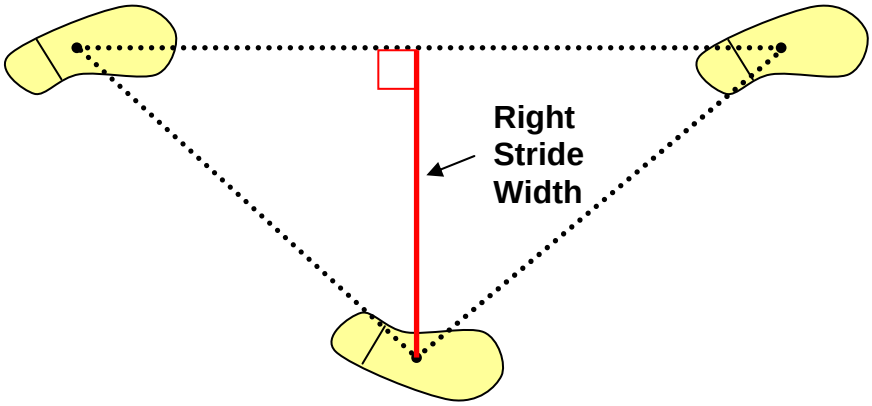


Figure 2.3: The measurement of stride width by the GAITRite walkway

The period of time to complete one stride cycle is termed stride time. The gait cycle is broken up into two phases, the stance phase and swing phase, which are further broken into sub-phases (Figure 2.4). The *swing phase,* measured temporally as *swing time,* is the interval of the gait cycle when the leg progresses backwards and upwards (recovers) before driving forwards and downwards through the air, and is not in contact with the ground. The *stance phase* or *support phase* is the period of time when the limb is in contact with the ground and consists of the time from heel contact through to toe off. The *double support* phase occurs when both limbs are in contact with the ground. This period of the gait cycle is characterised by the stance phase of one leg overlapping the stance phase of the contralateral leg. When only one limb is in contact with the ground it is described as the *single support phase*. One walking gait cycle has two single support and two double support phases (Craik & Dutterer, 1995; Sutherland et al., 1994; Whittle, 2002). Walking and running gait is distinguished by the presence or absence of a double support phase (Whittle, 2002).

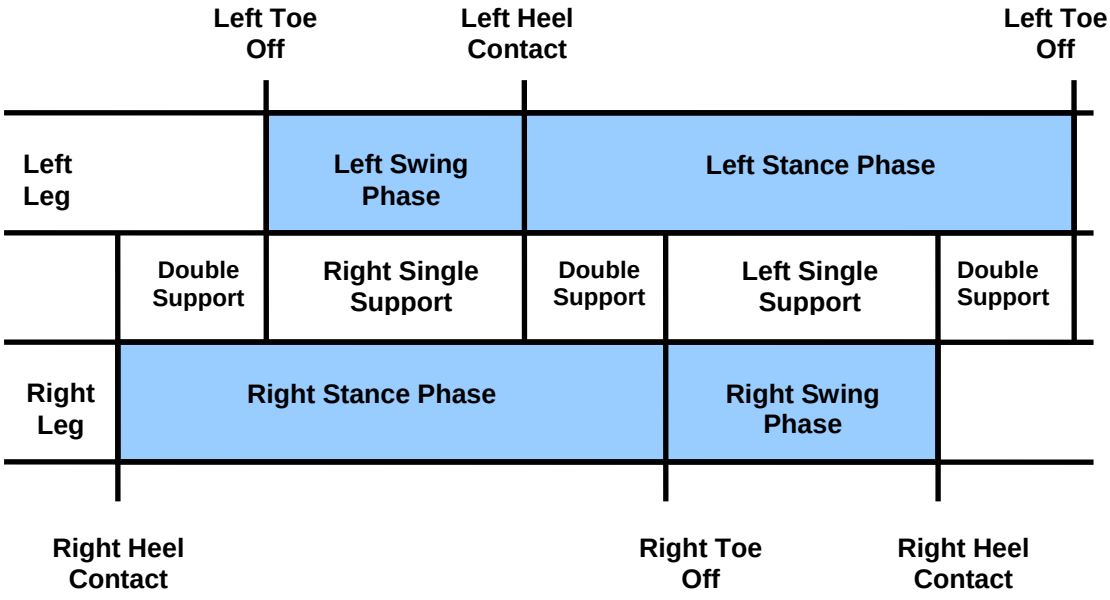


Figure 2.4: The major phases of the gait cycle (Adapted from Whittle, 2002).

2.1 Gait of Older Adults

Compared to healthy young adults, healthy older adults walk unequivocally slower at preferred and fast walking speeds (Bendall et al., 1989; Bohannon, 1997; Elbe et al., 1991; Hageman & Blanke, 1986; Leiper & Craik, 1991; Lord et al., 1996; Menz, Latt et al., 2004; Murray, Kory & Clarkson, 1969; Ostrosky et al., 1994; Winter et al., 1990; Woo et al., 1995). The average difference in walking velocity between the young and old has been reported to range between 7%-19% (Table 2.1).

Table 2.1: Walking Velocity (Vel.), stride lengths (SL) and percentage difference in walking velocity reported in the literature for young adults (YA) and older adults (OA) walking at normal speed.

Author(s)	Vel. of YA ($cm.s^{-1}$)	Vel. of OA ($cm.s^{-1}$)	% Diff. in Vel.	SL of YA (cm)	SL of OA (cm)	% Diff. in SL.
Elble et al., (1991) [a]	167.0	139.0	-16.8	132.0	108.0	-18.2
Hageman & Blanke (1986) [b]	159.5	131.9	-17.3	162.7	134.9	-17.1
Menz et al., (2004) [c]	143.0	116.0	-18.9	154.9	127.8	-17.5
Öberg et al., (1993) [d]	124.1	115.7	-6.8	118.2	110.6	-6.4
Ostrosky et al., (1994) [e]	138.0	127.0	-7.9	152.0	141.8	-6.7

Note: [a] males and females (YA = 20-39 years; OA = 65-87 years); [b] females (YA = 20-35 years; OA = 60-84 years); [c] males and females (YA = 22-40 years; OA = 76-87 years); [d] females (YA = 20-29 years; OA = 60-69 years); [e] males and females (YA = 22-39 years; OA = 60-80 years).

Gait velocity decreases somewhere between the third and fifth decades (Bohannon, 1997; Lord et al., 1996; Öberg et al., 1993) and this decline is larger for fast gait speed compared to preferred gait speed (Bohannon, 1997). After 65 years of age, normal gait velocity has been reported to decrease by 0.7% per year (Bendall et al., 1989) which may be accentuated during the first year of retirement where it has been reported to decrease by 4% for normal walking and

3% for fast walking speeds (Bassey, Fentem, MacDonald, Patrick & Scriven, 1977).

The shortened stride length exhibited by older adults (Table 2.1) is considered to be the most significant contributor to their slower walking velocity (Elble et al., 1991; Gabell & Nayak, 1984; Hageman & Blanke, 1986; Laufer, 2005; Lord et al., 1996; Menz, Latt et al., 2004; Murray et al., 1969; Öberg et al., 1993; Ostrosky et al., 1994; Winter et al; 1990; Woo et al., 1995). Although cadence has been reported to decline with age (Laufer 2005; Lord et al., 1996), this parameter has generally been reported to have little or no change with age for normal or fast walking velocities (Elble et al., 1991; Menz, Latt et al., 2004; Öberg et al., 1993; Ostrosky et al., 1994; Winter et al., 1990). These gait changes occur in both men and women, although women generally display a slower walking velocity (Bohannon, 1997; Öberg et al., 1993; Woo et al., 1995) and have smaller stride lengths than older men (Öberg et al, 1993; Kwon, Oldaker, Schrager, Talbot, Fozard & Metter, 2001; Woo et al, 1995).

The reduced stride velocity and stride length of older adults is associated with an increased stance time (Judge, Õunpuu & Davis, 1996; Lord et al., 1996; Murray et al., 1969; Winter et al., 1990) and a corresponding increase in double support time (Elble et al., 1991; Judge et al., 1996; Laufer, 2005; Winter et al., 1990). Although the literature is unanimous that stride length decreases with age, it is less clear what a normal stride length is for young or older adults (Table 2.1). At self-selected speeds, reported stride lengths for young adults have ranged between 118.2 cm (Öberg et al., 1993) to 162.7 cm (Hageman & Blanke, 1986)

and for older adults stride length has ranged between 108.0 cm (Elble et al., 1991) to 141.8 cm (Ostrosky et al., 1994).

Stride width is considered to be an important variable of balance control (Gabell & Nayak, 1984). Young and old adults appear to have similar stride widths (Elble et al., 1991; Gabell & Nayak, 1984; Grabiner, Biswas & Grabiner, 2001; Owings & Grabiner, 2004), until normalised with walking velocity where it decreases with age (Helbostad & Moe-Nilssen, 2003). In older adults there appears to be a U-shaped relationship between gait speed and step width, where step width widens at slow and very fast speeds and narrows at normal and fast speeds (Helbostad & Moe-Nilssen, 2003).

2.2 Issues Concerning the Reporting of Gait Data

Varied methods of collecting and analysing gait data have been reported by researchers, making comparisons across studies somewhat difficult. Three major areas of inconsistency throughout the gait literature have been: (1) the nomenclature used by researchers; (2) the methods for reporting gait data; and (3) the selection processes during participant recruitment. The discrepancies existing throughout the literature for gait nomenclature was addressed in Section 2.0. The following discussion provides an overview of the latter two points.

Although researchers have attempted to quantify the typical gait of older adults with parameters such as walking speed and stride length, methodological differences between studies have resulted in inconsistent findings. It appears that no two studies have used the same methodology. Inconsistent definitions of young and older adult age groups (Table 2.1), various methods of collecting and

interpreting spatio-temporal data, and inconsistent use of definitions across studies have all complicated interpretation. This problem is illustrated by the question of the normalisation of gait. Stature (Bohannon, 1997; Winter et al., 1990) or leg length (Leiper & Craik, 1991) are typically used to normalise gait parameters. These two parameters have a positive association with stride velocity and stride length (Dobbs et al., 1993; Woo et al., 1995), therefore tall people or individuals with longer legs will generally walk faster and have longer stride lengths than shorter people. It has been argued that normalising data with stature or leg length only, makes small differences to normal stride velocity and the benefit gained from accuracy is lost to obscure units such as “statures per second” or “stride length/lower extremity length” making the data less meaningful (Kirtley, Whittle & Jefferson, 1985). If the purpose of the research is to present descriptive data of typical gait, it is essential that the data is normalised, as without doing so the data is in effect meaningless for comparison with similar or different sample populations (Bohannon, 1997). For research with a repeated measures design investigating the impact of an intervention, it is unnecessary to normalise the gait data as the participants’ pre- and post-data are being compared rather than individuals with different statures and/or leg lengths.

Differences in reported gait parameters may also be due to the participant selection process. There is a large amount of variability across studies for the definition of healthy older adults. The most common exclusion criteria are neurological (Bohannon, 1997; Blanke & Hageman, 1989; Dobbs et al., 1993; Elble et al., 1991; Hageman & Blanke, 1986; Helbostad & Moe-Nilssen, 2003) and musculoskeletal disorders (Bohannon, 1997; Blanke & Hageman, 1989; Dobbs et al., 1993; Elble et al., 1991; Hageman & Blanke, 1986), and requiring

an assisting device when walking (Bohannon, 1997; Dobbs et al., 1993; Leiper & Craik, 1991; Lord et al., 1996; Woo et al., 1995). Several studies did not report screening for other variables that can have significant effects on gait. These include a history of falling (Hageman & Blanke, 1986; Ostrosky et al., 1994), medication use (Blanke & Hageman, 1989; Elble et al., 1991; Ostrosky et al., 1994; Winter et al., 1990) and cognitive function (Blanke & Hageman, 1989; Bohannon, 1997; Elble et al., 1991; Hageman & Blanke, 1986; Ostrosky et al., 1994; Winter et al., 1990). Other studies only described their participants as healthy (Kirtley et al., 1985; Öberg et al., 1993) or free from conditions that may impair gait (Leiper & Craik, 1991; Winter et al., 1990). Varied exclusion criteria may therefore also provide an explanation for the differences reported across studies.

Summary of Key Points

There are several confounding variables that have impacted upon the interpretation of the reported elderly gait kinematics. For clear interpretation and comparisons, accurate explanations of exclusion criteria should be provided. Despite these limitations, it is clear that walking velocity and stride length decrease with age, corresponding with increases in stance and double support time. The research has generally indicated that cadence and stride width do not change with age.

2.3 Mechanisms Leading to Changes in Gait

Numerous studies have identified the various age-related differences in gait, generally presenting these differences using spatio-temporal parameters. Research is now attempting to identify gait changes occurring due to the natural ageing process, pathological conditions and lifestyle behaviour. In order to develop appropriate rehabilitation programs, it is necessary to have an understanding of the processes associated with declining gait. This requires a multi-disciplinary approach (Larish, Martin & Mungiole, 1988) that considers the pathological, physiological and psychological variables that may contribute to the changed gait.

Diminishing gait function has been related to decreased functional status, reduced neuropsychological status (e.g. increased depressive symptoms, dementia), decreased physiological capacity (e.g. reduced exercise capacity and balance), lower physical activity levels, and lower health status (Hausdorff et al., 2001). The decline in functional mobility associated with age is most likely due to an accumulation of deficits throughout the sensory, effector, and central processing domains rather than the result of one underlying deficit (Duncan, Chandler, Studenski, Hughes & Prescott, 1993; Hausdorff et al., 2001; Lord et al., 1996).

Figure 2.5 illustrates the multifactorial relationship of physiological and neuropsychological variables leading to gait instability in older adults (Hausdorff et al., 2001). Visual, vestibular and proprioceptive function decline with age

compromising the quality of sensory information provided to the older adult (Borel, Harlay, Lopez, Magnan, Chays & Lacour, 2004; Lord & Dayhew, 2001; Lord et al., 1996; Woolley, Czaja & Drury, 1997). The quality of response output also declines corresponding to a reduced capacity for central processing presented as an increased response time (Lord & Fitzpatrick, 2001), decreased muscular strength and power (Akima et al., 2001; Frontera, Hughes, Fielding, Fiatarone, Evans, Roubenoff, 2000; Häkkinen, Kraemer, Kallinen, Linnamo, Pastinen & Newton, 1996; Hughes et al., 2001; Izquierdo et al., 2001; Landers, Hunter, Wetzstein, Bamman & Weinsier, 2001; Samson, Meeuwsen, Crowe, Dessens, Duursma & Verhaar, 2000) and impaired coordination (Barry, Riek & Carson, 2005; Paquette, Paquet & Fung, 2006).

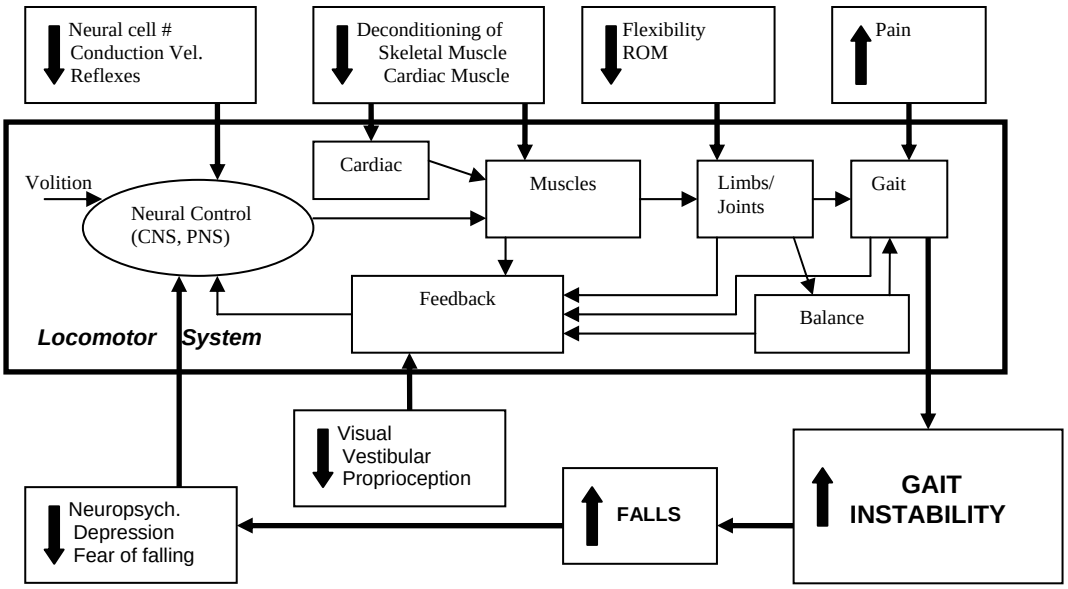


Figure 2.5: Diagram illustrating the multifactorial relationship of physiological and neuropsychological variables leading to gait instability in older adults (Hausdorff et al., 2001).

Although older adults tend to describe “old age” as the cause of their physical disability, chronic disease is the predominant etiology for these decrements

(Williamson & Fried, 1996). Geriatric conditions such as memory impairment and frailty are no longer considered the sole result of the ageing process but are now also attributed to pathophysiological and lifestyle factors (Hubert, Bloch, Oehlert & Fries, 2002; Simons, McCallum, Friedlander & Simons, 2000; Williamson & Fried, 1996).

The variables outlined in figure 2.5 are expanded upon in the following sub-chapters (i.e. sections 2.3.1-2.3.5). A review of how these variables impact upon gait provides a background to the reader regarding the development of the research methodology outlined in this research. The review also serves to establish a theoretical basis for justifying WBV as an intervention for enhancing the gait of older adults.

2.3.1 Cognition

Dementia is associated with declining functional status (Atkinson et al., 2005; Black & Rush, 2002; Visser, 1983; Woo et al., 1995). Patients with dementia walk slower than unaffected individuals (Bramell-Risberg, Jarnlo, Minthon & Elmståhl, 2005; Goldman, Baty, Buckles, Sahrmann & Morris, 1999; Tanaka, Okuzumi, Kobayashi, Murai, Meguro & Nakamura, 1995; Visser, 1983), have shorter step lengths (Tanaka et al., 1995; Visser, 1983), a lower stepping frequency and spend more time in double support (Visser, 1983). The most prevalent type of dementia suffered by the aged is Alzheimer's disease (Rahkonen, Eloniemi-Sulkava, Rissanen, Vatanen, Viramo & Sulkava, 2003). Compared to the healthy elderly, individuals displaying symptoms of Alzheimer's walk 30 to 40% slower, take shorter steps, have a lower cadence

and spend more time in double support (Alexander et al., 1995). Individuals with dementia are also more likely to fall than individuals without dementia (Allan, Ballard, Burn & Kenny, 2005; Bergland & Wyller, 2004).

The slower walking velocity of women may also be associated with cognition. Women have been reported to be more conservative during spatio-temporal judgement tasks such as when predicting the arrival of approaching vehicles. This perhaps cautious decision-making characteristic of women has been suggested to have survival importance as it reduces the potential for injury (Schiff & Oldak, 1990). Although this research was non-specific to gait, recent gait research may support this idea. When approaching a step, elderly females have been demonstrated to be more cautious than young women during the approach, making earlier and larger stepping adjustments (Lythgo, Begg & Best, 2006).

2.3.2 Disease

Neuromuscular and cardiovascular disease are associated with diminishing gait function in the elderly (Adamson, Hunt & Ebrahim, 2003; Bloem, Gussekloo, Lagaay, Remarque, Haan & Westendorp, 2000; Hausdorff, Herman, Baltadjieva, Gurevich & Giladi, 2003; Sofuwa, Nieuwboer, Desloovere, Willems, Chavret & Jonkers, 2005). The greater the number of chronic conditions an individual is afflicted by, the higher the level of disability (Adamson et al., 2003). Conditions that impact upon gait and are typically considered within gait research literature are Parkinson's disease, peripheral neuropathy, diabetes, congestive heart failure and hypertension.

An epidemiological study identified Sydney, Australia, to have one of the most prevalent Parkinson's disease estimates in the Western world (Chan et al., 2005). Parkinson's disease is a neurodegenerative disorder commonly found in the elderly. Prevalence estimates have been shown to be 15% for people aged 65 to 74 years, 30% for 75 to 84 years, and 50% in individuals 95 years and older (Bennett et al., 1996). People suffering Parkinson's disease walk with a shorter stride length and have a slower walking velocity than healthy older adults (Sofuwa et al., 2005).

Individuals with peripheral neuropathy typically demonstrate decreased sensitivity and movement impairment at the feet and ankles. This condition is found in approximately 8% of people aged 55 and over (England & Asbury, 2004) and increases in prevalence with age (Mold, Vesely, Keyl, Schenk & Roberts, 2004). These changes in the sensorimotor function negatively impact upon balance and increase fall risk (Koski, Luukinen, Laippala & Kivelä, 1998; Richardson, 2002). To compensate for the reduced balance function, these individuals walk with a shorter stride length and increased stride time (Richardson, Thies, DeMott & Ashton-Miller, 2004). However, peripheral neuropathy appears to have little to no impact upon step width (Richardson et al., 2004).

Peripheral neuropathy may occur in the later stages of diabetes mellitus (Sosenko et al., 1999). Diabetes is a worldwide problem that is becoming increasingly more common in the 65 years and older age group (Wild, Roglic, Green, Sicree & King, 2004). Diabetics present with an atypical gait pattern that has been attributed to several mechanisms including impaired somatic, visual and vestibular systems, as well as reduced sensitivity of the feet (Petrofsky, Lee,

& Bweir, 2005). The diminished function of these sensory systems has been linked to ineffective microcirculation (Stansberry, Hill, Shapiro, McNitt, Bhatt & Vinik, 1997) and may occur even before clinical diabetes diagnosis has been made (Sagliocco, Sartucci, Giampietro & Murri, 1999).

The inadequate circulation and reduced sensitivity of the diabetic foot leads to 1 in 20 diabetics experiencing foot ulcerations (Boulton, 2004). This impacts upon foot pressure distribution during the stance phase, possibly to abate stress of the injured region(s) (Sacco & Amadio, 2000). The reduced sensitivity of the feet, compromised somatic, visual and vestibular systems, and the tendency to protect the injured areas of the foot may all contribute to diabetics walking with a cautious gait. Compared to asymptomatic individuals, diabetics typically walk slower (Menz, Lord, St George & Fitzpatrick, 2004; Petrofsky et al., 2005), have reduced step lengths and cadence (Menz, Lord et al., 2004), display larger double support and single support times (Sacco & Amadio, 2000), and increase their step widths (Petrofsky et al., 2005).

Cardiovascular conditions such as hypertension and congestive heart failure have a detrimental impact upon elderly walking speed (Hausdorff et al., 2003; Sunnerhagen, Cider, Schaufelberger; Hedberg & Grimby, 1998) and gait stability (Hausdorff, Forman, Ladin, Goldberger, Rigney & Wei, 1994; Hausdorff et al., 2003) and may impact upon fall risk (Hausdorff et al., 2003). One study diagnosed five out of six fallers, who were without a known gait disorder, to have fallen due to a cardiovascular related cause (Montero-Odasso, Schapira, Duque, Soriano, Kaplan & Camera, 2005). This research indicates that when a gait

disorder is not diagnosed in an individual with a history of falls, cardiovascular causes should be considered (Montero-Odasso et al., 2005).

Osteoarthritis, particularly of the knee, is a disabling condition that results in diminishing gait speed (Kaufman, Hughes, Morrey, Morrey & An, 2001). This condition is characterised by joint symptoms, such as pain, and the erosion of the joint cartilage (Felson & Zhang, 1998). Women aged 50 years and over are 1.8 more times at risk than men to suffer from knee osteoarthritis (Felson et al., 1997). The reduced gait speed, characterised by shortened stride length and increased stance time, is believed to be a compensatory mechanism to reduce joint loading and pain (Kaufman et al., 2001) and increase joint stability (Al-Zahrani & Bakheit, 2002).

2.3.3 Fall History

Risk factors for falling, during everyday functional tasks such as walking, include a history of one or more falls (Gerdhem, Ringsberg, Åkesson & Obrant, 2005; Woo et al., 1995), stroke (Jørgensen, Engstad & Jacobsen, 2002), Parkinson's disease (Wielinski, Erickson-Davis, Wichmann, Walde-Douglas & Parashos, 2005), dementia (Allan et al., 2005; Bergland & Wyller, 2004), impaired gait (Gunter, White, Hayes & Snow, 2000; Lord et al., 1996; Woolley et al., 1997), muscle weakness (Lord et al., 1996; Moreland, Richardson, Goldsmith & Clase, 2004; Ozcan, Donat, Gelecek, Ozdirenc & Karadibak, 2005), psychotropic medications and multiple medications (Landi, Onder, Cesari, Barillaro, Russo, & Bernabei, 2005; Leipzig, Cumming & Tinetti, 1999a; Leipzig, Cumming & Tinetti, 1999b), and arthritis (Sturnieks, Tiedemann, Chapman, Munro, Murray & Lord, 2004).

Falls are a primary concern for older adults as they may lead to serious injury, loss of independence, and possible death. Injuries resulting from falls include bone fractures (representing the largest injury group), soft-tissue bruises and contusions, dislocations (Kannus et al., 1999), head trauma (Kannus et al., 1999; Nagurney, Borczuk & Thomas, 1998) and lacerations (Kannus et al., 1999).

Fallers demonstrate a slower walking velocity (Nelson et al., 1999; Woolley et al., 1997), a shortened stride length and smaller swing/stance ratio compared to non-fallers (Woolley et al., 1997). They also exhibit toeing out (forefoot abduction) accentuation (Woolley et al., 1997), have a longer stride time (Nelson et al., 1999; Woolley et al., 1997) and spend more time in the double support phase (Woolley et al., 1997). Multiple fallers have a reduced and more variable cadence and increased stance duration compared to non-multiple fallers (Lord et al., 1996). These differences are exacerbated by the fear of falling (Chamberlin, Fulwider, Sanders & Medeiros, 2005, Maki, 1997). Fallers who are fearful of falling, walk slower, have a shorter stride length, a wider stride width and longer double support time than fallers who do not fear falling (Chamberlin et al., 2005; Maki, 1997).

It is possible that the gait changes displayed by fallers are inherent mechanisms used to promote stability. A slower walking velocity may improve the ability to recover from a stumble or avoid obstacles, and the associated shorter stride length may reduce the centre of mass excursion beyond the base of support (Maki, 1997). The increased double support time may also improve stability by placing the centre of mass between the stance feet (Judge et al., 1996) and by

decreasing the amount of time spent standing on one leg (Maki, 1997). Alternatively, the reduced stride velocity and stride length may be due to deteriorated plantar flexor muscle torque (McGibbon & Krebs, 1999; Riley, DellaCroce & Kerrigan, 2001; Winter et al., 1990). The increased double support time may also simply be the consequence of the changes in these two parameters (Maki, 1997).

Stride width may (Maki, 1997) or may not (Heitmann, Gossman, Shaddeau & Jackson, 1989) be predictive of falling. These differing conclusions are possibly due to the different methodology between studies where fallers have been identified retrospectively (Heitmann et al., 1989) and prospectively (Maki, 1997), and inconsistent definitions of what constituted a fall were used. Although earlier researchers asserted that a wider stride width would also increase gait stability, an alternate view suggested that increases in stride width increase the possibility of falling (Maki, 1997). An increase in stride width may increase stability during the double support phase but may also reduce stability during single leg support (Maki, 1997). The lateral excursion that occurs during the wider stride requires greater effort to “recapture” the centre of mass as it moves laterally (Maki, 1997). Reduced stability during the single support phase is of major concern for “fall-risk” individuals. This period of the gait cycle is when they are already at their most unstable and vulnerable position due to the small base of support that is created by having only one foot in contact with the ground. The larger stride width may further promote instability by increasing lateral movement of the head, thereby reducing visual stability of the head (Maki, 1997).

Women are at a greater fall risk than men (Montero-Odasso et al., 2005; Morris et al., 2004; Pavol, Owings, Foley & Grabiner, 1999; Talbot, Musiol, Witham & Metter, 2005) and are more likely to suffer an injury from a fall (Talbot et al., 2005). A study that induced trips in both sexes reported that women were more likely to fall, compared to men who were more likely to recover and therefore prevent themselves from falling (Pavol et al., 1999). It may be hypothesised that the men had greater leg strength which allowed them to 'catch' themselves during the descent. Other possible mechanisms leading to falls include a fast walking speed, the body's position at the time of the fall and the inability to execute the recovery strategies after a trip (Pavol, Owings, Foley & Grabiner, 2001).

Interpretation of the falls literature can be somewhat ambiguous due to the inconsistent definitions used for a fall and this is highlighted by reference to table 2.2. A Tai-Chi intervention study has demonstrated this by using two different definitions (Wolf et al., 2003). The first definition of a fall was broad, requiring the individual to have unintentionally come to "rest on the ground, floor, or other lower level", while the other definition required the individual to have suffered some form of injury for the fall to be recorded. The broader definition recorded 209 falls compared to the narrower version which recorded only 110 falls. This particular study demonstrates that fall data may have a significantly different interpretation, dependent upon the fall definition that is used. Generally the studies agree that for a fall to occur it must be unintentional and for there to be contact with the ground with a body part other than the feet (Kannus et al., 1999; Maki 1997; Tinetti, Speechley & Ginter , 1988).

Table 2.2: Definitions of a fall from three studies.

Fall Criteria	Author(s)
(1) Hospitalised due to fall (2) Unexpected, sudden descent from an upright, sitting, or horizontal position (3) Descent height less than or equal to 1m (4) Emergency department visits not requiring hospitalisation not included	Kannus et al., (1999)
Any occasion on which they found themselves unintentionally on the floor, ground, or other lower surface, regardless of whether they sustained injury	Maki (1997)
Unintentionally coming to rest on the ground or at some lower level, not as a result of some major intrinsic event (e.g. stroke or syncope) or an overwhelming hazard that would result in a fall by most healthy young adults.	Tinetti et al., (1988)

2.3.4 Sensory Components of Gait

Three sensory systems interact to produce successful locomotion: visual, vestibular and somatosensory. The quality of information from these systems generally diminishes with age (Alexander, 1996; Baloh, Ying & Jacobson, 2003; Duncan et al., 1993; Lord et al., 1996; Menz, Lord & Fitzpatrick, 2003) and this has been linked with conservative walking patterns such as decreased stride velocity, stride length and cadence, and extended stance duration (Lord et al., 1996; Menz et al., 2003). These changes are most probably adaptive mechanisms to reduce the risk of falling (Menz et al., 2003).

The visual system guides the individual towards a desired endpoint and provides information about upcoming obstacles, allowing for proactive perturbation avoidance (Patla, 1997). Sub-optimal depth perception, visual acuity and contrast sensitivity are related to heightened fall risk (Lord & Dayhew, 2001; Lord et al., 1996; Woolley et al., 1997). The vestibular system acts as an internal reference, providing information on the heads orientation in space and acceleration (Brody, 1999). Vestibular defective individuals display slower walking speeds compared to healthy individuals due to shortened step lengths and reduced step frequencies (Borel et al., 2004). The somatosensory system includes the proprioceptive receptors that provide information about the orientation of the limbs and trunk in space. Proprioceptors include muscle spindles, golgi tendon organs, joint receptors (Ruffini endings, Pacinian Corpuscles) and cutaneous receptors (Brody, 1999; Riemann & Lephart, 2002). It has been postulated that individuals with poor proprioception lose the afferent feedback of where their feet are placed in space relative to the ground, and are therefore at greater risk of stumbling or falling (Sorock & Labiner, 1992).

2.3.5 Neuromuscular Changes

Overwhelming evidence exists for the relationship between ageing and the loss of muscular strength and power (Akima et al., 2001; Bassey, Fiatarone, O'Neill, Kelly, Evans & Lipsitz, 1992; Frontera, Hughes et al., 2000; Häkkinen, et al., 1996; Hughes et al., 2001; Izquierdo et al., 2001; Landers et al., 2001; Samson et al., 2000). Muscular strength is defined as the maximal force produced from a single maximum contraction and muscular power refers to the amount of work

performed with respect to time. Various methodologies and periods of time have been used to assess the loss of these muscular function parameters. One study demonstrated, as a percentage of muscle strength lost per decade, men and women are relatively similar, ranging between 11.1-16.7% per decade from 46 to 78 years of age (Hughes et al., 2001). The loss of muscular power is even greater than that of muscular strength (Bassey et al., 1992; Foldvari et al., 2000; Izquierdo et al., 2001; Lauretani et al., 2003; Macaluso & De Vito, 2003).

Sarcopenia, the age-related atrophy of skeletal muscle (Evans, 1995), plays an important role in the loss of muscle function (Akima et al., 2001; Frontera, Hughes et al., 2000; Häkkinen et al., 1996; Izquierdo et al., 2001; Landers et al., 2001; Lauretani et al., 2003; Petrella, Kim, Tuggle, Hall & Bamman, 2005). Muscle mass peaks during the third and fourth decades of life and begins to decline during the 4^{th} and 5^{th} decades (Lauretani et al., 2003; Lindle et al., 1997; Lynch et al., 1999). Although muscle loss occurs at a faster rate in men (Lauretani et al., 2003; Lynch et al., 1999), compared to women its impact is believed to be of a lesser consequence, as men have a larger peak muscle cross-sectional area than women and therefore have more muscle to lose over the lifespan (Lynch et al., 1999).

Strength and power differences between young and older females still exist after normalisation for cross-sectional area (Macaluso & De Vito, 2003). Sarcopenia, therefore, does not solely explain the loss of strength and power (Häkkinen et al., 1996; Häkkinen et al., 1998; Hughes et al., 2001; Macaluso & De Vito, 2003). Other explanations include reduced maximal voluntary neural input (Häkkinen et al., 1996; Häkkinen et al., 1998), increased agonist-antagonist co-activation

(Häkkinen et al., 1998), decreased muscle fibre capacity to generate force (Frontera, Suh, Krivickas, Hughes, Goldstein, Roubenoff, 2000) and altered muscle fibre expression (Andersen, Terzis & Kryger, 1999; Andersen, 2003). Altered muscle fibre expression has been used to explain the increased co-expression of MHC I (i.e. Type I – Oxidative) and MHC IIA (i.e. Type IIA – Oxidative Glycolytic) exhibited in the elderly (Andersen et al., 1999; Andersen, 2003). It was postulated that during this process, motor unit denervation occurs and is followed by reinnervation (Andersen et al., 1999; Andersen, 2003). Muscle fibres that were previously innervated by a different motor nerve subsequently receive different neural input, providing conflicting signals to their myogenic linage, and therefore changing their expression (Andersen et al., 1999; Andersen, 2003).

Sarcopenia alone appears to be an inconsistent predictor of mobility (Lauretani et al., 2003). Prediction of mobility should include the combination of measures of muscle function (i.e. strength and power) and muscle mass (Lauretani et al., 2003). Older adults with poor mobility have reduced strength and power of the ankle (Bendall et al., 1989; Duncan et al., 1993) and knee extensors (Bassey et al., 1992; Kwon et al., 2001; Samson et al., 2000; Lord et al., 1996). Although muscular strength and power are related attributes, muscular power has the greatest impact upon gait function (Bean, Leveille, Kiely, Bandinelli, Guralnik & Ferrucci, 2003).

Compared to older men, older female gait is more significantly impacted upon by the reduced strength and power (Bassey et al., 1992; Kwon et al., 2001; Lamoureux, Sparrow, Murphy & Newton, 2002). This is probably due to older

men having larger peak values of strength and power and therefore greater reserves of these parameters (Bassey et al., 1992; Kwon et al., 2001). Older women have been reported to have half the leg extensor power of older men (Bassey et al., 1992). This may explain their slower walking speed and greater prevalence of falls (Bassey et al., 1992; Morris et al., 2004; Pavol et al., 1999).

2.4 Exercise Interventions to Improve Gait Parameters

Muscular disuse and the 'normal' ageing process are the most likely causes of the neuromuscular changes in the healthy elderly (Aniansson, Sperling, Rundgren & Lehnberg, 1983; Hughes et al., 2001; Rantanen, Era & Heikinnen, 1997; Wiswell et al., 2001). It has been suggested that older adults should participate in exercise programs that increase muscular strength, and more specifically muscular power, since maintaining this variable in later life may reduce disability and maintain functional independence (Evans, 2000).

Physical activity has been demonstrated to improve several physiological attributes of older adults including VO_2 peak (Buchner et al., 1997; Cress, Buchner, Questad, Esselman, deLateur & Schwartz, 1999; Vaitkevicius et al., 2002), strength (Buchner et al., 1997; Cress et al., 1999; Fiatarone, Marks, Ryan, Meredith, Lipsitz & Evans, 1990; Heiwe et al., 2001; Lamoureux, Sparrow, Murphy & Newton, 2003; Rosario, Villani, Harris & Klein, 2003; Schlicht et al., 2001; Topp et al., 1996) and power (Jozsi, Campbell, Joseph, Davey & Evans, 1999). Physical activity is also considered to have a positive effect on gait parameters (Fiatarone et al., 1990; Hausdorff et al., 2001; Heiwe et al., 2000) and falls (Day, Fildes, Gordon, Fitzharris, Flamer & Lord, 2002).

Resistance training increases gait velocity in the healthy elderly (Fiatarone et al., 1990; Heiwe et al., 2001; Judge et al., 1993; Rubenstein et al., 2000; Schlicht et al., 2001; Topp et al., 1996) and in some chronic conditions such as uraemia (Heiwe et al., 2001). In one study, after eight weeks of high intensity resistance training, elderly participants ranging between 86-96 years of age improved their leg strength by an average of 174% ± 31% (Fiatarone et al., 1990). Tandem gait improved in these participants by 48% at the end of the intervention and two participants no longer used their canes to walk. This study demonstrated that: (1) Older adults can perform high-intensity resistance training without negative side-effects; (2) Older adults can improve their muscular strength from a high-intensity resistance training program; and (3) Improvements in strength from high-intensity resistance training can improve functional mobility.

Improvements in gait have not been found by all studies. One study, found protective effects for falls, but failed to demonstrate differences in gait following six months of resistance training (Buchner et al., 1997). Compared to the studies that found improvements in gait, this study failed to incorporate progressive overload (increasing the load on the muscle as it becomes habituated to the current load) and used relatively lower exercise volumes (sets and repetitions) and intensity. Older adults with “poorer” physiological function appear to be more sensitive to exercise intervention (Hausdorff et al., 2001). It is therefore possible that studies failing to demonstrate improvements for healthier older adults did so because of the “ceiling effect” (Hausdorff et al., 2001).

Several tests have been used to measure the potential benefits of resistance training on gait. A program incorporating strength and aerobic training increased

distance walked for a six-minute walk by an average of 48-metres (Rubenstein et al., 2000). The timed-up-and-go, a popular clinical method to assess functional status, has also demonstrated improvements after a period of resistance training of 13% and 23% in uraemic and healthy participants (Heiwe et al., 2001). It is difficult to identify whether the benefits have been the result of one or more training methods since most studies have used multiple training methods including resistance, aerobic and balance training.

Walking velocity is the primary dependent variable that has been measured to assess changes in gait following an intervention program (Heiwe et al., 2001; Judge et al., 1993; Rubenstein et al., 2000; Schlicht et al., 2001; Topp et al., 1996). This is a major limitation of gait intervention research as walking velocity only provides a global assessment of the changes in gait. Walking velocity fails to provide information about other mechanisms that may have contributed to the gait modifications (Lamoureux et al., 2003) such as stride length and stride time. The only study to have examined the kinematic gait changes of older adults following a resistance training program found increases in stride length and decreases in stride time to contribute to reduced obstacle crossing time. This research, however, only looked at the changes in obstructed gait kinematics (Lamoureux et al., 2003). There has been no research on level surface gait examining the effect of resistance training on spatio-temporal parameters such as stride length and stride time in the healthy elderly. It is therefore unclear which aspects of the gait cycle impact upon the improved walking velocity following resistance training.

Summary of Key Points

Numerous age-related changes and conditions may impact upon ageing gait. The mechanisms leading to gait impairment are multifactorial and include age–related changes in sensory and neuromuscular function as well as declining cognitive status, neuromuscular and cardiovascular disease, and fear of falling. Women are more susceptible to impaired neuromuscular function, which potentially has a larger impact upon their mobility and fall risk. Although not equivocal, interventions such as resistance training have been demonstrated to improve the gait function of older adults.

3.0 Whole Body Vibration

Vibration is often considered an occupational health hazard as it has been associated with nerve tissue damage (Dahlin, Necking, Lundström, & Lundborg, 1992), muscle tissue damage (Necking, Dahlin, Fridén, Lundborg, Lundström & Thornell, 1992), lower back disorders (Bovenzi & Hulshof, 1999), Raynaud's Disease (vibration white finger) (Herrick, 2005) and interference with cognitive processes such as that required for short-term memory (Sherwood & Griffin, 1990). Although vibration may produce undesirable side-effects, physiotherapists have been reported to use this modality as a therapeutic intervention such as for clearing the lungs and improving joint mobility (Griffin, 1994). More recently, vibration has been shown to have positive impacts upon the bone density of post-menopausal women and disabled children (Rubin, Recker, Cullen, Ryaby, McCabe & McLeod, 2004; Verschueren, Roelants, Delecluse, Swinnen, Vanderschueren & Boonen, 2004; Ward, Alsop, Caulton, Rubin, Adams & Mughal, 2004), back pain (Rittweger, Just, Kautzsch, Reeg & Felsenberg, 2002), stroke (van Nes, Geurts, Hendricks & Duysens, 2004), multiple sclerosis (Schuhfried, Mittermaier, Jovanovic, Pieber & Paternostro-Sluga, 2005) and muscle spasticity of cerebral palsy sufferers (Ahlborg, Andersson & Julin, 2006).

The potential benefits for human performance from vibration were originally reported by Nazarov & Spivak (1987). Russian gymnasts performed exhaustive gymnastic exercise on vibrating gymnastic rings. At the completion of the vibration exposure period, the gymnast's shoulders were 2-3 times stronger than the control group, who performed the same exercises without vibration. It was suggested by Nazarov & Spivak (1987) that vibration could increase strength 5-6 times faster than traditional methods. Further scientific inquiry has since

investigated and demonstrated improvements in isotonic strength (Issurin, Liebermann & Tenenbaum, 1994), explosive power (Bosco, Cardinale, Tsarpela, Colli, Tihanyi, von Duvillard et al., 1998; Bosco, Cardinale & Tsarpela, 1999; Bosco, Colli et al., 1999; Issurin & Tenenbaum, 1999), balance (Torvinen et al., 2002b) and flexibility (Issurin et al., 1994).

3.1 Devices Used to Produce Vibration Stimulation

Several methods of vibration stimulation have since been reported within the literature. Whole body vibration (WBV) has been the most commonly used method of vibration for enhancing human performance. For this type of vibration stimulation, the individual stands on an oscillating plate. At present, two methods of whole body vibration stimulation are commercially available: (1) tilting oscillation; and (2) vertical oscillation (Figure 3.1).

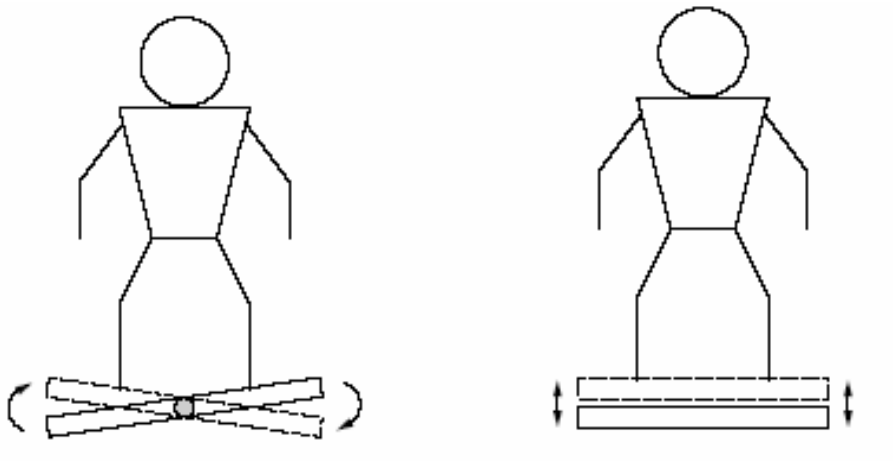


Figure 3.1: Two methods of whole body vibration stimulation: tilting oscillation (left) and vertical oscillation (right) (Cardinale & Wakeling, 2005).

Also, pulley systems (Figure 3.2) such as those typically seen in gymnasiums have been modified with a vibratory apparatus to oscillate the cable, producing

localised vibrations to the body parts contacting the vibrating device (Bosco, Cardinale et al., 1999; Issurin et al., 1994; Issurin et al., 1999).

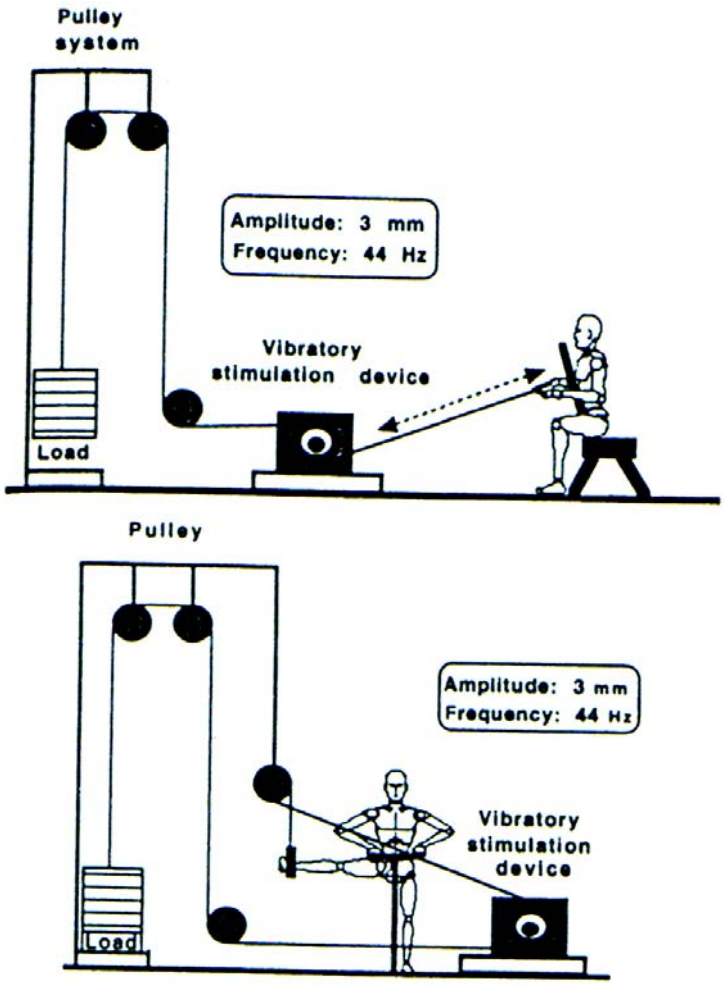


Figure 3.2: Apparatus employed by Issurin et al., (1994) for strength (top) and flexibility (bottom) exercises.

3.2 Theories explaining the human response to Whole Body Vibration

The exact mechanisms explaining the acute and chronic human response to WBV are currently unclear. The majority of theory has been indirectly based upon research that applied vibration onto the muscle belly, single muscle fibres and/or receptor organs (e.g. muscle spindles) (e.g. Burke, Hagbarth,

Löfstedt & Wallin, 1976a; Burke, Hagbarth, Löfstedt & Wallin, 1976b; Eklund & Hagbarth, 1966). The theories explaining the human response to WBV should therefore be considered cautiously, as WBV is a different stimulus compared to that used in most of these studies. According to Griffin (1994, p1), the "shaking of the human body – a complex, active, intelligent, dynamic structure – should not be expected to have a single, simple, or easily predictable consequence". As this statement implies, the whole body is vibrated and therefore several structures and mechanisms may be stimulated. Current technology is insufficient to identify with certainty which mechanism(s) contribute to the observed neuromuscular performance improvements. The following sections outline the current explanations for why human functional performance may improve following WBV.

3.2.1 Tonic Vibration Reflex

The application of mechanical vibration to the muscle or tendon elicits a reflexive concentric contraction and the reciprocal inhibition of its antagonists (Eklund & Hagbarth, 1966; Hagbarth & Eklund, 1966). This response is termed the Tonic Vibration Reflex (Hagbarth & Eklund, 1966) and is closely related to the stretch reflex (Figure 3.3).

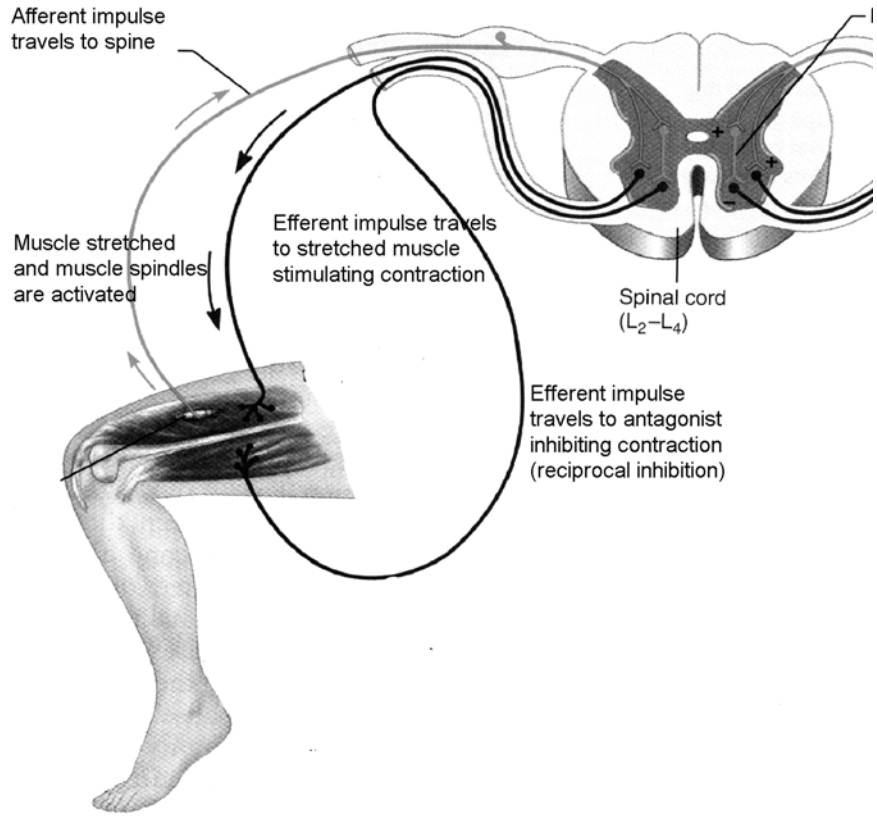


Figure 3.3: Illustration of the stretch reflex (Adapted from Marieb, 2004).

The mechanical action of vibrations produces short fast deformation of the muscle and tendon, stimulating the primary Ia afferents of muscle spindles (Burke et al., 1976a; Burke et al., 1976b). Secondary muscle spindle endings and Golgi tendon organs also respond to vibration (Burke et al., 1976a; Burke et al., 1976b), but these receptors are not as responsive to vibration as the primary afferent endings (Brown, Engberg and Matthews, 1967; Trott, 1976). Upon stimulation of the primary Ia afferents, afferent neurons transmit impulses to the spinal cord. From the spinal cord the sensory neurons synapse with the alpha motor neuron of the stretched muscle, exciting the muscle's extrafusal fibres producing a contraction. At the same time, the afferent neurons also connect

with inhibitory motor neurons of the antagonist muscle inhibiting contraction of this muscle.

Vibration stimulates both monosynaptic and polysynaptic reflex pathways (Burke & Schiller, 1976). Monosynaptic reflex pathways (Figure 3.4) are the simplest reflex pathways consisting of an afferent neuron, one synapse, and an efferent neuron (Latash, 1998).

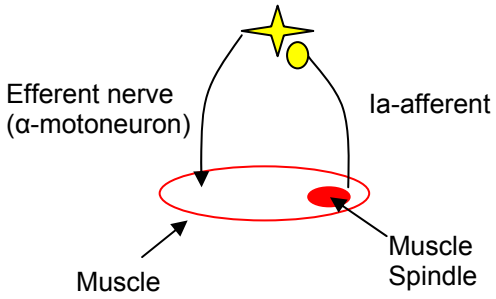


Figure 3.4: Diagram of a monosynaptic reflex (Adapted from Latash, 1998).

Polysynaptic reflex pathways contain numerous (more than three) synapses (Latash, 1998). Although the majority of muscle reflexes are believed to be polysynaptic they are still poorly understood (Latash, 1998). Figure 3.5 illustrates the theoretical understanding of the polysynaptic reflex in the context of vibration stimulation. Vibration stimulates multiple peripheral receptors including muscles spindles and golgi tendon organs, resulting in muscular contraction (Latash, 1998).

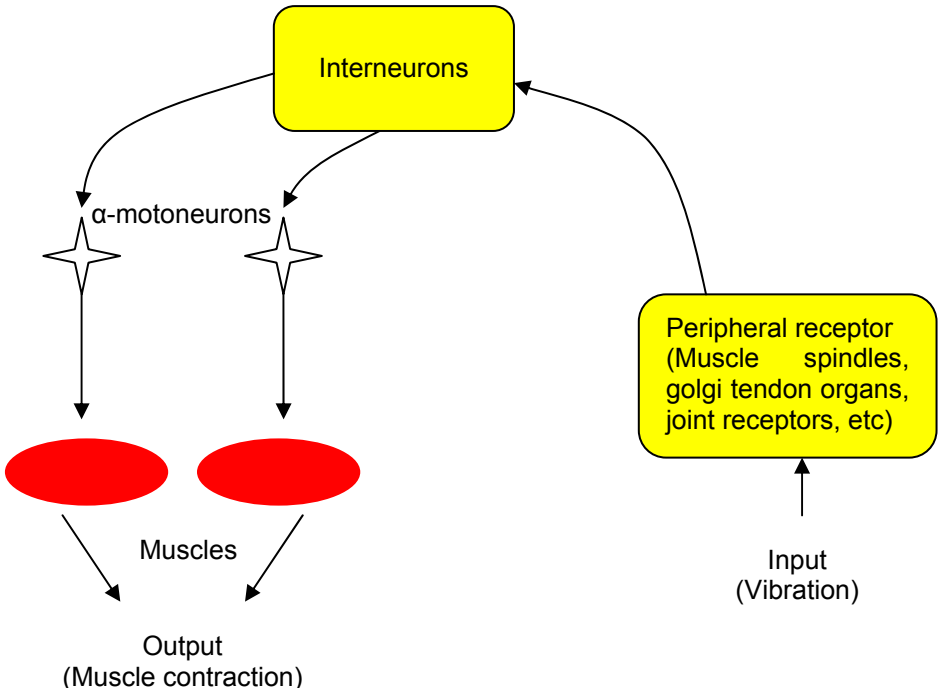


Figure 3.5: Illustration of the polysynaptic reflex pathways during vibration stimulation (Adapted from Latash, 1998).

Muscular contractions of vibrated muscle increase during WBV (Cardinale & Lim, 2003). Although vibration is not placed directly upon the muscle or tendon during WBV, as required by the definition of the Tonic Vibration Reflex, the increased muscle activity that occurs during WBV is explained by this reflexive response. When vibrations are superimposed over an isotonic, isometric or isokinetic contraction, a more forceful muscular contraction is produced (Warman, Humphries & Purton, 2002). It has been suggested that the acute enhancement following vibration is due to enhanced muscle receptor sensitivity following activation of the stretch reflex (Cardinale & Bosco, 2003; Issurin & Tenenbaum, 1999). A warming effect produced by the friction of vibrating tissues and increased blood flow have also been suggested to contribute to acute performance enhancement (Issurin & Tenenbaum, 1999).

It has been proposed that the improvements in strength and power experienced after WBV may be comparable with resistance training (Delecluse, Roelants & Verschueren, 2003). Resistance training and WBV both place increased gravitational load on the neuromuscular system (Cardinale & Bosco, 2003). Gravitational loading has the capacity to modify the functional capacity of the neuromuscular system through neurogenic and myogenic factors (Carroll, Riek & Carson, 2001). The initial gains in strength during the early stages of resistance training are attributed to neurogenic factors (Kraemer et al., 2002; Sale, 1988; Staron et al., 1994) and although protein synthesis (i.e. the process of hypertrophy) may occur after just one resistance training session (Chesley, MacDougall, Tarnopolsky, Atkinson, & Smith, 1992), it is not until after approximately 6 weeks that muscular hypertrophy becomes evident (Kraemer et al., 2002).

Several of the mechanisms contributing to neural adaptations after resistance training are also attributed to the changes that occur following WBV. They include changes in: (1) maximal voluntary activation; (2) co-activation; (3) motor unit synchronisation; and (4) central motor command activity.

3.2.2 Maximal Voluntary Contraction

Humans are often unable to recruit all motor units during a maximal voluntary contraction (Dowling, Konert, Ljucovic & Andrews, 1994; Gandevia, 2001; Knight & Kamen, 2001). Improving the number of motor units recruited will therefore increase strength capacity (Enoka, 1997). Superimposing vibration on isotonic, isometric and isokinetic maximal voluntary contraction's simultaneously elevates force output (Warman et al., 2002) and muscular activity of the vibrated muscle

(Warman et al., 2002). Following vibration, elevated maximal voluntary contractions have been shown to continue only for isotonic contractions (Warman et al., 2002). The mechanism for these improvements is unclear as EMG activity has been shown to return toward resting levels, despite elevated contraction forces (Warman et al., 2002). This has also been demonstrated during explosive extension movements, where leg extensor power increased and muscle activity decreased (Bosco et al., 2000). The improvements in maximal voluntary contraction post vibration have been interpreted as due to improved neuromuscular efficiency (Warman et al., 2002). It may also be due to elevated muscle temperature produced by the vibration (Warman et al., 2002).

3.2.3 Motor Unit Synchronisation

Motor unit synchronisation refers to increasing muscle force output as a result of the synchronous discharge of motor units (Enoka, 1997). A benefit of motor unit synchronisation is an increased rate of force development (Semmler, 2002), a characteristic that is essential for maximising explosive movements such as the vertical jump. Although evidence for motor unit synchronisation is inconclusive, it is a commonly cited explanation for increased muscle force output following resistance training (Enoka, 1997; Gabriel, Kamen & Frost, 2006; Semmler, 2002). Enhanced motor unit synchronisation has also been cited as a possible explanation for strength and power increases following WBV (Cardinale & Bosco, 2003; Cardinale & Lim, 2003). Increased surface EMG activity during vibration stimulation of the arms with vibrating dumbbells (Bosco, Cardinale & Tsarpela, 1999) has been presented as possible evidence for improved motor unit synchronisation (Cardinale & Bosco, 2003). Surface EMG, however, has been

demonstrated to be a poor indicator of motor synchronisation (Yue, Fuglevand, Nordstrom & Enoka, 1995). Therefore, further research is needed to determine whether in fact improved motor unit synchronisation contributes to enhanced force output.

3.2.4 Co-Activation

Placing a vibration device on the bicep tendon promotes involuntary flexion of the elbow (Hagbarth & Eklund, 1966). An inhibitory effect on the antagonistic muscles (i.e. reciprocal inhibition) is also observed preventing elbow extension (Hagbarth & Eklund, 1966). It has been suggested that this effect may improve intramuscular co-ordination and therefore performance by reducing the braking forces produced by co-activation of the joint being vibrated (Cardinale & Bosco, 2003). It should be noted that the vibration research examining co-activation has been on single muscle groups only (Hagbarth & Eklund, 1966). To date there have been no studies examining the changes in agonist and antagonist activity following WBV.

3.2.5 Central Motor Command

It has been suggested that the motor cortex should also be considered as a possible site stimulated by WBV, and subsequently a contributor to performance enhancement (Cardinale & Bosco, 2003). The sensitivity of the corticospinal pathway to muscle tendon vibration suggests that this is a possibility (Steyvers, Levin, Verschueren & Swinnen, 2003). The optimal frequency for muscle-tendon vibration to increase corticospinal excitability has been approximated as 75 Hz for the flexor carpi radialis, while no observed

changes have been found at 20 Hz (Steyvers et al., 2003). This would suggest that the frequency typically used in WBV research would not stimulate the corticospinal pathway. However, conclusions such as this should be made cautiously as only one muscle was stimulated by Steyvers et al., (2003). It is possible that WBV at the lower frequencies stimulates the corticospinal pathway via the stimulation of multiple muscle groups. Further research is required in this area before generalisations can be made.

3.2.6 Hormonal Response

The endocrine system has an important role in muscle protein synthesis following resistance training (Crewther, Keogh, Cronin & Cook, 2006). Testosterone and growth hormone have an anabolic effect upon muscle tissue, while cortisol has a catabolic effect (Crewther et al., 2006). These hormones may also have an effect upon neural pathways (Crewther et al., 2006). The isolation and stimulation of muscle spindle afferents by vibration has increased blood growth hormone concentrations (McCall, Grindeland, Roy & Edgerton, 2000). Immediately following WBV, blood testosterone (Bosco et al., 2000) and growth hormone (Bosco et al., 2000; Kvorning, Bagger, Caserotti & Madsen, 2006) concentrations have been shown to increase and cortisol levels to decrease (Bosco et al., 2000; Kvorning et al., 2006) but these results have not been consistently demonstrated. One study found increased blood growth hormone and lowered cortisol concentrations but failed to demonstrate increases of blood testosterone (Kvorning et al., 2006). Moreover, the authors of this study highlighted the difference in protocols where the earlier work used 10 x 1 minute efforts with 1 minute

recoveries, compared to their study which used 6 x 30 second efforts with 2 minute recovery (Kvorning et al., 2006). The shorter WBV time and the longer recovery times used by the latter study may be the variables that created these different results (Kvorning et al., 2006). Based upon experiments with birds (Bleisch, Luine & Nottebohm, 1984), it has been suggested that the elevated testosterone levels may contribute to the acute enhancement of strength and power (Bosco et al., 2000). Consequently, further research must be performed to understand the impact WBV has on neuromuscular performance.

3.3 Whole Body Vibration Nomenclature

The review of literature has shown that there is a lack of consistency and clarity with the parameters being reported for WBV, making it difficult to replicate and further advance the published investigations. There are two major discrepancies in the manner that WBV experimentation is currently being reported. Firstly, uniform terminology is not being used for reporting of the magnitude of vibration; and secondly, the method to calculate maximal acceleration is inconsistent.

3.3.1 Magnitude of Vibration

Although one study reported only one vibration parameter (frequency) which therefore made it impossible to replicate their research (Rønnestad, 2004), most studies have typically provided the reader with information consisting of a combination of frequency, amplitude, displacement, peak-to-peak displacement, acceleration and/or gravitational acceleration (Table 3.1, p. 53). Vibration frequency (Hz) appears to be the most clearly and accurately reported of these

parameters and is measured by the number of cycles of oscillations per second (Griffin, 1994). However, there is less consistency in reporting the magnitude of the vibratory oscillations. Past work for example, has generally described this parameter by reference to amplitude, displacement or peak-to-peak displacement (Table 3.1). Essentially, this parameter has lacked consistency and scientific rigor to enable other researchers to reliably compare studies.

Practices in reporting amplitude, displacement and peak-to-peak displacement vary so widely that it is imperative some guidelines are established in an attempt to gain consistency across studies. In the case of vibration, amplitude is the maximum displacement of a vibrating point from a mean position (Figure 3.6). Another term, peak-to-peak displacement is used to describe the total vibration excursion of a point between its positive and negative extremes (Figure 3.6).

Table 3.1: Parameters reported by studies investigating whole body vibration

Author(s)	Vibration Parameters and Terminology Reported
Bosco et al., (1998)	Frequency (Hz), Displacement (mm), Acceleration ($m.s^{-2}$)
Bosco, Colli et al., (1999)	Frequency (Hz), Displacement (mm), Acceleration ($m.s^{-2}$)
Bosco et al., (2000)	Frequency (Hz), Displacement (mm), Acceleration (g)
Cardinale & Lim (2003)	Frequency (Hz), Amplitude (mm)$^{\alpha}$
Cronin et al., (2004)	Frequency (Hz), Amplitude (mm), Displacement (mm), Acceleration (g)*
Delecluse et al., (2003)	Frequency (Hz), Amplitude (mm), Peak-to-Peak Amplitude (mm), Acceleration (g)$^{\#}$
Delecluse et al., (2005)	Frequency (Hz), Amplitude (mm), Acceleration (g)
De Ruiter, van Raak et al., (2003)	Frequency (Hz), Amplitude (mm)
Rittweger et al., (2000)	Frequency (Hz), Amplitude (cm), Peak Acceleration ($m.s^{-2}$ and g)
Rittweger et al., (2001)	Frequency (Hz), Amplitude (mm)$^{\wedge}$
Rittweger, Ehrig et al., (2002)	Frequency (Hz), Amplitude (mm)
Rittweger et al., (2003)	Frequency (Hz), Amplitude (mm)
Roelants et al., (2004)	Frequency (Hz), Amplitude (mm), Acceleration (g)
Rønnestad (2004)	Frequency (Hz)
Runge et al., (2000)	Frequency (Hz), 7-14 mm thrusts upwards
Russo et al., (2003)	Frequency (Hz); Acceleration (g)
Torvinen et al., (2002)	Frequency (Hz), Peak-to-peak amplitude (mm)
Torvinen et al., (2002a)	Frequency (Hz), Peak-to-peak amplitude (mm), Maximal Acceleration (g)
Torvinen et al., (2002b)	Frequency (Hz), Peak-to-peak amplitude (mm), Maximal acceleration (g)$^{\#\#}$
Torvinen et al., (2003)	Frequency (Hz), Peak-to-peak amplitude (mm), Acceleration (g)

$^{\alpha}$ Defined amplitude as peak-to-peak

* Amplitude and displacement were used synonymously

$^{\#}$ Amplitude and peak-to-peak amplitude were used synonymously

$^{\wedge}$ This paper reported how to calculate acceleration but did not provide the actual acceleration for this study

$^{\#\#}$ Amplitude was measured at the end of the tilting platform

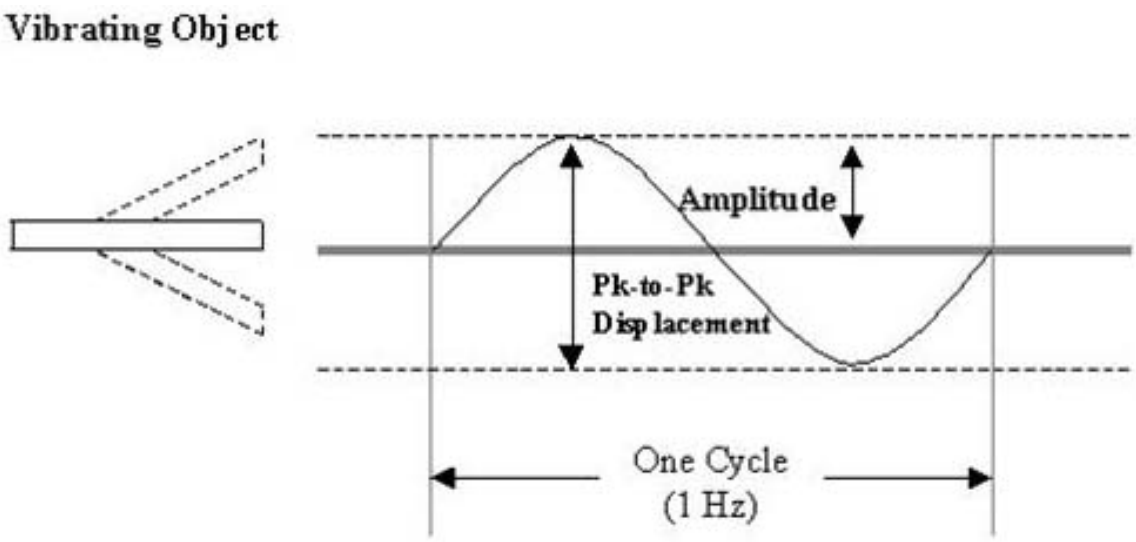


Figure 3.6: Graphical diagram demonstrating frequency, amplitude and peak-to-peak displacement of a vibrating object

How the researcher measured amplitude is also of concern. Figure 3.7 demonstrates that the further each point of the plate is from the axis, the greater the amplitude. Although amplitude has been reported to be measured at the end of the tilting platform (Torvinen et al., 2002b), the majority of studies have not reported at which point from the axis the amplitude was measured. Since the size of the platform and therefore maximal tilting amplitude may differ between devices, a more relevant measurement would be the placement of the feet (e.g. middle toe) from the centre of axis of rotation.

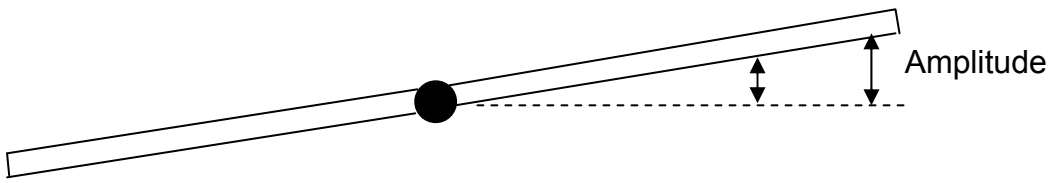


Figure 3.7: Any fixed point on the plate undergoes the same rotation, however each point undergoes a different linear displacement based upon the distance from the axis.

Disconcertingly, several studies have not used the standard definitions of amplitude. Amplitude and peak-to-peak displacement have been employed synonymously by some authors (Cardinale & Lim, 2003; Delecleuse et al., 2003) when by definition they are not the same. It appears that one study (Rittweger, Beller & Felsenberg, 2000) incorrectly reported that they used an amplitude of 1.05 cm (10.5 mm). A frequency of 26 Hz and maximal acceleration of 147 $m.s^{-2}$ were also reported, providing the opportunity to recalculate amplitude (maximal acceleration is the peak rate of change of velocity of the vibrating plate). Calculation of amplitude (A), using the provided frequency (f) and maximal acceleration (a_{max}) demonstrates that there is a major discrepancy in what was reported for frequency, amplitude and acceleration. The calculations below indicate an amplitude of 5.5mm in comparison to the 10.5mm reported. Although it is most likely that the authors reported peak-to-peak amplitude rather than amplitude, this does not explain a 0.25 mm discrepancy for amplitude. Assuming vibration frequency is maintained at 26 Hz, further calculations suggest there to be a 6.7 $m.s^{-2}$ (0.7 g) discrepancy between what is reported and the figures that have been provided (See Appendices K-R for tables displaying calculated plate accelerations for various frequencies and amplitudes) .

$$A = \frac{a_{max}}{(2\pi f)^2}$$ **(Formula 1, Griffin, 1994)**

$$A = \frac{147}{(2.\pi.26)^2}$$

$$A = 5.5 \text{ mm.}$$

It is therefore important that vibration variables are reported accurately as erroneously applied values significantly change the acceleratory forces. Table 3.2 demonstrates the acceleration maximum with various amplitudes and frequencies. By maintaining the same frequency and doubling the amplitude the accelerative forces will have increased twofold. Future research attempting to replicate these previous studies, and who are unaware of such discrepancies, may inflict injury to participants with potentially toxic vibration intensities.

Table 3.2: Various combinations of amplitude and frequency produce different acceleration maximum values.

Amplitude (mm)	Frequency (Hz)	Acceleration max ($m.s^{-2}$)
1	15	8.9
1	30	35.5
2	15	17.8
2	30	71.1

3.3.2 Acceleration

Calculations and reporting of acceleration and gravitational acceleration show a similar range of inconsistency as that reported for amplitude. Of the 20 WBV papers presented in table 3.1, 12 studies reported acceleration. Only three of these studies reported this parameter as maximum (peak) acceleration. Maximum acceleration ($a_{\max}$) is dependant upon the frequency (f) and amplitude (A) of the vibrating plate and can be calculated as:

$$a_{\max} = A(2\pi f)^2$$ **(Formula 2)**

Therefore, at an amplitude of 2 mm and a frequency of 15 Hz, a_{max} = 17.8 m.s^{-2}. For the purpose of clarity, maximal acceleration, amplitude and frequency should be the minimum parameters reported by all studies. By providing these variables, future studies may confidently replicate the specifications of previous research.

Formula 2 demonstrates maximal acceleration to be the product of amplitude and frequency. The resulting maximal acceleration will therefore depend upon these two variables. Studies have used a wide variety of amplitudes and frequencies with various combinations producing an array of accelerations (Table 3.3). Reporting the maximal acceleration will clearly demonstrate to the reader the vibration intensities that are being applied to the participants, allowing comparisons to be more easily made.

Of the studies to have reported maximum acceleration, there appears to be disagreement with the methods used to calculate this parameter. Bosco et al., (1999), implemented a frequency of 26 Hz and a displacement of 10 mm (amplitude = 5 mm), and reported acceleration to be 54 m.s^{-2}. However, when using formula 2, the acceleration at this frequency and amplitude equates to 133.5 m.s^{-2}. It is difficult to provide an explanation for this discrepancy as the authors did not indicate how they determined acceleration.

Table 3.3: Examples of frequencies, amplitudes and the resulting maximal accelerations utilised by whole body vibration studies

Authors	Frequency (Hz)	Amp. (mm)	Acceleration ($m.s^{-2}$)
Bosco, Colli et al., (1999)	26	5	133.5
Bosco et al., (2000)	26	2	53.4
Cardinale & Lim (2003)	30, 40, 50	5	177.7, 315.9, 493.6
De Ruiter, van Raak et al., (2003)	30	8	284.3
Rittweger et al., (2001)	26	6	160.2
Rittweger et al., (2002)	18, 26, 34	5	64, 133.5, 228.2
Rittweger et al., (2003)	26	6	160.2
Torvinen et al., (2002a)	25, 30, 35, 40	1	24.7, 35.5, 48.4, 63.2
Torvinen, Sievanen et al., (2002)	25, 30, 35, 40	1	24.7, 35.5, 48.4, 63.2
Torvinen et al., (2003)	25, 30, 35, 40, 45	1	24.7, 35.5, 48.4, 63.2, 80.0

Furthermore, Bosco et al., (2000) has reported accelerations as high as 17 g (g = gravitational force). Gravitational force can be calculated by dividing $a_{\max}$ by gravity (9.81 $m.s^{-2}$) :

$$g = \frac{A(2\pi f)^2}{9.81} \quad \textbf{(Formula 3)}$$

Their paper reported a frequency of 26 Hz and an amplitude of 2 mm. Using the above formula, the g-forces equal 5.4 g. The calculations again disagree with the figures reported within the literature.

3.3.3 Whole Body Vibration Nomenclature Recommendation

For the purpose of clarity all published papers need to use a standard terminology to report whole body vibration intensity. Amplitude should be reported as the maximum displacement from the plate's horizontal position, and

should be measured by researchers from an anatomical landmark of the foot such as the middle toe. Furthermore, all studies should report the amplitude (mm), frequency (Hz) and acceleration max ($m.s^{-2}$). As is evident in table 3.2, manipulation of various frequencies and amplitudes can significantly affect acceleratory forces. Acceleration max or gravity more accurately indicate the forces being imposed on the human body during whole body vibration and should therefore be reported by all studies.

3.4 Application of Frequency & Amplitude

Although it has been suggested that WBV intensities approaching, or greater than 1 g may be 'toxic' (Rubin, Pope, Fritton, Magnusson, Hansson & McLeod, 2003; Rubin et al., 2004), the vast majority of WBV research examining its effect on neuromuscular performance has far exceeded these values. In the healthy young, the frequencies and amplitudes reported have ranged between 15-50 Hz and 1-8 mm respectively. For older adults the frequencies and amplitudes have ranged between 10-40 Hz and 1.5-5 mm respectively.

Unlike resistance training, where there are generally recognised progressions of exercise intensity based upon individual training status and goals (Kraemer et al., 2002), there are no such guidelines for WBV. This explains the numerous combinations that have been used by researchers. In some cases, the choice of intensity may have been prompted by the WBV device used. Commercial devices, such as the PowerPlate, limit the user to frequencies of 30, 35, 40 and 50 Hz, and amplitudes of 2.5 and 5 mm, producing accelerations ranging between 88.8 $m.s^{-2}$ to 493 $m.s^{-2}$. Other research appears to use the frequencies

and amplitudes used by previous research. Further research should identify the most appropriate frequencies and amplitudes based upon the individual's health and fitness status.

A unique method to identify the appropriate WBV intensity is to monitor EMG activity during the vibration stimulus (Cardinale & Lim, 2003). For the vastus lateralis, EMG activity has been demonstrated to be greatest when standing in a half squat position (knee angle at 100°) at a frequency of 30 Hz (Cardinale & Lim, 2003). Further studies are warranted to investigate specific muscle responses to vibration.

3.5 Health Risks of Whole Body Vibration

Vibration has long had negative associations with human health. As early as the 17th Century it was recognised that vibration produced by horse coaches may be the cause of the back pain being experienced by coachmen (Mester, Spitzenfeil, Schwarzer & Seifriz, 1999). In a modern context, vibration is considered to be an occupational risk factor producing adverse effects in drivers of trucks, fork-lift trucks, tractors, cranes, helicopter pilots (Bovenzi & Hulshof, 1999) and individuals who operate hand held machinery (Bovenzi, Lindsell & Griffin, 2000; Strömberg, Dahlin, Brun & Lundborg, 1997). For vibration exposure of less than 30 minutes, ISO 2631 establishes the discomfort level for vibration at 0.4 g at 30 Hz. Criticism has been directed to the vibration exposure approaching and exceeding 1 g suggesting such intensities should be avoided due to the risk of pathological responses (Rubin et al., 2003; Rubin et al., 2004). Neuromuscular

whole body vibration research has typically used accelerative forces far exceeding these loads.

Although some minor effects have been reported, the growing literature on WBV as an exercise intervention, has documented few pathological side effects. The most common side-effects are erythema and oedema of the lower limbs (Crewther, Cronin & Keogh, 2004; Rittweger et al., 2000). It is possible that this response is caused by the vasodilation that has been reported to occur in the lower limbs (Kerschan-Schindl et al., 2001; Rittweger et al., 2000).

More significant adverse effects have also been reported including hip (Crewther et al., 2004) and knee pain (Roelants et al., 2004; Russo et al., 2003), and discomfort from the vibration induced head motion (Crewther et al., 2004). One study (Cronin, Oliver & McNair, 2004) reported participants to complain of pain in the jaw, neck, and the lower extremity (particularly in the tibialis posterior). The pain required physiotherapy treatment in some participants and took seven to ten days to subside. In the healthy untrained young, the severe adverse effects are probably due to the excessive g-forces placed on these untrained participants. The studies to have reported such adverse effects vibrated participants at amplitudes as high as 5-6 mm and frequencies of 30 Hz, with accelerations of 177.7 $m.s^{-2}$ to 213.2 $m.s^{-2}$ (Crewther et al., 2004; Cronin et al., 2004). Although these two studies were not the first to use these intensities with untrained individuals, it is possible that the accelerations were too high for this group. Although it was identified that the adverse effects generally occurred when the frequency was at 30 Hz, it was concluded that frequency and amplitude

combinations that produce large g-forces should be cautiously applied, especially to individuals with a low training status (Crewther et al., 2004).

Another study reported two out of six subjects to have ill effects at specific frequencies (Rubin et al., 2003). One individual experienced faintness at 27 Hz and another individual complained of feeling “sea-sickness” at 17 Hz. The symptoms passed after each of the individuals laid down. This study surgically inserted pins into the greater trochanter and L_4 of the lumbar vertebra. Local anaesthetic was used for this process. It is possible that direct relationship between the WBV and side-effects may have been confounded by the anaesthesia and/or the surgical procedure. The participants may not have had these side-effects without the medical intervention.

In older adults, the most typical side effect has been transient itching, tingling and erythema of the feet and lower legs (Bruyere et al., 2005; Russo et al., 2003). There have been single reports of back (Roelants et al., 2004) and groin (Bautmans, Van Hees, Lemper & Mets, 2005) pain that has precluded participants from continuing. Knee pain has also been reported in participants with pre-existing osteoarthritis (Roelants et al., 2004; Russo et al., 2003), subsiding after several days of rest (Russo et al., 2003). There has been one case where an institutionalised elderly person refused to continue due to an apparent fear of the room the WBV device was in (Bautmans et al., 2005).

3.5.1 Human Resonance

When considering the adverse effects of WBV, human resonance should be considered. Figure 3.8 demonstrates the concept of resonant frequency. A body, in this case a pendulum, which has an external force exerted upon it by a body with the same frequency, will experience an increase in its amplitude. Maximal external force transmissibility occurs when the frequency of external force is phasic with the pendulum's frequency.

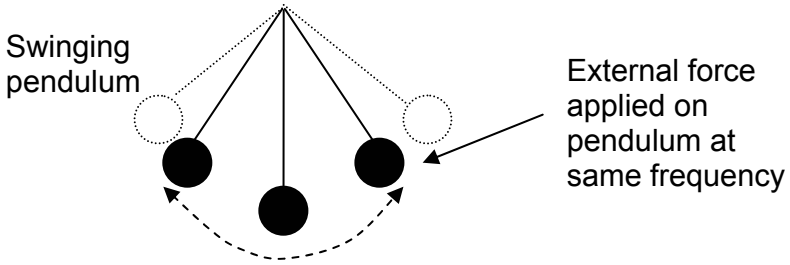


Figure 3.8: Resonant frequency of a pendulum. Broken line indicates increased amplitude as external force exerts maximal transmissibility at same frequency of pendulum.

The complexity of the human structure together with individual human differences means that there is not one all encompassing resonant frequency (Randall, Matthews & Stiles, 1997). Studies have reported varying resonant frequencies during standing posture ranging between 5.5 to 15.7 Hz (Matsumoto & Griffin, 1998; Randall et al., 1997). Individuals may have one main resonant frequency (Harazin & Grzeik, 1998) but may also have resonant peaks at other frequencies (Matsumoto & Griffin, 1998). Height, weight and gender have been demonstrated to bear no relationship with resonance (Randall et al., 1997).

Exposure to frequencies at human resonance increases the transmissibility of vibration, increasing the forces transmitted through some body parts and potentially causing a pathogenic response. In the erect and relaxed standing postures, transmissibility at the hip has been demonstrated to exceed 100% at frequencies lower than 20 Hz (Rubin et al., 2003). Research with the elderly has been performed within the reported ranges of human resonance, without causing detrimental effects (Bruyere et al., 2005).

3.5.2 The Muscle Tuning Hypothesis

The muscle tuning hypothesis is a relatively new concept that has been introduced to the WBV literature (Cardinale & Wakeling, 2005). This hypothesis suggests that the body has a strategy to reduce the impact of vibration (Figure 3.9). During heel-toe running, the force impact produces vibration frequencies ranging from 10-20 Hz acting on the foot (Nigg & Wakeling, 2001). These vibrations travel throughout the individual's body and are received by the sensory receptors (e.g. muscle spindles) of the lower limbs and transmitted to the central nervous system (Nigg & Wakeling, 2001). These frequencies that are produced from the impact of the foot on the ground (10-20 Hz) are within the range of resonant frequencies (5 to 65 Hz) of the major muscle groups of the lower limb (Nigg & Wakeling, 2001). It is therefore plausible to suggest that resonance may occur to the musculature and cause injury. Muscle tuning proposes that skeletal muscle alters its activity to dampen the vibrations, thereby preventing the resonance phenomenon. Soft tissue vibration amplitudes have been reported to be less than 5 per cent of the initial amplitude after two oscillations (Nigg & Wakeling, 2001). Vibration frequencies close to the resonance of lower limb

muscles have demonstrated increased muscular activity with corresponding damping of the vibration (Wakeling, Nigg & Rozitis, 2002).

Vibration damping has also been demonstrated during human walking (Wakeling, Liphardt & Nigg, 2003). At heel-strike an impact shock is transmitted to the foot and travels up through the lower limb as vibration (Wakeling et al., 2003). Evidence for muscle tuning was presented by increased gastrocnemius and biceps femoris activity along with corresponding muscle damping (Wakeling et al., 2003).

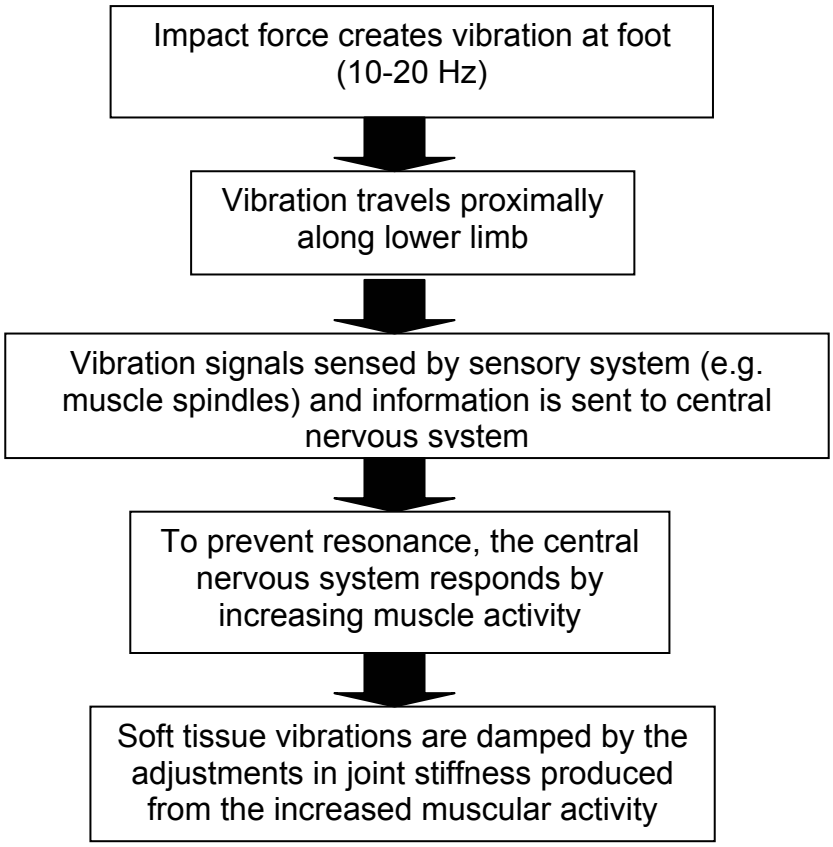


Figure 3.9: Flow chart outlining the muscle tuning hypothesis

3.6 Acute and Chronic Effects of Whole Body Vibration

3.6.1 Acute Effects

Superimposing WBV on dynamic exercise increases the rating of perceived exertion (Rittweger, Schiessl & Felsenberg, 2001) and time to exhaustion (Rittweger, Mutschelknauss & Felsenberg, 2003) compared to dynamic exercise alone. EMG analysis has demonstrated that WBV increases activity of the leg musculature (Cardinale & Lim, 2003; Delecluse et al., 2003) and this response appears to be muscle specific (Delecluse et al., 2003). The specificity of muscular response was shown by participants standing in a static squat position. In this position, WBV only mildly increased rectus femoris activity compared to that of the triceps surae, which presented with activity five times baseline static squat values (Delecluse et al., 2003). The smaller response by the thigh musculature is probably due to damping (Rubin et al., 2003). Muscle resonance also appears to have an impact upon muscle activity. The vastus lateralis, for example, has been shown to have its highest activity at 30 Hz, decreasing at higher and lower frequencies (Cardinale & Lim, 2003).

Evidence for muscular activity is further supported by increased metabolic activity, which has been shown to be similar to that observed during walking (Rittweger et al., 2000; Rittweger, Ehrig, Just, Mutschelknauss, Kirsch & Felsenberg, 2002). An early study combined WBV and squatting exercise with an additional load of 35-40% of bodyweight and reported oxygen consumption to increase by 50% of maximal oxygen uptake (Rittweger et al., 2000). This study

was poorly controlled making generalisations to the effect of WBV impossible. A group performing squat exercise only was not used making it unclear whether the elevated metabolism was due to the combination of WBV and squatting exercise or due to the squat exercise alone. A more recent paper by the same research group confirmed that WBV exercise does elevate the metabolic rate as measured by oxygen uptake (Rittweger, Ehrig et al., 2002). Manipulation of frequency and/or amplitude increased oxygen consumption and therefore energy expenditure (Rittweger, Ehrig et al., 2002). Applying an additional load of 40% of lean body mass also increased oxygen consumption (Rittweger, Ehrig et al., 2002). Even if metabolic activity is increased it appears to have little effect on fat mass (Roelants et al., 2004).

Although several studies have examined the acute effects of WBV on neuromuscular performance, generalisations are again difficult to make due to a lack of agreement in findings, most probably due to inconsistent methodologies. Despite EMG activity indicating neuromuscular fatigue (Torvinen et al., 2002b), non-exhaustive WBV has been demonstrated to acutely facilitate the strength and power abilities of active individuals and elite athletes (Bosco, Colli et al., 1999; Bosco et al., 2000; Cochrane & Stannard, 2005; Torvinen et al., 2002b). After a single WBV session in the static squat position, participants have improved countermovement jump height by 4% (Bosco et al., 2000). Recently, elite female field hockey players improved vertical jump height by 8% following 5 minutes of simultaneous WBV and dynamic exercise at 26Hz and an amplitude of 3 mm (maximum acceleration = 80.1 $m.s^{-2}$) (Cochrane & Stannard, 2005). After ten times one minute exposures performed in a static squat position, with one minute recovery periods, national level female volleyballers have displayed right

shifts of the velocity-force and power-force curves (Bosco, Colli et al., 1999). These findings, however, have been contradicted by other work that exposed untrained individuals to five times one minute WBV exposures (30 Hz; 8mm; 284.3 $m.s^{-2}$) and two minute rest periods (de Ruiter, van der Linden, van der Zijden, Hollander & de Haan, 2003). Additionally, this study showed the knee extensors to have a reduced ability to perform maximal voluntary contractions up to 60 minutes after static squat WBV (de Ruiter, van der Linden et al., 2003). Similarly, participants described as non-elite, failed to show acute improvements in vertical jump height after experiencing five times two minute exposures, separated by 40 second rest periods, at a frequency of 26 Hz and amplitude of 5.5 mm (maximum acceleration = 146.8 $m.s^{-2}$) (Cochrane, Legg & Hooker, 2004). Furthermore, they failed to demonstrate improvements in sprint and agility performance. It is unclear why these differences occurred. The failure to demonstrate improvement may have been due to large vibration intensity the untrained individuals were exposed to. It is possible that this stimulus was excessively high for untrained individuals.

Two factors that impact upon the outcome measures are the level of induced fatigue produced during WBV exposure and the WBV intensity. Compared to non-exhaustive WBV exercise, WBV combined with simultaneous squat exercise to exhaustion reduced maximal jump height, serial jump height, isometric knee extensor torque and increased EMG knee extensor activity (Rittweger et al., 2000; Rittweger et al., 2003). WBV amplitude also impacts upon the acute response. Despite a trend of improvement, healthy young adults exposed to four minutes of WBV with an amplitude of 1 mm and frequency range of 25 to 40 Hz (maximum acceleration range = 24.7 $m.s^{-2}$ to 63.2 $m.s^{-2}$), failed to demonstrate a

significant increase in isometric lower limb strength (1.1% increase compared to control) and vertical jump height (1.6% increase compared to control) (Torvinen, Sievanen, Jarvinen, Pasanen, Kontulainen & Kannus, 2002). In comparison, another group of healthy young adults who were exposed to the same frequencies and time frame, but a larger amplitude of 5 mm, showed respective improvements of 3.2% and 2.5% in isometric lower limb strength and vertical jump height compared to the control (Torvinen et al., 2002b). It should be noted however, that these acute improvements, although statistically significant are arguably not meaningful from a practical point of view. As an example, the improvement of 2.5% in vertical jump amounted to a 0.7 cm improvement (Torvinen et al., 2002b).

3.6.2 Chronic Effects

Longer term studies have ranged from 10 days to six months. As for the acute WBV research, short term WBV research lasting up to 10 days has produced inconsistent results. Physically active individuals who were exposed to 10 periods of WBV over 10 days have demonstrated 6% and 12% increases in power output and vertical jump height respectively (Bosco et al., 1998). Another group, however, found no improvement in knee extensor strength in physically active individuals that performed six WBV sessions over two weeks (de Ruiter, van der Linden et al., 2003).

In longer term studies, WBV evidence is also contradictory. In untrained individuals, two months of static and light dynamic exercises while exposed to WBV (dynamic squatting = 0-10 seconds; erect stance 10-20 seconds; static

squat = 20-30 seconds; light jumping = 30-40 seconds; alternating weight from one leg to the other 40-50 seconds; standing on the heels = 50-60 seconds) improved knee extensor strength and countermovement jump height by 7.8% and 2.5% respectively, compared to a control. A further two months of WBV, however, failed to find a difference between the WBV and control group for either performance measure (Torvinen et al., 2002a). Over an eight-month period using the same exercise protocol, untrained individuals improved vertical jump height by 7.8% more than a control group, but failed to show any change in knee extensor strength (Torvinen et al., 2003). A comparison of unloaded WBV with dynamic exercise was compared to traditional resistance training alone in untrained individuals after a 12-week period (Delecluse et al., 2003). Similar improvements in knee extensor strength occurred for both groups but countermovement jump improved in the WBV group only (Delecluse et al., 2003).

In physically active individuals, static squats and unloaded dynamic exercises on the WBV platform appears to be insufficient to improve neuromuscular performance. WBV exercise in the static squat position over 11-weeks has demonstrated no change in knee extensor strength and countermovement jump height (de Ruiter, van Raak, Schilperoort, Hollander & de Haan, 2003). Similarly, sprint trained athletes performing unloaded static and dynamic loaded exercises over a five week period demonstrated no change in sprint running velocity, knee extensor strength or knee extensor velocity (Delecluse, Roelants, Diels, Koninckx & Verschueren, 2005). Trained individuals appear to require additional loading combined with dynamic WBV exercise to improve dynamic performance. The effect of performing squats with or without simultaneous WBV on the one-repetition maximum squat (1-RM; the maximum weight that can be lifted for one

repetition) and the countermovement jump were compared over a five week period (Rønnestad, 2004). Both groups performed Smith machine squats at 6-10 RM during the intervention period and both produced improvements in their 1 RM squat. The improvements were largest for the WBV group but the difference was reported as non-significant between the two groups (p=0.046), at an alpha level of 0.01. Despite a trend for improvement in countermovement jump for the WBV group, and no change in the resistance training group, the difference was again reported as non-significant (p=0.088). The small sample sizes (n=7) in this study suggest the need for larger samples to confirm this finding.

Summary of Whole Body Vibration and Healthy Young Adults

Several formats of WBV exercise have been used by researchers. They are: (1) Standing in a static squat position; (2) Performing non-exhaustive dynamic leg exercises while standing on a WBV device; (3) Performing exhaustive dynamic leg exercises while standing on a WBV device; and (4) Performing weighted dynamic leg exercises while standing on a WBV device. Direct comparisons of these methods have not been methodically made, making generalisations somewhat difficult. Based on the current evidence, for long-term training programs it may be suggested that trained individuals require additional loading above WBV exercise alone to sufficiently improve muscular performance. Static and unloaded WBV exercise, with amplitudes as low as 1 mm, may be sufficient for individuals who are untrained.

3.7 Whole Body Vibration and Older Adults

Considering the benefits that have thus far been demonstrated for athletes and young adults, the question arises – will older adults experience similar benefits? A paucity of literature currently exists pertaining to the effect of WBV on older adults. The majority of the literature has focused upon elderly women. Although not highlighted by previous researchers, this is most likely due to at least one of two possible explanations. Firstly, as cited earlier in this literature review, the functional performance of women is more sensitive to strength and power loss (Bassey et al., 1992; Kwon et al., 2001; Lamoureux et al., 2002). A second explanation is that the research has simultaneously examined the neuromuscular and osteogenic effects of WBV (Russo et al., 2003; Verschueren et al., 2004). Post-menopausal women are more susceptible to osteoporotic fractures, therefore making this group an obvious candidate for WBV research.

The evidence for neuromuscular function improvements is promising. As discussed in earlier chapters, knee extensor strength and power are predictors of functional mobility of older adults. Six months of WBV, performed twice a week at frequencies ranging between 12 to 28 Hz and accelerations between 0.1 to 10 g (the amplitudes were not cited), increased the leg muscle power of post-menopausal women by 5% (Russo et al., 2003). Frequencies were progressively increased over the intervention period to “allow for gentle adaptation” (Russo et al., 2003). Participants stood in a static squat position for the entirety of the intervention. Initially, participants were exposed to three times one minute periods of WBV, separated by one minute of recovery. The exposure time was

progressively increased to three times two minute periods of WBV by the end of end of the six months.

The effects of 24-weeks of WBV and resistance training were compared in post-menopausal women and demonstrated the two forms of strength training to have similar increases in isometric and dynamic strength (Verschueren et al., 2004). Participants performed various squat and lunge exercises on the WBV plate. The WBV frequency ranged between 35 to 40 Hz and amplitude between 1.7 to 2.5 mm. The intensity was also increased by changing the execution of the exercises from two legs to one leg. Lean muscle mass did not increase in the WBV group indicating that the strength increases were neurogenic. It is generally understood that the early gains in strength after resistance training are largely the result of neurogenic adaptations (Kraemer et al., 2002; Sale, 1988; Staron et al., 1994) and this appears to be similar for WBV training as well.

Another study also compared the effect of WBV and resistance training on knee extensor strength and speed, as well as countermovement jump performance in post-menopausal women aged approximately 65 years of age (Roelants et al., 2004). Again participants performed various forms of squat and lunge exercises. The WBV frequency and amplitude ranged between 35-40Hz and 2.5 to 5 mm respectively. After 24-wks of training the WBV and resistance training groups displayed comparable changes, increasing in isometric strength by 15% and 18.4% respectively, and dynamic knee extensor strength by 16.1% and 13.9% respectively. The majority of improvements occurred during the first 12-weeks; from weeks 13 to 24 both forms of training failed to show any significant changes in either type of strength. Both groups displayed decreases in speed of

movement and improved countermovement jump height. Similar to knee-extensor strength, the majority of countermovement jump improvement occurred during the first 12-weeks for WBV (16%) and resistance training (12.1%).

Balance is an important component for every day activities such as standing and walking and relies upon strength and proprioceptive function. An eight-month whole body vibration program performed 3-5 times per week had no effect on balance in healthy young adults (Torvinen et al., 2003). This is probably due to the young age of these participants who in this case may be presenting with a "ceiling effect". In post-menopausal community dwelling women, however, balance has been demonstrated to improve after a 24-week WBV training program (Verschueren et al., 2004). Although, no change in static balance was shown (Verschueren et al., 2004), WBV appeared to have a positive effect on dynamic balance after perturbation (Verschueren et al., 2004).

In nursing home residents, WBV has been shown to improve clinical measures of balance and gait (Bautmans et al., 2005; Bruyere et al., 2005). The Tinetti test is a subjective test performed by clinicians to assess gait and balance abnormalities. The timed-up-and-go is also used to assess functional performance. For this test, participants are asked to rise from a standard sized chair, walk to a marker three metres away, turn and return to the chair. Performance is determined by the time measured to complete the test. In one study participants were exposed to four times one minute periods of WBV over six weeks (Bruyere et al., 2005). During the first and third minutes, the frequency was at 10 Hz and amplitude at 1.5 mm. The second and fourth minutes consisted of a frequency of 26 Hz and amplitude of 3.5 mm. At baseline,

participants had a Tinetti Global score of 14.9/28. By the end of the 6-week period participants had a mean score of 20.5/28 and decreased the time to complete the timed-up-and-go by 13.8 seconds. These improvements were not consistent with another study that examined nursing home residents and WBV (Bautmans et al., 2005). In this study, the timed-up-and-go improved by three seconds only and the Tinetti score failed to show any improvements (Bautmans et al., 2005). It is unclear why these differences were reported. It may be due to the different forms of WBV used where one study used tilting oscillations (Bruyere et al., 2005) and the other vertical (Bautmans et al., 2005). Tilting oscillations may have a larger impact upon balance mechanisms in the elderly. A unique component of this study was measuring blood pressure and heart rate during WBV. Changes in both of these parameters were shown to be clinically insignificant (Bruyere et al., 2005).

Summary of Whole Body Vibration and Older Adults

WBV has been shown to improve the strength and power of elderly nursing home residents. These improvements are believed to transfer across to the functional task of walking. To date, only clinical measures such as the Tinetti test and the timed-up-and-go have been used to identify the changes in functional status following WBV.

4.0 Overview

A purpose built whole body vibration platform (ACUWBV) was developed for the intervention studies forming the pilot and major research. The ACUWBV produced vibration via tilting oscillations and allowed the researcher to manipulate vibration amplitude without the participant changing stance width, and therefore stance posture. The development of the ACUWBV consisted of three main phases: (1) Identifying the platforms specifications; (2) Measuring the reliability of setting the amplitude of the ACUWBV; and (3) Frequency calibration. Statistical analysis demonstrated that the 'typical' stance width of 26 individuals was 31.8 ± 5.4 cm between the third digits of each foot. The platform was therefore designed to allow participants to place the third digit, of each foot, 16 cm from the platforms tilting axis. The design of the variable eccentric cam used for this research allowed for the adjustment of vibration amplitude without the participants changing their stance width. The reliability of adjusting the amplitude with this device was determined to be acceptable with typical errors of ±0.02 mm for an amplitude setting of 0.5, and ± 0.01mm for an amplitude setting of 1.0 mm. The final stage of the ACUWBV consisted of the calibration of the platforms vibration frequency with a strobe light.

4.1 Introduction

An ACU designed and purpose built whole body vibration plate was used for this research (Figure 4.1). The following report outlines the development of the custom-made WBV plate (ACUWBV), its specifications and measures of reliability.

Figure 4.1: The whole body vibration plate (ACUWBV) that was custom-made for this research.

The ACUWBV produces sinusoidal vibrating perturbations via tilting oscillation, similar to the Galileo 2000 (NovotTec; Pforzheim, Germany), a commercial WBV plate used by several WBV researchers (e.g. Cochrane et al., 2005; de Ruiter et al., 2003; Rittweger et al., 2003). To increase WBV amplitude and therefore intensity on the Galileo 2000, it is necessary to widen the foot placement on the plate. Figure 4.2 demonstrates how the amplitude of the tilting board increases as the distance from the axis increases. Previous research (Harazin & Grzesik, 1998; Matsumoto & Griffin. 1998) has established that different standing postures influence the transmission of WBV from the vibrating platform to the various structures of the human body (Figure 4.3). The ACUWBV was therefore developed to allow for a consistent stance width by participants throughout this research, that is, participants would not have to change their stance width to increase the vibration amplitude.

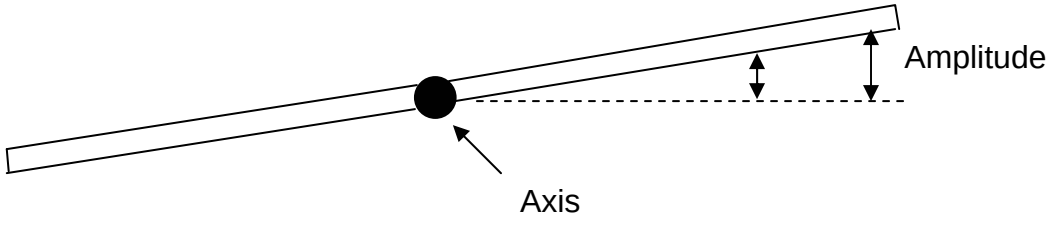


Figure 4.2: Any fixed point on the plate undergoes the same rotation, however each point undergoes a different linear displacement based upon the distance from the axis.

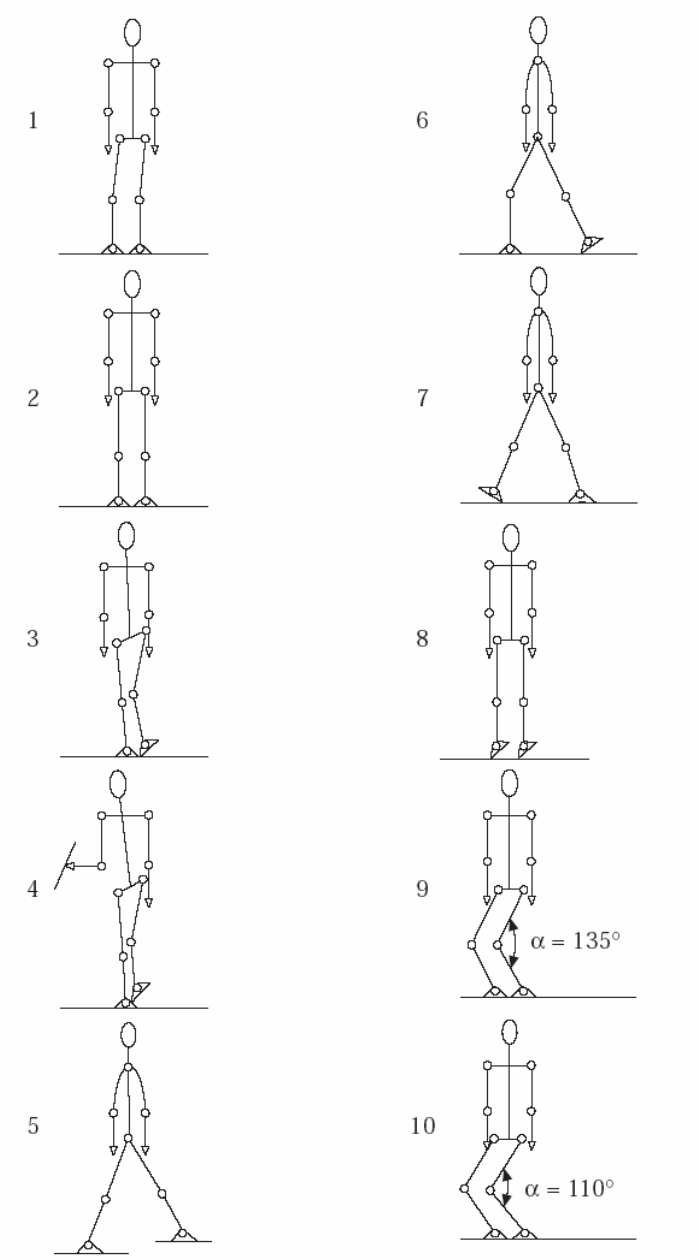


Figure 4.3: Different standing postures examined by Harazin & Grzesik (1998) during WBV exposure. 1. Relaxed standing, hands at sides; 2. Standing with legs stiffened in knee-joints; 3. Standing on the right leg with the support of the left foot toes; 4. Standing on the right leg, while the heel of the left foot was raised up to the level of right medial ankle; 5. Standing in step; 6. Standing in step, the left foot behind, supported on the toes; 7. Standing in step, the left foot ahead, supported on the heel; 8. Standing on the toes with the heels raised about 2-4 cm; 9. Standing with feet astride, knees bent at the angle of 135 degrees; 10. Standing with feet apart, knees bent at 110 degrees.

4.2 Platform Size

For the pilot and major studies, participants were requested to stand at a relatively constant stance width to reduce the influence of posture on vibration transmission (Harazin & Grzesik, 1998; Matsumoto & Griffin. 1998). To identify a 'typical' stance width, twenty-six female participants with an age range of 30-65 years were recruited via an e-mail sent out to the ACU National (Melbourne Campus) staff. Participants took off their shoes and marched on the spot for 50 steps. The distance between the third digits (middle toe) of the feet was then measured. Table 4.1 lists the descriptive statistics for the stance width data.

Table 4.1: Descriptive statistics of the participants (n=26) stance width (cm).

Mean	31.8
Standard Deviation	5.4
Standard Error	1.1
Median	30.0
Mode	30.0
Range	21.0
Minimum	22.0
Maximum	43.0

The mean stance width was 31.8 ± 5.4 cm, with the smallest stance being 22.0 cm and the largest 43.0 cm. The most common stance width was 30.0 cm (n=5). Based upon these data, it was decided that the mean stance width of this group would be used as the constant stance width for the research to standardise the vibration transmission (Harazin & Grzesik, 1998; Matsumoto

& Griffin. 1998). Therefore, during the WBV intervention research, participants stood with their third digit 16 cm from the axis of rotation on the tilting board.

4.3 Specifications of WBV device

A photo of the mechanism that produced the oscillations for the ACUWBV is presented in figure 4.4 and the major components are illustrated in figure 4.5. Sinusoidal platform oscillations were produced using a tilting board that rotated about a central axis of rotation. To generate movement of the tilting board, a lever arm was attached to the tilting board and an eccentric cam. A motor, controlled by a Penta-Drive™ DC Motor Speed controller (Nema-4x/IP-65), drove the eccentric cam to turn at the desired frequency.

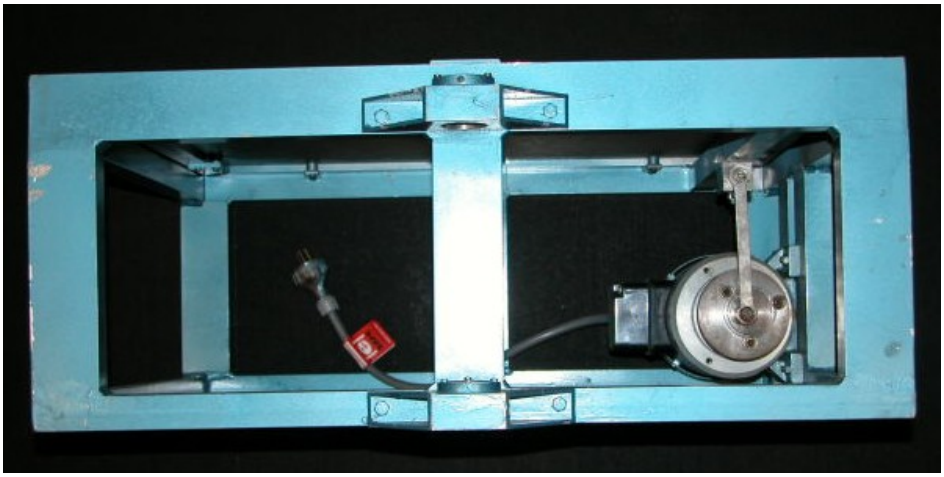


Figure 4.4: Photo of the mechanism that produced the oscillations for the ACUWBV

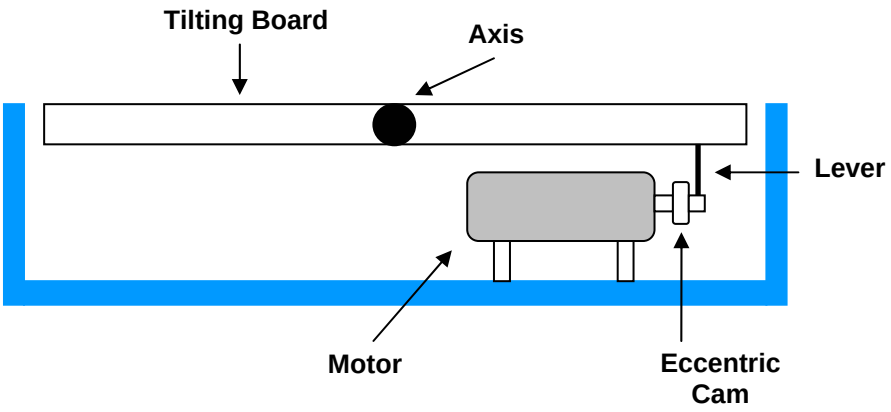


Figure 4.5: The major components of the WBV platform.

As explained above, the sinusoidal platform oscillations were produced using an eccentric cam. An eccentric cam is a disc that has its axis of rotation off-centred. Therefore, when the axis of rotation was at the high or low point, the platform was respectively at its high or low point (Figure 4.6).

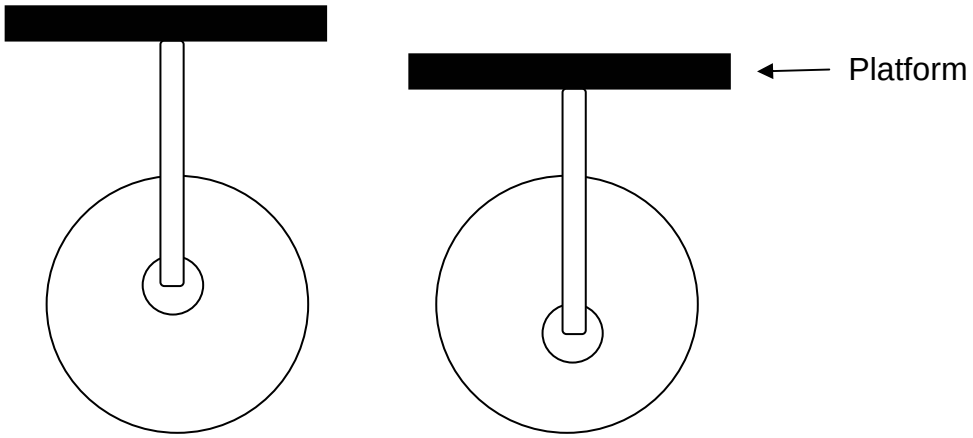


Figure 4.6: A diagram of the principle of the eccentric cam. When the axis of rotation is at its high or low points the platform was respectively at its high or low point.

Commercial WBV plates currently on the market, such as the Galileo 2000, require the individual to vary their stance width to alter the vibration amplitude. The unique aspect of the ACUWBV is the variable eccentric cam (Figure 4.7)

that does not require changes in stance width to alter the amplitude. This device has a central eccentric cam placed within a housing disc. Rotation of the central eccentric cam within the housing disc varies the position of the drive shaft and lever arm, therefore altering the amplitude of the platform as it oscillates. A line on the housing disc is matched to the eccentric cam to determine the amplitude magnitude.

(a)

(b)

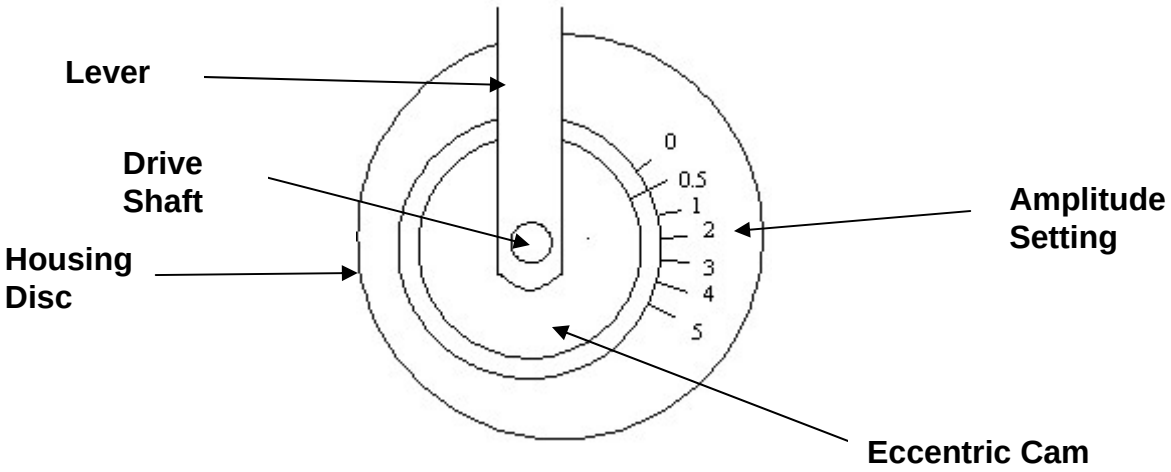


Figure 4.7: (a) Photo of variable amplitude eccentric cam; (b) Diagram of variable amplitude eccentric cam.

By placing the amplitude of the eccentric cam on 0 mm, the eccentric cam developed for this research also allows the use of a placebo. For this setting, participants can hear the motor running and can also feel a mild sensation of vibration created by the motor. Few WBV studies have used a placebo as a control effect. A previous study that used a placebo utilised a tape recorded sound of the WBV plate motor running while participants stood on the device (Bautmans et al., 2005). Another study reported that participants "could hear the motor and experienced tingles on their foot soles" and that the "acceleration of the platform was only 0.4 g" but they did not explain how they created the placebo effect (Delecluse et al., 2003).

4.4 Reliability of Setting the ACUWBV Eccentric Cam

The ACUWBV eccentric cam was adjusted within a housing disc each time a different amplitude was utilised. Adjusting the eccentric cam meant that it was susceptible to some level of error. It was therefore necessary to identify the reliability of setting this device. The following section reviews statistical procedures commonly cited in reliability research and then presents the study that was performed to assess the reliability of setting the ACUWBV eccentric cam.

4.4.1 Review of Statistical Procedures for Reliability Studies

Reliability has been defined as the ability to provide a measurement that is consistent and free from error (Portney & Watkins, 2000), or as the acceptable amount of measurement error for the effective practical use of a measurement

tool (Atkinson & Nevill, 1998). Without measures of reliability, generalisations about data cannot be confidently made (Portney & Watkins, 2000). Measurement errors can be attributed to three factors: (1) the person recording the measurements (the rater); (2) the measurement instrument (the dial indicator used to measure the accuracy of setting the amplitude); and (3) the variability of the attribute being measured (in this case the amplitude setting) (Portney & Watkins, 2000).

Systematic bias and random error are associated with measurement error (Atkinson & Nevill, 1998). The combination of these two components is called the total error (Atkinson & Nevill, 1998). Systematic errors are predictable errors of measurement, where estimation is consistently above or below the correct value (Portney & Watkins, 2000). Systematic error with the ACUWBV may be made each time the lines on the housing disc and eccentric cam are matched. In this case, the rater consistently under or overestimates the true amplitude measure. Random errors occur due to chance and can be the result of factors such as fatigue, poor attention or mechanical error (Portney & Watkins, 2000). In the case of the present study poor attention to aligning the lines may lead to random error.

4.4.1.1 Reliability Tests

The paired t-test may be used to compare test and retest means. This form of analysis should be interpreted cautiously as random variability can disguise the systematic variation between tests (Atkinson & Nevill, 1998). That is, no significant difference may be found between test-retest conditions due to large

amounts of random error between tests (Atkinson & Nevill, 1998). When more than one retest is to be analysed, a repeated measures ANOVA combined with an appropriate post hoc test may be used to identify systematic bias during test-retest observations (Atkinson & Nevill, 1998). However, this test needs to be treated with the same caution as the paired t-test since systematic bias can again be hidden by random variation (Atkinson & Nevill, 1998).

Although reliability studies have reported Pearson's correlation coefficient to measure reliability, it is considered a limited method of reliability analysis (Portney & Watkins, 2000; Vincent, 1995). Firstly, instead of providing a measure of agreement, it provides the extent to which two variables move together, or their covariance (Vincent, 1995). Pearson's correlation coefficient is therefore unable to identify systematic error in measurement (Vincent, 1995). Secondly, the bivariate nature of correlations allow analysis of only two sets of data at the one time, even though several variables may be of interest (Vincent, 1995).

Some researchers use more than one reliability index within a single study to improve reliability analysis and interpretation (Portney & Watkins, 2000). A correlation and a t-test, for example, may be used to assess stability and agreement between two sets of data (Portney & Watkins, 2000). However, the combination of the correlation and t-test may be ambiguous and still make interpretation difficult. For example, there is no clear method of interpreting

the data if the data are correlated but significantly different or if they are poorly correlated but are not significantly different (Portney & Watkins, 2000).

The intraclass correlation coefficient (ICC) is considered to be a more desirable index for reliability analysis (Atkinson & Nevill, 1998; Hopkins, 2000a; Portney & Watkins, 2000; Vincent, 1995), partly because it is more sensitive to systematic error than the Pearson's correlation (Atkinson & Neville, 1998). The ICC is considered to be more advantageous because it is univariate (whereas Pearson's correlation is bivariate) and can be used when more than one retest has been performed (Atkinson & Neville, 1998). As with other reliability coefficients it ranges between 0.00 and 1.00, a score of 1.0 indicating excellent reliability.

4.4.1.2 Confidence Limits for Reliability Studies

Although the ICC is supported as a valuable tool in reliability research (Atkinson & Nevill, 1998; Hopkins, 2000a), it has been suggested that this measure should not be used alone to address reliability (Hopkins, 2000a). Reliability tools such as the ICC provide the researcher with a gauge of reliability but give no indication of confidence limits for the true value, i.e. the value that would be recorded if measurement error did not exist. It has therefore been recommended that confidence limits are included with all reliability estimates (Hopkins, 2000a). The standard error of measurement (SEM) and 95% limits of agreement (LOA) are two statistical methods that quantify measurement error.

The SEM is also known as the within-subject standard deviation (Atkinson & Nevill, 2000), and has more recently been popularised within exercise science statistical literature as the 'typical error' (Hopkins, 2000a). This statistical tool will be termed as typical error hereon in. The typical error (TE) is the standard deviation from the mean that is calculated from a set of observed values. Theoretically, there is a 68% or 95% probability that the true value will fall within one or two standard deviations of the mean value of the observed value (Portney & Watkins, 2000). Where the ICC is used, the typical error is calculated from the following formula:

$$\text{TE} = SD\sqrt{1 - ICC}$$

where SD is the standard deviation, and ICC is the intraclass coefficient derived from the ICC(3,1) model (Hopkins, 2000a). The ICC(3,1) model is calculated using the formula below:

$$ICC(3,1) = \frac{BMS - EMS}{BMS + (k-1)EMS}$$

where the first number (3) designates the model and the second number (1) denotes the number of raters, BMS is the between-tests mean squared, EMS is error mean squared and k is the number of tests (Portney & Watkins, 2000; Shrout & Fleiss, 1979).

Application of the TE is demonstrated in the following example. With reference to the reliability of setting the amplitude of the eccentric cam used in this research, assume the TE is 0.1 mm after being derived from predetermined measures of the standard deviation and ICC. If the eccentric cam was set at 1 mm, the researcher can be 68% confident that the amplitude

will be set between 0.9 to 1.1 mm (Mean ± 1TE). Furthermore, confidence can be held at 95% that the amplitude will be set at an amplitude between 0.8 to 1.2 mm (Mean ± 2 TE).

According to Hopkins (2000b, p. 375), "LOA represent a reference interval for 95% of an individual's test-retest differences". Conceptually, this has been presented as:

Observed value - LOA < true value < observed value + LOA,

where the true value falls somewhere between the observed value ± 95% LOA. Using the ANOVA approach, where multiple retests are performed, the 95% LOA can be calculated from:

$$\text{LOA (95\%)} = 1.96\sqrt{2MSE}$$

where MSE is the mean squared error calculated from repeated measures ANOVA.

Again, using the eccentric cam used in this research as the example, the LOA provides the expected error range when setting the amplitude over multiple occasions, that is, 95% of test-retest scores will fall between a statistically determined reference range. Assume the amplitude of the eccentric cam is set by the researcher to 1 mm by aligning the marked lines. The accuracy of this setting is then measured with a dial indicator and recorded to be 1.07 mm. If the researcher was to later reset the amplitude to 1 mm by again aligning

the lines on the eccentric cam, and on this later occasion the dial indicator indicates the actual amplitude to be 1.10 mm, an increase of 0.03 mm has occurred. Assume the LOA for the amplitude setting are 0.10 mm. The 95% confidence limit would be 0.03 ± 0.10 mm, or more specifically the researcher would expect that 95% of test-retest measures for the eccentric cam would lie between 0.07 and 1.13 mm.

The validity of each of these statistical methods for describing measurement error has been strongly debated within the exercise science literature (Atkinson & Nevill, 1998; Atkinson & Nevill, 2000; Hopkins, 2000a, Hopkins, 2000b). A major limitation of the LOA that has been recognised by both groups is that its accuracy is based upon sample size, the smaller the sample size the greater the risk of bias (Atkinson & Nevill, 1998; Hopkins, 2000a). A sample size of at least 40 individuals should be tested if LOA is to be used (Atkinson & Nevill, 1998). The typical error produces an expected value that is not reliant upon sample size (Hopkins, 2000a). Therefore, in studies where sample size is small, typical error is a more robust indicator of measurement error.

4.4.2 Experiment – Test-Retest Reliability for Setting the Amplitude of ACUWBV

The amplitudes for the purpose built ACUWBV plate was positioned by manually aligning two lines on an eccentric cam system (Figure 4.7b). The configuration of these two lines is susceptible to systematic and random error.

The reliability and typical error of these manual settings were therefore measured.

4.5 Method

Since only one rater would be responsible for setting the amplitude of the ACUWBV, intra-rater reliability was measured only. The rater randomly performed three tests for each amplitude setting. There were no individuals standing on the platform during this procedure. The rater was required to align the lines for the corresponding amplitudes that were randomly selected. Upon alignment, the rater measured the peak-to-peak displacement of the WBV plate using a dial indicator which measured to the nearest 0.01 mm (Figure 4.8). This value was then halved to calculate amplitude. These measurements were recorded on the same day over a period of approximately two hours.

Figure 4.8: The dial indicator used to measure the accuracy of amplitude settings on the ACUWBV.

An intraclass correlation coefficient (ICC) incorporating a repeated measures ANOVA design was used to test the reliability of setting the amplitude. The following formula was used for to determine the ICC (Hopkins, 2000a; Portney & Watkins, 2000, p565):

$$ICC(3,1) = \frac{BMS - EMS}{BMS + (k-1)EMS}$$

where the first number designates the model and the second number denotes the number of raters, BMS is the between-tests mean square, EMS is error mean square and k is the number of tests.

The typical error was calculated using the following formula:

$$\text{TE} = SD\sqrt{1 - ICC}$$

where SD is the standard deviation, and ICC is the intraclass coefficient derived from the ICC(3,1) model.

4.6 Results & Discussion

Descriptive statistics for the ACUWBV is presented in table 4.2. The SD was relatively similar for each of the amplitudes. The smallest range was 0.01 mm for the 1 mm amplitude setting, and the largest was 0.12 mm for the 2mm amplitude setting.

Table 4.2: Descriptive statistics for reliability data of setting the ACUWBV amplitude.

True Amplitude (mm)	Observed Mean Amplitude (mm)	SD	Range (mm)	Typical Error (mm)	Expected Amplitude Error Range (mm)[a]	True WBV Acceleration ($m.s^{-2}$)[b]	Expected WBV Acceleration Error Range ($m.s^{-2}$)[c]
0.5	0.52	0.02	0.03	0.02	0.48-0.56	7.90	7.58-8.85
1	1.09	0.01	0.01	0.01	1.07-1.11	15.80	16.90-17.53
2	2.11	0.06	0.12	0.05	2.01-2.21	31.60	31.75-34.91
3	3.01	0.04	0.07	0.03	2.95-3.07	47.40	46.60-48.49
4	4.01	0.05	0.09	0.04	3.93-4.09	63.20	62.08-64.60
5	5.05	0.03	0.05	0.02	5.01-5.09	79.00	79.14-80.40

[a] Expected amplitude error range was calculated from the typical error (2TE).
[b] True WBV acceleration was calculated from the true amplitude and a constant frequency of 20 Hz.
[c] Expected WBV acceleration error range was calculated from the expected amplitude error range and a constant frequency of 20Hz.

The ICC value was 0.4 for the ACUWBV which is considered to represent low reliability (Portney & Watkins, 2000). Visual inspection of the raw data demonstrated a tendency for over estimation of amplitude. This was further supported by the mean amplitude data (Table 4.2) where all recorded amplitudes were above the desired value. Although the calculated ICC is considered to indicate poor reliability, the typical errors for each amplitude demonstrate the error range to be quite small and in final outcome terms (i.e. plate acceleration) the impact is quite small (Table 4.2). It was therefore concluded that setting the amplitude of the ACUWBV eccentric cam could be reliably performed.

4.7 Frequency Calibration

Once amplitude reliability was determined, a strobe light (Figure 4.9) was used to calibrate the frequency of the spinning cam, and therefore the frequency of the vibrating platform.

Figure 4.9: The strobe light used for calibration of the frequency of the ACUWBV plate.

Strobe lights produce flashing light. The strobe light was switched on and manually adjusted to flash at the desired frequency of the spinning cam. The speed controller was switched on and was manually adjusted until a marker that had previously been placed on the spinning cam appeared to be stationary (Figure 4.10 (a)). This stationary illusion was produced by the simultaneous frequencies of the spinning cam and the flashing strobe light. Figure 4.10 (b) demonstrates that when the strobe light and spinning cam are out of synchronisation there appears to be multiple markers, despite only one marker actually being on the cam. The frequencies to be set were 10, 20, 25 and 30 Hz. It was accepted that a certain level of error would be produced

using the strobe light as visual inspection was made of the moving marker placed on the spinning cam.

(a)  (b) 

Figure 4.10: Frequency calibration of ACUWBV. (a) Marker appears stationary when the cam is spinning at 20 Hz and strobe light is flashing at 20 Hz; (b) Multiple markers appear when the cam is spinning at a different frequency (20 Hz) to the flashing strobe light (30 Hz).

Summary

This research outlined the development of the ACUWBV. In summary: (1) The setting of amplitude with the ACUWBV eccentric was established to be reliable; (2) The frequency of the spinning cam was calibrated using a strobe light; and (3) A stance width of 16 cm each side of the tilting axis of the ACUWBV was determined for the pilot and major research.

5.0 Overview

This pilot study aimed to identify the effect of low intensity whole body vibration (WBV) on the spatio-temporal parameters of elderly female fast gait. Seven female participants aged 77.43±2.07 years volunteered for three WBV sessions per week for three weeks. Throughout the intervention period, participants were exposed to progressively increasing WBV accelerations. The amplitude was set at 0.5 mm for all sessions and the frequency increased over the three weeks from 10 Hz to 25 Hz. The participants stood on a purpose built WBV platform, with knees bent, for five one minute efforts. The WBV exercise bouts were interspersed with one minute rest periods. To determine the effect of WBV on the spatio-temporal variables of gait, participants performed fast walks over an electronic walkway (GAITRite) at the end of each WBV session. Time-series analysis displayed a linear increase in stride velocity over the three week intervention period. Conversely, stride time, stance time and double support time exhibited linear decreases. A repeated measures ANOVA was performed on gait data from sessions one, four and eight only. Stride time ($p<0.05$) and stance time ($p<0.05$) significantly decreased over three weeks. Stride velocity ($p=0.10$) and double support ($p=0.08$) time failed to demonstrate significant differences. A power analysis from the pilot data demonstrated that a sample size of at least eight participants should be used for future research. No adverse effects were reported during this study. As a result of this pilot study, future research should consist of a larger sample size, include a control group, and be of longer intervention duration.

5.1 Introduction

Older adults are slower walkers than healthy young adults at fast and normal walking speeds (Bohannon, 1997; Murray et al., 1969). The primary explanation for the slower walking speed is a reduced stride length (Murray et al., 1969), while increases in stride time, stance time, single support time and double support time have also been demonstrated (Murray et al., 1969).

A multifactorial relationship of physiological and neuropsychological variables exists, contributing to gait instability in older adults (Hausdorff et al., 2001). From a neuro-physiological perspective, the quality of sensory information from the visual, vestibular and proprioceptive system decline compromising the quality of afferent information provided to the older adult. Furthermore, the quality of response output declines due to decayed efferent pathways and atrophied skeletal muscle.

These changes in mobility with age place a burden on the individual and the larger community (Hall & Hendrie, 2003). For the individual, a slower gait may impact upon daily activities such as negotiating traffic intersections (Oxley, 2004). For the community, this burden is reflected by economic costs. In Australia, older adults are the fastest growing part of the population, especially in the 85 year age group (Australian Bureau of Statistics, 2003). Strategies are currently being investigated and developed to address this economic issue.

One such strategy is resistance training which aims to increase or maintain strength and power. It is reasonable to suggest that the increases in strength and

power that older adults may experience will cross-over to functional performance such as gait. To date, resistance training has shown equivocal results for increasing the gait speed of older adults whereby gait speed has either increased (Fiatarone et al., 1990; Heiwe et al., 2001; Schlicht et al., 2001; Topp et al., 1996) or shown no changes (Buchner et al., 1997; Cress et al., 1999). Where increases in gait speed have been reported following a resistance training intervention, there has been no indication of which gait parameters (e.g. stride length, stride time) were associated with the changes. Decreased stride length is considered to be the primary contributor to the reduced walking velocity with age (Lord et al., 1996; Elble et al., 1991; Hageman & Blanke, 1986; Öberg et al., 1993). It is therefore plausible that an increased stride length contributed to these increases in walking velocity.

Recently, WBV has shown promising results for increasing leg extensor strength and power of older adults (Roelants et al., 2004; Russo et al., 2003; Verschueren et al., 2004). For this intervention, participants stand on an oscillating plate that stimulates the spinal pathway via the stretch reflex. Nursing home residents have also demonstrated improvements in clinical measures of balance and gait over six week periods (Bautmans et al., 2005; Bruyere et al., 2005). Although two studies have documented improvements for the timed-up-and-go (Bautmans et al., 2005; Bruyere et al., 2005), only one group has been able to demonstrate improvements in the Tinetti test score (Bruyere et al., 2005). The timed-up-and-go examines functional mobility by measuring the time it takes to get out of a chair, walk three metres, turn around and return. The Tinetti Test qualitatively evaluates gait, assessing parameters including step length, foot floor clearance,

step symmetry and step width. It is unclear what parameters contributed to the faster timed-up-and-go times. The observed improvements, for example, may not have been due to the walking velocity component of this test but due to chair rise time. This research therefore aimed to quantify changes that may occur in gait and its associated parameters following WBV. As there had been no previous research examining the response of the spatio-temporal parameters of elderly female gait to WBV, a power analysis was also performed to identify statistical power for future research.

The intensity of WBV is measured by its frequency (the number of vibrations per second) and amplitude (the distance moved by the plate from the horizontal during each oscillation). The most appropriate frequencies and amplitudes for older adults are currently poorly understood and have often been inadequately reported. For athletes and healthy young adults, frequencies have typically ranged between 20 to 35 Hz (Bosco et al., 1998; Bosco et al., 2000; Rittweger et al., 2001). The most significant impact on the accelerative forces is the amplitude, which has varied significantly between studies ranging between 2.5 to 8 mm (Bosco et al., 2000; Cochrane et al., 2004; Cronin et al., 2004; de Ruiter, van Raak et al., 2003; Rittweger et al., 2001; Roelants et al., 2004). For older adults, frequencies have ranged between 10 to 40 Hz and amplitudes 1.5 to 5 mm (Bryuere et al., 2005; Roelants et al., 2004). Based upon the reported frequencies and amplitudes, the g-forces experienced by the participant at the point of contact of the plate can be extremely high. For example, a study that reported acute adverse effects in young adults used a frequency of 26 Hz and an amplitude of 6 mm, which calculates to an

acceleration of 16.3 g (Cronin et al., 2004). Potentially, older adults susceptible to osteoporotic fractures are at a greater risk of injury from such high forces. There is, therefore, a need to assess the impact lower intensities may have on older adults.

There is limited literature reporting the response of older adults to WBV accelerations at the lower end of the spectrum. The primary aim of this study therefore was to examine the response of older female gait to relatively lower gravitational accelerations.

Presently, all of the WBV research has performed baseline measures on targeted outcome measures, such as the Tinetti score, and re-tested these at the end of the intervention period. This method of assessment does not provide information about the changes in gait that occur over time. As previously stated, improvements in Tinetti scores and timed-up-and-go times have been identified, but it is unclear how long it takes to see changes from the WBV and whether improvements continue to increase, plateau or even decrease. This study therefore aims to perform a time-series analysis, and chart the response of gait throughout the intervention period.

Based upon the limited observations and findings presented in the WBV research, the following hypotheses were made:

(1) That the stride velocity of elderly females exposed to low intensity WBV would increase over a three week intervention period;

(2) That the increased stride velocity of elderly female fast gait would be associated with increased stride length;

(3) That the increased stride velocity of elderly female gait would be associated with decreases in stride time, stance time, single support time and double support time

5.2 Method

5.2.1 Participants

Seven participants, 75 to 81 years old, were recruited from church and Returned Soldiers League (RSL) groups for this study. Eligible participants were ambulatory, able to follow simple commands, and not suffering from unstable cardiovascular disease or other uncontrolled chronic conditions that would interfere with the safety and conduct of this study (Fiatarone et al., 1990).

5.2.2 Screening

For participant safety, screening tests were used to identify individuals at risk of falling. A two-stage screening process was performed and those deemed at risk were excluded from this study. The first stage required participants to complete a self-reported medical history and to visit their physician for a medical clearance (sections 5.2.2.1 to 5.2.2.3). The second stage evaluated physiological risk factors that have previously been identified as relatively reliable predictors of falls (sections 5.2.2.4 to 5.2.2.6).

5.2.2.1 Self-Reported Medical History

The first stage of screening required participants to complete a self-reported medical history form (Appendix E). The form identified whether the participant had ever experienced hospitalization or any fractures throughout their lifetime and whether they had experienced any falls in the past 12-months. The

participants were also asked to highlight whether they had any medical conditions, including: (1) musculo-skeletal dysfunction, (2) neuromuscular dysfunction, (3) overuse injuries, (4) vascular disorders, (5) diabetes, (6) arthritis; (7) visual impairment; (8) persistent vertigo; (9) light-headedness; and/or (10) epilepsy. Participants who had a fall within the previous 12-months were excluded from the study.

5.2.2.2 Prescribed Medication

Participants were asked to specify the medication that they were currently using. Participants who were taking medication that may affect their balance were excluded from the study. Non-prescribed medication and vitamin supplements were not documented.

5.2.2.3 Medical Clearance

Participants were required to have a medical examination to identify medical conditions that would preclude exercise testing and training. After completing the self-reported medical history, the participants were requested to show this form to their physician who was asked to write his/her recommendation that it was safe to exercise and provide any other additional comments. A letter outlining the study was also provided to the physician (Appendix D).

5.2.2.4 Vision

Visual Acuity

Individuals with poor visual acuity are more likely to walk slower and are considered to have a high risk of falling (Black & Wood, 2005). For this study, binocular visual acuity was measured with participants sitting six metres from a Snellen Chart (Verbaken, J., Australian Vision Charts, Melbourne, Australia, 1988). Testing was conducted in a room lit by artificial light above the participant and natural lighting behind the participant. The Snellen chart has several lines of letters, decreasing in size. If habitually worn for walking, participants wore corrective glasses for this test, and were requested to read aloud the letters on each line of the chart, starting from the top. The score from the lowest correct line was recorded. Participants were excluded from the study if they had a logmar score higher than 0.4 (six metres).

Contrast Sensitivity

The visual information provided by the various spatial frequencies and contrast levels of the visual environment contribute to the avoidance of trip hazards (Black & Wood, 2005). The Melbourne Edge Test (MET) is a well-accepted measure of contrast sensitivity (Lord & Dayhew, 2001) and has a good external validity as a predictor of falls (Lord & Dayhew, 2001; Lord, Clark & Webster, 1991; Lord, Ward, Williams & Anstey, 1994). Participants sat in a room illuminated by artificial light above them and natural light behind them. A MET card (Figure 5.1) was presented to the participant at the individual's normal reading distance. The MET card consists of 24 circles of 25 mm in

diameter, divided into two halves with variable orientation and reducing contrast. The angle of division of these halves varied between vertical, horizontal, left diagonal and right diagonal. The participant reported the orientation of the shading. When the participant was unable to perceive the contrast, the last correctly identified circle was recorded as the contrast sensitivity score. The maximum score that could be achieved was 24 circles. Scores lower than or equal to 16 have been identified as an indicator of poor contrast sensitivity, therefore, only the participants that recorded scores of 17 or more were included in this study.

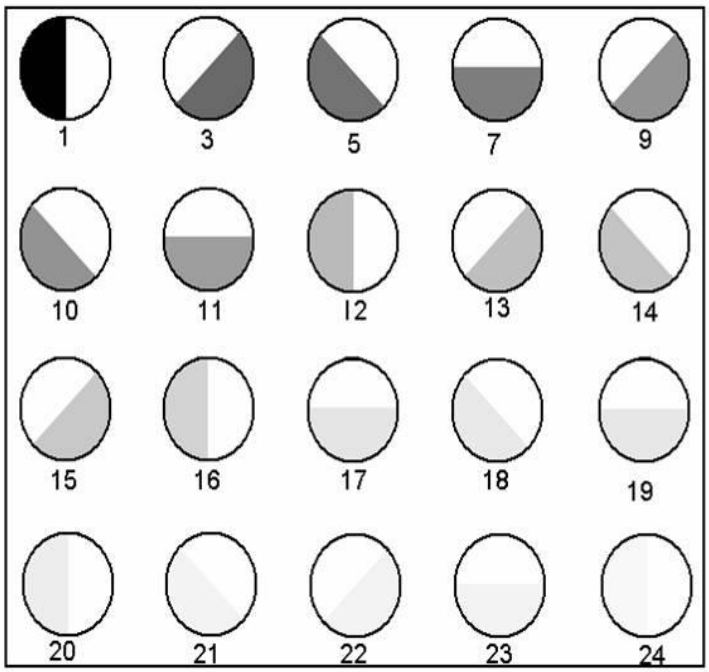


Fig 5.1: Melbourne Edge Test

5.2.2.5 Romberg Test

Various versions of the Romberg test have been developed, all to identify reduced proprioceptive control of the lower limbs. The standard Romberg test was used, where participants were requested to stand with their feet together and close their eyes for 30 seconds (Hill, 1997). Participants were excluded from the study if they lost balance and took a step, opened their eyes during the 30 second testing period, or needed manual steadying by the researcher.

5.2.2.6 Cognition

Mental status is associated with fall risk in older adults (Rait et al., 2005). The mini-mental state examination is an 11-item questionnaire that is recognized as a reliable tool for identifying cognitive impairment and fall risk (Appendix F). The maximum score for the MMSE is 30. A score below 24 is considered to be an indicator of cognitive impairment and was therefore used as an exclusion criterion for this study.

5.2.3 Gait Analysis

Participants deemed to be healthy and not at risk of falling were then invited to continue with the study. The third and final stage, before administration of the WBV intervention, was to perform baseline gait analysis on participants. The following section outlines the gait analysis instrumentation and procedures used for this study.

5.2.3.1 Gait Analysis Instrumentations

A standard GAITRite walkway (CIR Systems Inc., Philadelphia, USA) was used for this study (Figure 5.2). This system has been shown to have high concurrent validity and repeatability (Bilney, Morris & Webster, 2003; Menz et al., 2004; van Uden & Besser, 2004; Wilson, Lorenzen & Lythgo, 2002). The walkway is comprised of six sensor pads encased in a roll-up carpet that has an active area of 61 cm wide by 366 cm long and has a sampling rate of 80Hz. The sensors are organized in a (48x288) grid pattern and placed on 1.27 cm centres. As the participant walks across the walkway, the sensors detect the foot contacts. The spatio-temporal parameters are calculated via the number of sensors activated, the distance between the activated sensors and the time of activation/deactivation. This information was transferred to a laptop computer, where application software processed the raw data into footfalls and spatio-temporal parameters (GAITRite Manual).

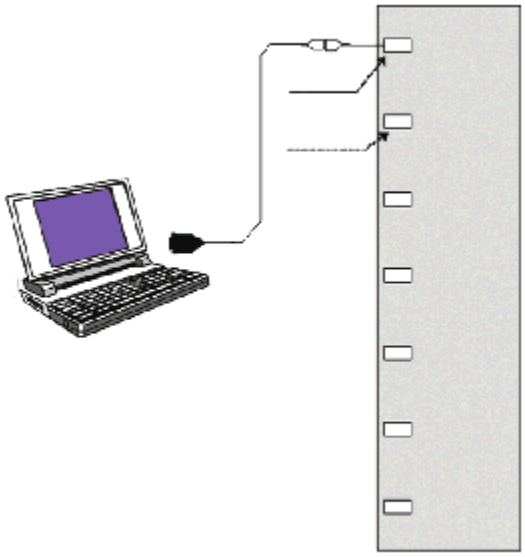


Figure 5.2: Schematic diagram of GAITRite walkway

(Adapted from: http://www.gaitrite.com/Publications/GAITRite_files/GAITRite_slide1/slide9.html)

5.2.3.2 Gait Analysis Parameters & Definitions (GAITRite Operating Manual)

The gait parameters used for this study and their GAITRite definitions were:

(a) Stride Velocity – Obtained after dividing stride length by stride time.

(b) Stride Length – Measured on the line progression between the heel points of two consecutive footfalls of the same foot (left to left, right to right).

(c) Stride Time – The time measured to complete one stride.

(d) Stance Time – The time elapsed between the first contact and the last contact of the foot for two consecutive footfalls of the same foot (Figure 5.3).

(e) Single Support Time – The time elapsed between the last contact of the current footfall to the first contact of the next footfall of the same foot. During this time, the opposite foot is in the swing phase. Therefore the single support time is equal to the swing time of the opposite foot (Figure 5.3).

(f) Double Support Time – Measured as the time elapsed between first contact of the current footfall and the last contact of the previous footfall, added to the time elapsed between the last contact of the current footfall and the first contact of the next footfall (Figure 5.3).

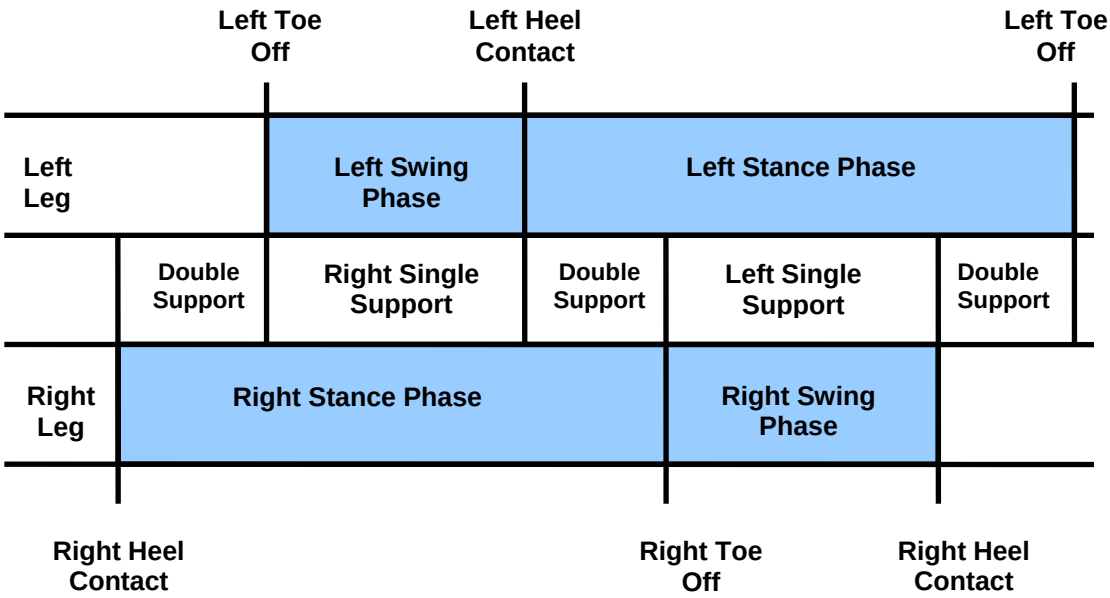


Figure 5.3: The temporal phases of the gait cycle
(Adapted from Whittle, 2002)

5.2.3.3 Baseline & Post-Intervention Gait Measurements

The baseline and post-intervention gait measurements consisted of participants walking across the GAITRite walkway for ten fast walks. Participants were requested to walk across the walkway at a "fast but comfortable walking speed", where they felt confident of not losing balance. Participants wore flat soled shoes that they would 'typically' walk in. In order to reduce acceleration and deceleration over the GAITRite, participants initiated and ceased walking two metres before and after the mat. Ten trials were recorded and averaged for subsequent analysis.

The baseline measures were collected one week prior to beginning the intervention, and the post-intervention gait measurements were recorded at the completion of the final WBV session. The baseline measures were taken during the final health screening stage.

5.2.3.4 Gait Analysis During the Intervention Period

The gait of each participant was assessed immediately at the end of each WBV session (Figure 5.4). The gait analysis during this period replicated the protocol for the baseline and post-intervention tests. During the intervention, participants also walked over the GAITRite at a "fast but comfortable walking speed", where they felt that they were confident of not losing balance. Participants wore the same walking shoes that they wore for the baseline sessions. In order to reduce the confounding effect from acceleration and deceleration over the GAITRite, participants initiated and ceased walking two

metres before and after the mat. Ten trials were recorded and averaged for subsequent analysis.

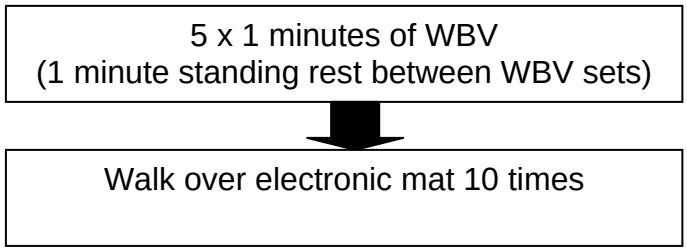


Figure 5.4: Intervention and testing procedures for WBV sessions one to nine

5.2.4 Intervention

During WBV participants were requested to stand in a squat posture approximately 30° from the vertical. This method has been demonstrated to increase skeletal muscle activity (Delecluse et al., 2003). Bipolar surface EMG has previously been used to examine muscle activity in the squat posture for WBV and placebo conditions. Individuals standing in the squat posture during WBV have been found to have elevated rectus femoris and gastrocnemius activity compared to individuals who stand in the same position for the placebo condition (Delecluse et al., 2003). To ensure that the participant's bodyweight was distributed on the WBV plate and not the handlebars, they were requested to hold onto the handlebar but not place their bodyweight onto it.

WBV was performed with one minute 'on' and one minute 'off'. The participants training volume progressed from four times one minute exposures in the first week to five times one minute exposures in the subsequent weeks.

To introduce WBV to the participants, the intensity over the initial two sessions was relatively low (Table 5.1). The frequency pattern used was similar to that employed by Bruyere et al., (2005) for the first session but was progressively increased over the following nine sessions (Bautmans et al., 2005; Roelants et al., 2004; Russo et al., 2003). Participants completed the WBV sessions in the order outlined in table 5.1. If they were unable to attend a WBV session, they completed the session they missed during the following session of attendance. During this session, they only performed the session they had missed.

The participants were requested to wear the same pair of walking shoes for all WBV sessions. They were also asked regularly whether there were any acute side-effects or any side-effects they may have felt during, after or between WBV sessions.

Table 5.1: The Protocol Used for the Three Weeks of WBV

	Monday	**Wednesday**	**Friday**
Week One	**Frequency:**	**Frequency:**	**Frequency:**
	1^{st} min = 10 Hz	1^{st} min = 10 Hz	1^{st}-4^{th} min = 20 Hz
	2^{nd} min = 20 Hz	2^{nd} min = 20 Hz	**Amplitude:** 0.5 mm
	3^{rd} min = 10 Hz	3^{rd} min = 20 Hz	**Acceleration:**
	4^{th} min = 20 Hz	4^{th} min = 20 Hz	1^{st}-4^{th} min = 7.9 $m.s^{-2}$
	Amplitude: 0.5 mm	**Amplitude:** 0.5 mm	
	Acceleration:	**Acceleration:**	
	1^{st} and 3^{rd} min = 2 $m.s^{-2}$	1^{st} min = 2 $m.s^{-2}$	
	2^{nd} and 4^{th} min = 7.9 $m.s^{-2}$	2^{nd}, 3^{rd} and 4^{th} min = 7.9 $m.s^{-2}$	
Week Two	**Frequency:**	**Frequency:**	**Frequency:**
	1^{st}-5^{th} min = 20 Hz	1^{st}-5^{th} min = 20 Hz	1^{st}-5^{th} min = 20 Hz
	Amplitude: 0.5 mm	**Amplitude:** 0.5 mm	**Amplitude:** 0.5 mm
	Acceleration:	**Acceleration:**	**Acceleration:**
	1^{st}-5^{th} min = 7.9 $m.s^{-2}$	1^{st}-5^{th} min = 7.9 $m.s^{-2}$	1^{st}-5^{th} min = 7.9 $m.s^{-2}$
Week Three	**Frequency:**	**Frequency:**	**Frequency:**
	1^{st}-5^{th} min = 25 Hz	1^{st}-5^{th} min = 25 Hz	1^{st}-5^{th} min = 25 Hz
	Amplitude: 0.5 mm	**Amplitude:** 0.5 mm	**Amplitude:** 0.5 mm
	Acceleration:	**Acceleration:**	**Acceleration:**
	1^{st}-5^{th} min = 12.3 $m.s^{-2}$	1^{st}-5^{th} min = 12.3 $m.s^{-2}$	1^{st}-5^{th} min = 12.3 $m.s^{-2}$

5.2.5 Statistical Analyses

Statistical analyses were performed on sessions 1, 4 and 8 of the WBV intervention. Although it was understood that a MANOVA may be used to identify changes in multiple dependent variables, it was decided that it was not appropriate for this research. To have adequate degrees of freedom for MANOVA, it is recommended that there is a ratio of at least three subjects per group to each dependent variable (Vincent, 1995). For this research there was a ratio of seven participants to six dependent variables. The effect of the WBV on gait parameters was therefore analysed by repeated measures ANOVA (GLM) based upon independent statistical advice which was sought outside the university. A Kolmogorov-Smirnov test was performed to test for normality and based upon results from this test, normality was assumed. To avoid a type I error, sphericity was tested with Mauchly's test for sphericity (Portney & Watkins, 2000). Where the assumption of sphericity was violated, an adjustment was made using the Greenhouse-Geiser (GG) correction (Portney & Watkins, 2000). After obtaining a significant F-value, Helmert contrast analyses were performed, based upon the design of previous time-series research (Rauramaa et al., 2004) and advice from the statistician. Helmert contrast analysis detects patterns over time by comparing the mean of the dependent variable with successive time points, indicating the time point of the intervention effect (Crowder & Hand, 1990; Rauramaa et al., 2004). Where differences were found, partial eta squared was used to assess the effect size of WBV on that parameter. Analyses were performed using the

SPSS 12.0.1 for Windows statistical package. Significance levels were set at $p<0.05$.

5.3 Results

5.3.1 Recruitment

Participants were recruited after a presentation to a ladies auxiliary meeting. Nine ladies demonstrated original interest and all participated in the pre-screening process. Following this stage, one participant chose to cease participation as she was concerned about the time commitment. Another participant indicated that she was going into hospital for eye surgery and would miss the last two weeks. This participant was allowed to complete the first week but her data were not included in this study.

5.3.2 Participant Characteristics

The group's characteristics are presented in table 5.2. The mean number of medical conditions was 2.14 with all participants reporting at least one medical condition. The types of conditions included arthritis (n=5), visual impairment (n=4), back pain (n=1) and hypertension (n=3). All participants indicated that they were not taking any medications at the time of the study.

Table 5.2: Baseline Participant Characteristics (n = 7)

Characteristic	Mean ± SD
Age (Yrs)	77.43 ± 2.07
Mass (kg)	76.61 ± 16.40
Stature (m)	1.62 ± 0.08
Number of Medical Conditions	2.14 ± 1.21
Medications	0

5.3.3 Compliance and Adverse Events

Only one participant completed all nine sessions over the three weeks. All other participants fulfilled eight of the nine possible sessions. Therefore session nine was not included in the analyses. Explanations for non-attendance included social events, previously booked appointments and sporting events. There were no adverse events reported by participants over the three weeks, and more specifically, disorders such as osteoarthritis and hypertension were not aggravated. Participants reported the vibration sensation as pleasant and enjoyable.

5.3.4 Gait Parameters

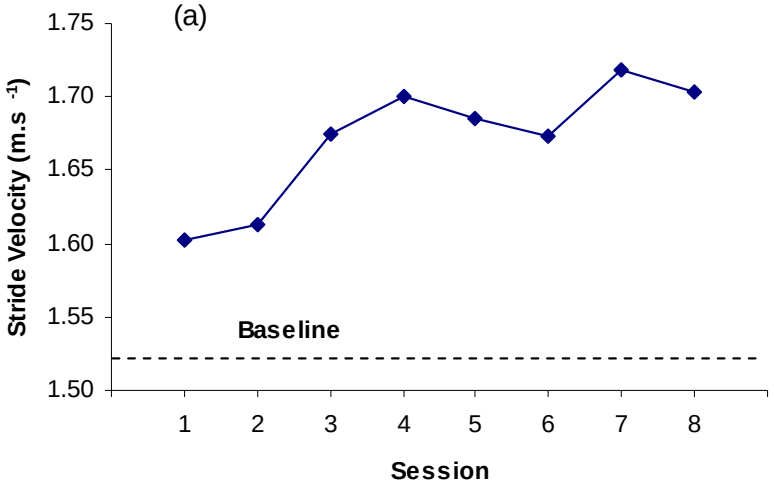
The following sections outline the response of gait parameters over the three weeks. The gait parameters are broken up into six sections. The first section examines stride velocity (5.3.4.1). The following two sections examine the two major parameters of stride velocity - stride length and stride time (5.3.4.2 – 5.3.4.3). The succeeding section examines the stance phase component of stride time (5.3.4.4). The final two sections divide stance time into two sub-phases - single support time and double support time (5.3.4.5 – 5.3.4.6). Swing time was not statistically examined as it constitutes the equivalent time period as the single support time phase and therefore produces the same spatio-temporal calculations.

5.3.4.1 Stride Velocity

There was an increase in stride velocity, compared to baseline, after just one session (p=0.03). Although figure 5.5 displays a linear increase, analysis of sessions one, four and eight (Table 5.3) reported no significant difference (F(1.11, 6.67)=3.79, p=0.10). Figure 5.5 illustrates that by session four, stride velocity had plateaued.

Table 5.3: The effect of WBV on stride velocity ($m.s^{-1}$) for sessions one, four and eight (p=0.10).

Gait Parameter	WBV One	WBV Four	WBV Eight	F
Stride Velocity	1.60	1.70	1.70	3.55
(M,SD)	(0.12)	(0.10)	(0.12)	(1.11, 6.70)



(b)

Session	Mean	SD
Baseline	1.52	0.16
1	1.60	0.12
2	1.61	0.12
3	1.67	0.10
4	1.70	0.10
5	1.69	0.09
6	1.67	0.12
7	1.72	0.11
8	1.70	0.12

Figure 5.5: Time-series graph (a) and time-series table (b) for group mean stride velocity ($m.s^{-1}$) following WBV intervention (n=7)

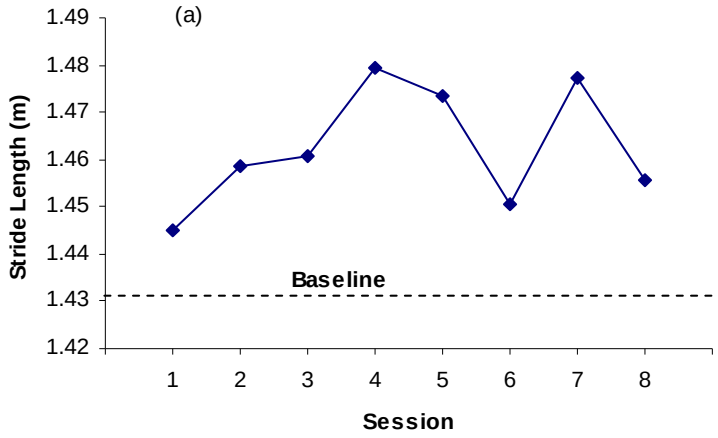
5.3.4.2 Stride Length

Within-subject effects were not found for sessions one, four and eight, suggesting WBV had little to no impact on stride length (Table 5.4).

Table 5.4: The effect of WBV on stride length (m) for sessions one, four and eight (p=0.19)

Gait Parameter	WBV One	WBV Four	WBV Eight	F
Stride Length	1.45	1.48	1.45	2.09
(M,SD)	(0.11)	(0.10)	(0.10)	(1.20, 7.21)

Stride length was not shown to be affected by WBV exercise (Figure 5.6), and presented with an undulating pattern over the three weeks.



(b)

Session	Mean	SD
Baseline	1.43	0.11
1	1.45	0.11
2	1.46	0.09
3	1.46	0.09
4	1.48	0.10
5	1.47	0.09
6	1.45	0.10
7	1.48	0.09
8	1.46	0.10

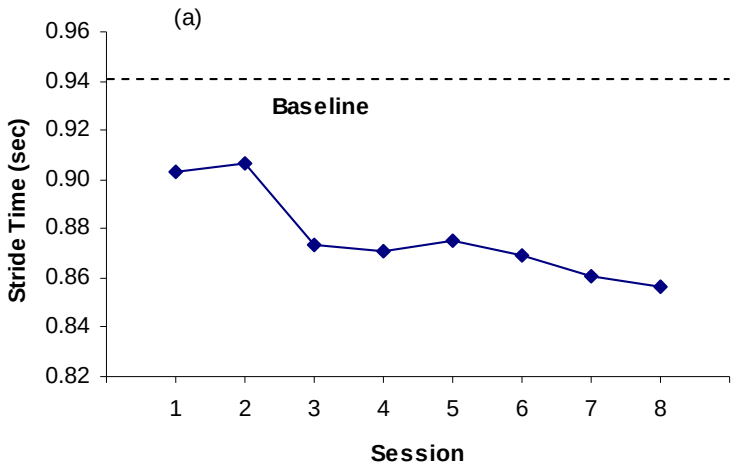
Figure 5.6: Time-series graph (a) and time-series table (b) for group mean stride length (m) following WBV intervention (n=7)

5.3.4.3 Stride Time

Compared to baseline, stride time was significantly reduced after the first WBV session (p=0.02) and continued to decrease (Figure 5.7) over the three weeks (F(1.11, 6.64)=3.79, p=0.04). The largest decrease in stride time occurred between sessions one to four (0.03 seconds). Contrast analysis demonstrated that session one was significantly different to the means of sessions four and eight only (p=0.04). Partial eta squared was calculated as 0.54 indicating a moderate effect of WBV on stride time. Although session eight was less than session four, no significant difference was found.

Table 5.5: The effect of WBV on stride time (sec) for sessions one, four and eight (p=0.04). Contrast significant effects calculated for level 1 vs. later (p=0.04)

Gait Parameter	WBV One	WBV Four	WBV Eight	F
Stride Time	0.90	0.87	0.86	6.43
(M,SD)	(0.02)	(0.03)	(0.04)	(1.11, 6.64)



(b)

Session	Mean	SD
Baseline	0.94	0.04
1	0.90	0.04
2	0.91	0.04
3	0.87	0.04
4	0.87	0.04
5	0.88	0.04
6	0.87	0.06
7	0.86	0.05
8	0.86	0.05

Figure 5.7: Time-series graph (a) and time-series table (b) for group mean stride time (sec) following WBV intervention (n=7)

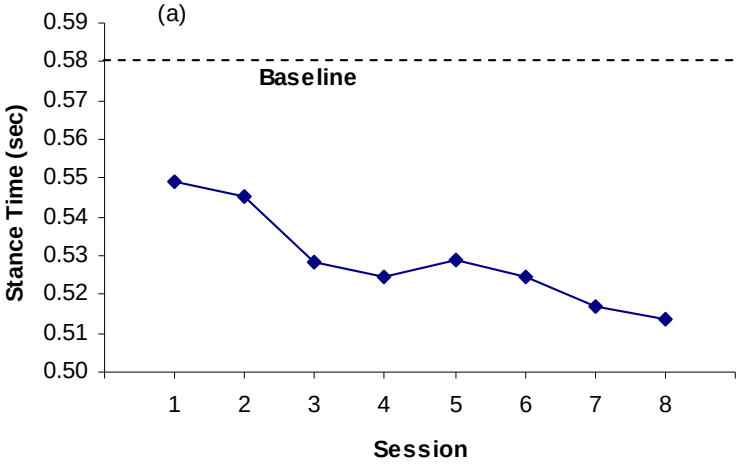
5.3.4.4 Stance Time

Stance time displayed a decrease from baseline to the completion of the first WBV session that was borderline significant (p=0.06) and demonstrated a trend to decrease linearly (Figure 5.8) from sessions one to eight (F(1.07, 6.45)=6.81, p=0.04). Contrast analysis indicated session one to be significantly different to the mean of sessions four and eight only (p=0.02). Partial eta squared indicated that 61% of the variance could be explained by the WBV. No statistical difference was found for sessions four and eight.

Table 5.6: The effect of WBV on stance time (sec) for sessions one, four and eight (p=0.04). Contrast significant effects calculated for level 1 vs. later (p=0.02)

Gait Parameter	WBV One	WBV Four	WBV Eight	F
Stance Time	0.55	0.53	0.51	6.81
(M,SD)	(0.02)	(0.03)	(0.04)	(1.07, 6.45)

(a)



(b)

Session	Mean	SD
Baseline	0.58	0.03
1	0.55	0.02
2	0.55	0.03
3	0.53	0.03
4	0.52	0.03
5	0.53	0.03
6	0.52	0.04
7	0.52	0.04
8	0.51	0.03

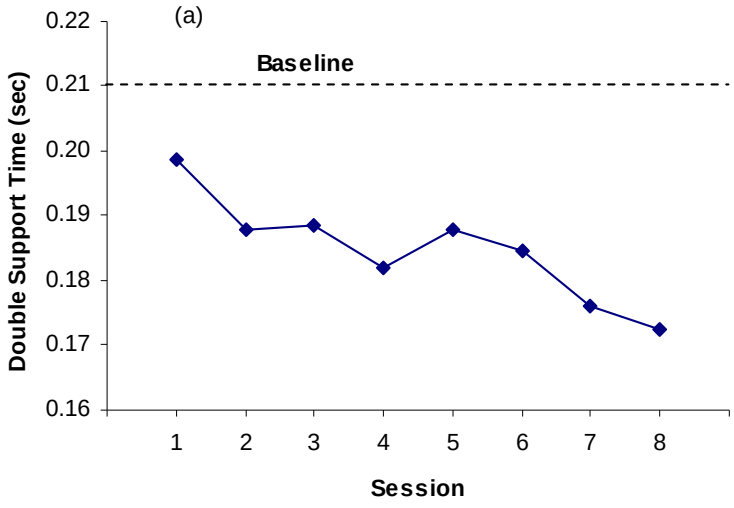
Figure 5.8: Time-series graph (a) and time-series table (b) for group mean stance time (sec) following WBV intervention (n=7)

5.3.4.5 Double Support Time

Double support time did not show a significant difference between baseline and session one. Throughout the three weeks, this parameter decreased linearly and moved towards borderline significance ($F(1.16, 6.95)=3.98$, $p=0.08$). Although significance was not found, figure 5.9 displays a visible decrease in double support time. A contrast analysis was therefore performed which found a significant difference for session one vs. sessions four and eight ($p=0.03$). Partial eta squared indicated a moderate effect, whereby 57% of the change in double support time could be explained by the WBV intervention.

Table 5.7: The effect of WBV on double support time (sec) for sessions one, four and eight ($p=0.08$). Contrast significant effects calculated for session one vs. later ($p=0.03$)

Gait Parameter	WBV One	WBV Four	WBV Eight	F
Double Support Time (M,SD)	0.20 (0.02)	0.18 (0.02)	0.17 (0.03)	3.98 (1.16, 7.00)



(b)

Session	Mean	SD
Baseline	0.21	0.03
1	0.20	0.03
2	0.19	0.04
3	0.19	0.03
4	0.18	0.02
5	0.19	0.04
6	0.18	0.04
7	0.18	0.04
8	0.17	0.03

Figure 5.9: Time-series graph (a) and time-series table (b) for group mean double support time (sec) following WBV intervention (n=7)

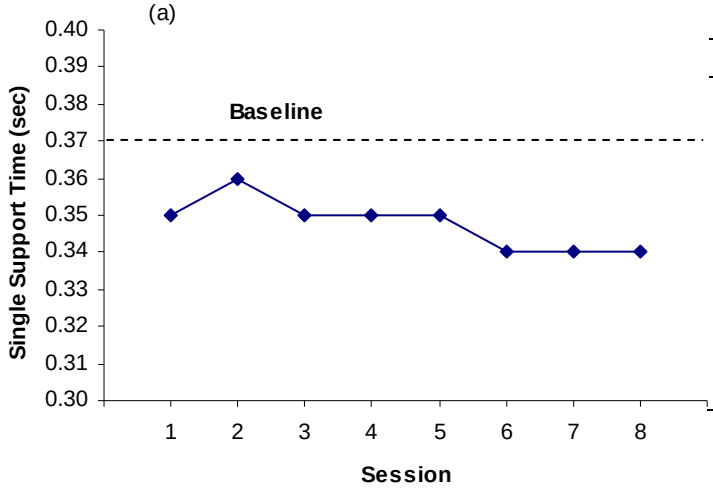
5.3.4.6 Single Support Time

Although single support time decreased from baseline to session one, the difference was not significant. No significant differences were found for sessions one, four and eight (Table 5.8). This variable showed very little to no change over the three weeks of WBV (Figure 5.10).

Table 5.8: The effect of WBV on single support time (sec) for sessions one, four and eight (p=0.11)

Gait Parameter	WBV One	WBV Four	WBV Eight	F
Single Support Time (M,SD)	0.35 (0.03)	0.34 (0.02)	0.34 (0.03)	2.90 (2,12)

(a)



(b)

Session	Mean	SD
Baseline	0.37	0.03
1	0.35	0.03
2	0.36	0.04
3	0.35	0.03
4	0.35	0.02
5	0.35	0.04
6	0.34	0.04
7	0..34	0.04
8	0.34	0.03

Figure 5.10: Time-series graph (a) and time-series table (b) for group mean single support time (sec) following WBV intervention (n=7)

5.3.5 Power Analyses

As was outlined above, stride velocity and double support time suggested changes following the WBV intervention. It was felt that the lack of significance may have been due to the small sample size. Power analyses were therefore performed to determine the minimum sample size for future research of these variables (Table 5.9). The calculations for the estimation of sample size are presented in Appendix I and Appendix J. For a significant difference to be calculated for each of these variables, a minimum of eight participants was necessary for the major study.

Table 5.9: Minimum participants required for statistical significance for stride velocity and double support time

Variable	F Index	Minimum participants required for statistical significance
Stride Velocity	0.77	8
Double Support Time	0.82	8

5.4 Pilot Study Discussion

There has been a great deal of variability reported across the literature debating what constitutes a fast walking velocity for elderly women in their eighth decade. At baseline, participant stride velocity (1.52 $ms.^{-1}$ ± 0.16) in this study was generally faster than that reported by most studies (Brach et al., 2001; Elble et al., 1991; Leiper & Craik, 1991; Öberg et al., 1993), although faster walking velocities (1.74 $m.s^{-1}$) have been reported (Bohannon, 1997). Explanations for these differences include sample size, participant characteristics (e.g. height, limb length) and inclusion/exclusion criteria. It must be noted that it was not a goal of this study to compare walking speeds to other population groups, but it should be observed that participants walked faster than has generally been reported in the literature. These individuals may have been walking nearer to their maximum capacity, as determined by their physical attributes, and therefore their walking velocity may not have been as sensitive to the effect of the intervention.

Stride velocity and stride time showed significant differences between baseline and session one. Although it is possible that one session of WBV had such an impact on fast-walking, it is more likely that the results were due to the motivation of the participants. Nevertheless, the first session demonstrated that WBV at the intensity prescribed in the present study did not have an acute negative outcome on fast walking velocity of elderly women.

This pilot study is the first to perform a time-series analysis examining the response of spatio-temporal parameters of fast gait to WBV. The graph of

stride velocity showed an upward slope over the three weeks. Notably, a 10 $cm.s^{-1}$ difference was recorded between WBV one and WBV eight. Although these changes were not significant (p=0.10), either a larger sample size or longer intervention duration may have produced a significant difference. An estimate of sample size from the data produced from this study indicates that sample size of eight would have produced a significant difference (Table 5.9).

It is generally accepted that the age-associated decrease in stride length contributes to the slower stride velocity (Lord et al., 1996; Elble et al., 1991; Hageman & Blanke, 1986; Öberg et al., 1993). It was therefore expected that stride length would increase following WBV, and contribute to an increased stride velocity. This study demonstrated that WBV had no functional impact on the stride length of elderly females. It is understood that decreasing power of the triceps surae contributes to the decreasing stride length that is associated with increasing age and therefore contributes to the decreasing stride velocity (Judge et al., 1996; McGibbon & Krebs, 1999; Riley et al., 2001; Winter et al., 1990). Although WBV has augmented strength and power of the knee extensors of post-menopausal women (Roelants et al., 2004), and demonstrated increased EMG activity of the triceps surae during exposure (Torvinen et al., 2002b), a study is yet to demonstrate strength or power increases of the latter muscle group. It would be expected that if WBV had a positive impact on the functional power of the triceps surae, stride length would increase, but this pilot study indicates that this is probably not the case.

The increased stride velocity was related to decreases in stride time. Stride time is comprised of swing time and stance time. Swing time showed no change throughout the three weeks and therefore changes in stride time and stride velocity were linked to changes in stance time. Stance time can be broken down into two further components, single support time and double support time. The balance between these two variables shifts with age, whereby single support time decreases (Judge et al., 1996; Murray et al., 1969) and double support time increases (Judge et al., 1996). The final period of single limb support consists of the largest amount of work and power produced by the triceps surae and is the most significant contributor to forward propulsion (Kepple, Siegel & Stanhope, 1997), yet no change in single support time was recorded. The failure of single support time to decrease, combined with no change in stride length, further supports the argument that this intervention did not increase the functional power of the triceps surae during fast gait.

As previously indicated, the coinciding phase to the stance period is double support time. Double support time is considered to contribute to balance control (Gabell & Nayak, 1984; Maki, 1997; Winter et al., 1990). Although this variable did not demonstrate a significant difference ($p=0.08$), reference to figure 5.9 displays a downward slope of double support time. Similar to stride velocity, a larger scale study with a larger sample size may have produced a significant difference. An estimate of sample size demonstrated that a sample size of eight would have produced a significant difference (Table 5.9). Exposure to WBV over a longer duration of weeks may have also produced a

significant difference for this parameter. It is possible that the decreasing double support time was a biomechanical response to increased stride velocity (Maki, 1997; Winter et al., 1990). Conversely, increased stability in single limb support may have reduced the need to spend time standing in the double support phase and therefore increased stride velocity.

The time-series method of analysis provided insight on the response of fast gait to WBV over the three weeks. For those variables that indicated a significant change, the impact was within the first 4 sessions. Stride velocity, for example, increased from 1.60 $m.s^{-1}$ to 1.70 $m.s^{-1}$ from sessions one to four. This change may have been due to systematic errors such as a practice effect or simply the effect of a placebo. For variables such as stance time and double support time, there was a trend to continue to decrease from session four to eight while stride velocity appeared to plateau. This study was three weeks in length, therefore due to this limited duration, it is difficult to determine whether this plateau would have continued, or whether stride velocity would have shown further increases or even decreases. A longer intervention period may produce a larger positive effect on gait and would highlight whether a plateau effect really was occurring.

In the present study, no adverse effects were reported by the participants. Individuals described the experience as pleasant and enjoyable, which possibly explains the high compliance. The intensity of the vibration (10-25 Hz; 0.5 mm; 2.0 $m.s^{-2}$-12.3 $m.s^{-2}$) was lower than reported in previous studies, reducing the potential for harm. For the small number of studies to have

investigated WBV for older adults, frequencies ranging between 10 to 40 Hz and amplitudes varying between 1.5 to 5 mm have been used (Bryuere et al., 2005; Roelants et al., 2004).

The absence of a control group is a major limitation to this study. It is impossible to emphatically state that WBV had an effect on any of the analysed gait parameters. The large change after the first session was possibly a placebo response, but without a control, conclusions cannot be made. The linear trends that were exhibited over the three weeks may have also been due to a placebo response, or may have been the result of a practice effect from walking fast at each session. Future time-series research should include a control group that performs the same fast walks as the WBV intervention group to identify whether a practice effects occurs.

A final observation for future investigation was the perceived benefits reported by participants throughout this study. The majority of this cohort indicated that they felt the intervention had a positive impact on their functional mobility. Statements from participants included, “It makes my legs feel as though they are alive”, “I was able to pull weeds from out of my garden and get up without someone assisting me”, and “I feel as though it gives me a kick-start for the day”. Further research should investigate the impact WBV has on self-efficacy.

The perceived benefits may explain the high compliance over the three week period. Although only one participant attended all nine sessions, all other

participants completed eight of the nine possible sessions. Explanations for missing the WBV session were for previously organised commitments (e.g. voluntary service) or sporting events (i.e. lawn bowls).

Summary

In summary, this research set out to examine the response of elderly female gait to WBV of relatively low magnitude acceleration and determine the merit of performing a larger scale study. Based on the findings of this pilot, WBV intensities consisting of 20-30Hz and an amplitude of 0.5 mm (2.0 $m.s^{-2}$-12.3 $m.s^{-2}$) appear to be of sufficient stimulation to increase the fast walking velocity of elderly females over a three week period and warrants further investigation with larger participant numbers and a longer duration. Time-series analysis demonstrated that the largest response occurred during the initial intervention period (sessions one to four). It is possible that long-term WBV will not provide progressive benefits. A study of longer duration is necessary to provide further information. There were no adverse effects from WBV at this intensity and compliance was high over a three week period. It is recognised that conclusions drawn from this study should be cautiously interpreted, as there was no control group and a small sample was used. The major study should aim to rectify these limitations.

6.0 Overview

This cross-over design study examined the fast walking of elderly females following WBV. Participants were requested to attend three 15 minute sessions a week over a twelve week period. Group placebo/WBV (n=12; 77.3±6.2 years) was exposed to a placebo intervention for weeks 1-6, and a WBV intervention for weeks 7-12. Group WBV/placebo (n=10; 75.6±7.9 years) was exposed to a WBV intervention for weeks 1-6, followed by a placebo intervention for weeks 7-12. During the WBV period, the maximum plate acceleration ranged between 2.0 $m.s^{-2}$ to 24.7 $m.s^{-2}$ (0.5 to 1.0 mm; 10 - 30 Hz) and the intervention duration progressed from four times one minutes of WBV (or placebo) to five times one minute efforts, interspersed with one minute recovery periods. At the end of each WBV or placebo session fast walking gait was assessed. Both groups demonstrated linear increases in walking velocity during the WBV period. During the placebo periods, neither group exhibited improvements in walking velocity. Group WBV/placebo maintained their improved walking velocity during the detraining period (weeks 7-12). Temporal gait changes (e.g. stride time) were the major contributors to the faster walking speeds. Although group WBV/placebo demonstrated similar patterns to the pilot study for double support and single support times, group placebo/WBV failed to do the same. The activity-specific and balance confidence scale (ABC) did not show any change in perceived balance confidence for either group following any of the intervention periods. This research demonstrated that low frequency-low amplitude WBV does increase the walking velocity of community dwelling elderly females.

6.1 Introduction

The pilot study indicated promising results, whereby stride velocity displayed an upward slope over eight training sessions. These increases in stride velocity were shown to be associated with decreases in certain temporal components of gait, specifically, stride time, stance time and double support time. There were a number of limitations to this pilot work that make conclusions difficult to specify. The most obvious and significant limitation is the omission of a control group. It is possible that participants improved their walking speed due to the Hawthorne effect, that is, participants improved performance based upon an expectancy of the outcome (Portney & Watkins, 2000). They may also have improved due to the fast walks that were performed at each session. Although previous research has demonstrated 12 weeks of fast walking does not improve the fast walking ability of elderly men (Paillard, Lafont, Costes-Salon, Riviere & Dupui, 2004), it cannot be discounted that a training effect occurred in the elderly females who participated in this study. The major objective of this study was to account for this. A control group was therefore asked to perform the same fast walks as the WBV intervention group.

The pilot study was conducted over a three week period. A longer term study would identify whether the improvements would continue or plateau. Over three weeks, the participants demonstrated excellent compliance. All of the participants completed at least eight of the nine (89 per cent) possible sessions. Another objective of this study was to examine compliance to WBV over a longer duration. Previous reports of the compliance of older adults to

WBV have ranged between 73 and 96 per cent over six week periods (Bautmans et al., 2005; Bruyere et al., 2005). These studies have conducted the WBV intervention program at the residence of nursing home patients. Attendance may have been affected by the high level of accessibility to the WBV for these participants. This study will extend the pilot work, as well as the current literature, on the compliance of older adults to WBV by extending the intervention duration to twelve weeks and by recruiting community dwelling older adults.

The relatively high level of compliance that was shown in the pilot study may have been due to the perceived benefits experienced by the participants. Although this feedback was noted, measures were not taken regarding these perceived benefits. This study will attempt to address this issue by using a previously validated measurement tool, the Modified Activities-Specific Balance Confidence scale (Outlined in section 6.2.2.4), to identify the self-perceived quality of balance of participants.

Presently, there has been no reported research examining the long-term maintenance of the effects following WBV. To extend the current understanding of the long-term benefits of WBV on the spatio-temporal parameters of gait, a group of participants will be exposed to WBV for six weeks. Gait data, without WBV, will be collected over the subsequent six weeks to observe any changes in gait that may occur.

Based upon observations and findings from the pilot study and the WBV literature, the following hypotheses were addressed:

(1) That the stride velocity of elderly females exposed to WBV would increase linearly during the intervention period;
(2) That the increased stride velocity of the elderly female gait would be associated with decreases in stride time, stance time and double support time;
(3) That there would be no change in stride length and single support time of elderly female gait following exposure to WBV;
(4) That upon completion of the WBV intervention, the walking velocity of elderly females would decrease over a period of six weeks;
(5) That participants would report an improved self-perceived quality of balance following the WBV intervention; and
(6) That the spatio-temporal gait parameters and self-perceived quality of balance of elderly females would not change over the placebo period.

6.2 Method

6.2.1 Recruitment of Participants

Four methods of recruitment were used for this study. The most successful method was via an article written by a journalist in the Senior Victorian newspaper. A presentation was also made to senior delegates of the Victorian Lawn Bowls Association in the South Eastern Metropolitan area of Melbourne. Articles were also written for a local Returned Soldiers League and Probus club inviting interest from members.

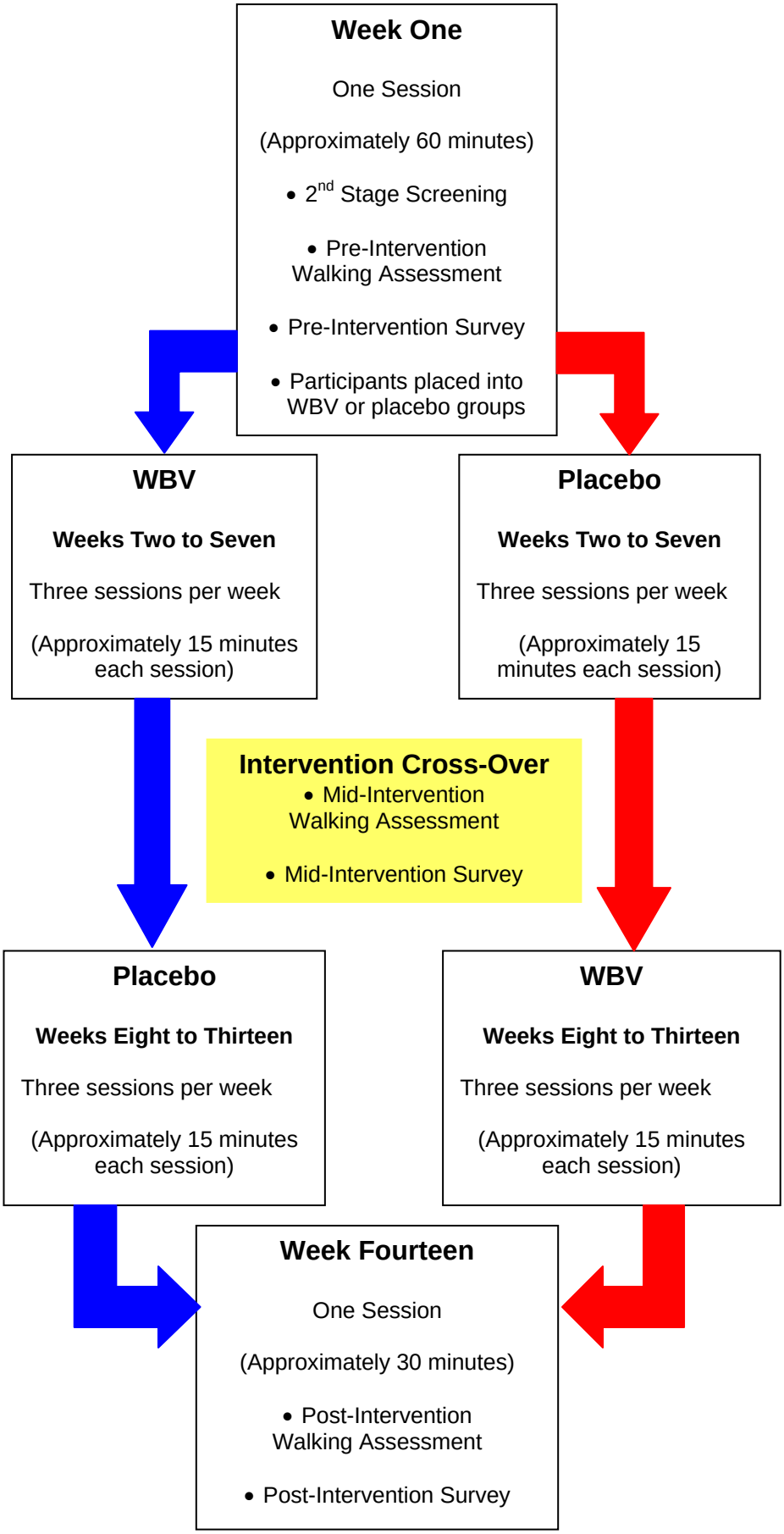
6.2.2 Procedure

The procedure for the study is outlined in figure 6.1. Following a phone conversation initiated by the participants, an introduction pack consisting of a letter to the participant (Appendix G), a letter to their physician (Appendix H) and a health screening form (Appendix E) were mailed to the participant.

6.2.2.1 Health and Medical Screening

After reading the letter to the participants and indicating interest in further participation, individuals were requested to complete the health screening form. Following completion of this form, participants made an appointment with their General Practitioner for a medical screening to determine whether their personal health was judged adequate to participate in the study. The ACU National's School of Exercise Science met all out of pocket expenses for

Figure 6.1: Procedural Outline for weeks one to thirteen.



this consultation. Participants were requested to provide their General Practitioner with the letter to the physician outlining the study protocol and potential benefits for the participant, and were also requested to show the health screening form to the General Practitioner. The General Practitioner was asked to read the health screening form and provide further details on any medical contraindications that may have influenced safe participation in this study. The final request of the General Practitioner was to sign the space provided on the health screening form if the participant was able to safely participate in this study.

6.2.2.2 Baseline and Post-Intervention Assessments

Following consent from the General Practitioner, participants were booked in for the second stage of medical screening and baseline measurements. The second stage of screening was previously outlined in the pilot study (sections 5.2.2.4-5.2.2.6). This stage examined the participant's vision, the proprioceptive control of their lower limbs and their cognition.

The baseline measures were collected one week prior to beginning the first intervention. Participants were requested to attend a one hour session which finalised the health screening, collected baseline gait measurements and had participants complete the ABC-UK surveys (Outlined in section 6.2.2.4). The same measures were performed during the intervention cross-over (mid-study) and post-intervention.

6.2.2.3 Gait Assessment

For logistic purposes, since the group's placebo/WBV and WBV/placebo were tested at different venues, two GAITRite walkways were used. Group placebo/WBV used the same walkway that was used in the pilot study. The specifications for this walkway have previously been outlined in section 5.2.3.1. Group placebo/WBV performed ten fast walks over the walkway. Session data were averaged for subsequent analysis. Due to different walkway specifications (60 cm wide by 720 cm long with a 48x576 sensor grid pattern; 80 Hz sampling rate) and the available area for testing, group WBV/placebo performed five fast walks that were recorded and averaged for subsequent analysis. Both groups were requested to "walk at a comfortable fast walk", where comfortable was defined as a speed with which they felt confident. To minimise any confounding effect of gait acceleration and deceleration over the GAITRite, participants initiated and ceased walking two metres before and after the mat.

6.2.2.4 Survey - The Modified Activities-Specific Balance Confidence Scale (ABC-UK)

The ABC-UK is a 16 item scale which asks participants to rate their confidence of maintaining their balance and steadiness when performing tasks of various difficulties. The tasks listed are progressively more difficult and the participant is asked to rate how they feel about each task in multiples of 10%, ranging from 0% (no confidence) to 100 % (absolutely confident).

The Activities-specific Balance Confidence Scale (ABC) was originally developed in North America to quantify the functional mobility of older adults (Powell & Myers, 1995) The ABC-UK was developed as a British specific scale due to the American-English idiom in the ABC that is unfamiliar to the British population (Parry, Steen, Galloway, Kenny & Bond, 2001). Table 6.1 lists the questions asked in the ABC-UK. Examples of the unfamiliar word translations include “sidewalk” into “pavement”, “mall” into “shopping centre”, and “closet” into “cupboard”. The language used by the ABC-UK is more familiar to the Australian language and was therefore used for this study.

Table 6.1: The Modified Activities-specific Balance Confidence Scale (ABC-UK)

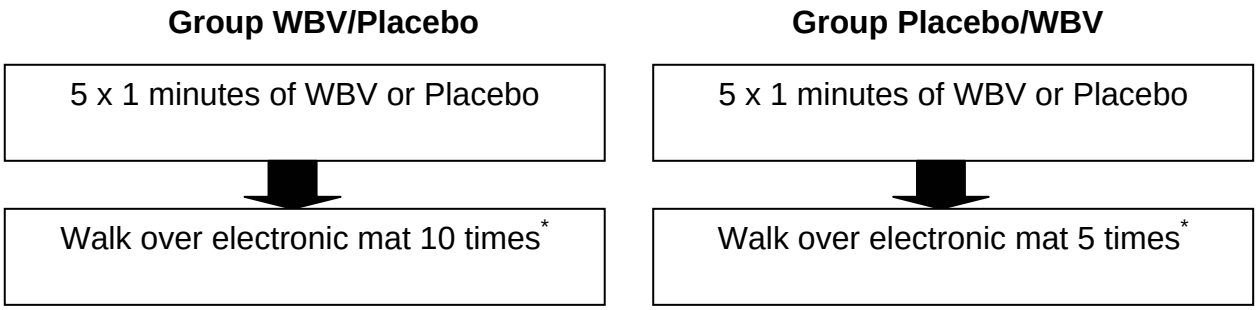
How confident are you that you can maintain your balance and remain steady when you…
1. …walk around the house?
2. …walk up or down stairs?
3. …bend over and pick up a slipper from the floor at the front of a cupboard?
4. …reach for a small tin of food from a shelf at eye level?
5. …stand on your tip toes and reach for something above your head?
6. …stand on a chair and reach for something?
7. …sweep the floor?
8. …walk outside the house to a parked car?
9. … get into or out of a car?
10. …walk across a car park to the shops?
11. …up or down a ramp?
12. …walk in a crowded centre where people walk past you quickly?
13. …are bumped into by people as you walk through the shopping centre?
14. …step onto or off an escalator while holding onto the handrails?
15. …step onto or off an escalator while holding onto parcels such that you cannot hold onto the handrail?
16. …walk outside on slippery pavements?

The ABC-UK has been demonstrated to detect differences in falls related quality of life between fallers and non-fallers and has a test-retest reliability of 0.89 and internal consistency of 0.98. For the present study, the ABC-UK was used to identify the self-perceived quality of balance in participants before and after WBV.

6.2.3 Intervention Program

Figure 6.2 outlines the procedure for the intervention program. Following baseline measurements, participants were placed into one of two groups. Group placebo/WBV began with the placebo for the first 6-wks of the intervention and group WBV/placebo began with WBV. At week seven the groups crossed over, group WBV/Placebo was exposed to the placebo and group Placebo/WBV, was exposed to WBV. At the end of the WBV and placebo intervention sessions, gait measures were recorded for each participant. The total session time for intervention and gait assessment totalled 15 minutes for each participant. Participants were requested to attend intervention sessions, three times a week over a 12 week period.

Figure 6.2: Daily procedure for intervention program



* The explanation for the number of walks was outlined in section 6.2.2.3

6.2.3.1 Whole Body Vibration Intervention

Whole body vibration intensity was progressively increased over the six weeks. Table 6.2 outlines the progression. Participants completed the WBV sessions in this order. If they missed a WBV session, they completed the session they missed during the following session of attendance.

Participants stood on the WBV plate with their knees slightly flexed. To ensure that the participant's bodyweight was distributed on the WBV plate and not the handlebars, they were requested to hold onto the handlebar but not place their bodyweight onto it.

Table 6.2: WBV protocol

	Monday			Wednesday			Friday		
	Freq. (Hz)	**Amp. (mm)**	**Accel. ($m.s^{-2}$)**	**Freq. (Hz)**	**Amp. (mm)**	**Accel. ($m.s^{-2}$)**	**Freq. (Hz)**	**Amp. (mm)**	**Accel. ($m.s^{-2}$)**
Week One									
1st minute	10	0.5	2	10	0.5	2	20	0.5	7.9
2nd minute	20	0.5	7.9	20	0.5	7.9	20	0.5	7.9
3rd minute	10	0.5	2	20	0.5	7.9	20	0.5	7.9
4th minute	20	0.5	7.9	20	0.5	7.9	20	0.5	7.9
Week Two									
1st-5th minute	20	0.5	7.9	20	0.5	7.9	20	0.5	7.9
Week Three									
1st-5th minute	25	0.5	12.3	25	0.5	12.3	25	0.5	12.3
Week Four									
1st-5th minute	30	0.5	17.8	30	0.5	17.8	30	0.5	17.8
Week Five									
1st-5th minute	20	1.0	15.8	20	1.0	15.8	20	1.0	15.8
Week Six									
1st-5th minute	25	1.0	24.7	25	1.0	24.7	25	1.0	24.7

WBV was performed with one minute 'on' and one minute 'off'. The participants' training volume progressed from four one minute exposures in the first week to five times one minute exposures in the subsequent weeks. Similar to previous WBV research with older adults (Bautmans et al., 2005; Roelants et al., 2004; Russo et al., 2003), the training intensity increased progressively over the intervention period. The frequency increased from 10Hz to 30 Hz over the first four weeks. In week five, the amplitude was increased from 0.5 mm to 1.0 mm and the frequency was reduced back to 20 Hz. For the final week of WBV, the amplitude was maintained at 1.0 mm and the frequency was increased to 25 Hz. The acceleration increased from 2 and 7.9$m.^{-2}$ in the first week, to 24.7$m.s^{-2}$ in the final week of this condition.

The participants were requested to wear the same pair of walking shoes for all WBV sessions. They were also asked to report any acute side effects from the WBV and any other adverse effects they may have felt between WBV sessions.

6.2.3.2 Placebo

For the placebo condition, the participants stood on the platform with their knees slightly flexed. The participants could hear the motor running under the platform and could feel a tingling sensation on the soles of their feet. Although the changes in acceleration were negligible, the increase in frequency of vibration for the placebo group coincided with that for the WBV group.

6.2.4 Time-Series Gait Assessment During Intervention Period

As previously stated, the gait of each participant was assessed immediately after the WBV or placebo interventions. Participants were instructed to "walk at a fast comfortable walking speed" over the GAITRite. In order to reduce acceleration and deceleration over the GAITRite, participants initiated and ceased walking two metres before and after the mat. Due to the different GAITRite sizes (outlined in section 6.2.2.3), ten trials were recorded for group placebo/WBV and five trials were recorded for group WBV/placebo, and each group's trials were averaged for subsequent analysis. Participants wore the same walking shoes that they wore for the baseline and the intervention sessions.

6.2.5 Statistical Analyses

Statistical analyses were performed on sessions 1, 5 and 10 of the WBV and placebo interventions for both groups. The choice of statistical analyses was based upon the same argument presented in section 5.2.5 for the pilot research. To test for normality, a Kolmogorov-Smirnov test was performed. Based upon results from this test, normality was assumed. The effect of the WBV and placebo interventions on gait parameters were analysed by two-way repeated measures ANOVA [2(treatment) x 3(session)] (GLM). After obtaining a significant F-value, Helmert contrast analyses were performed to assess the main effects for intervention, session, and interaction effects for intervention*session. Statistical analyses were performed using the SPSS 12.0.1 for Windows statistical package. Significance levels were set at $p<0.05$.

6.3 Results

6.3.1 Recruitment

Thirty-four participants took part in the pre-enrolment evaluations. In total, seven were excluded from this study. Reasons for exclusion included: (a) General Practitioner did not believe WBV was appropriate (n = 1); and (b) Individual did not meet selection criteria (n = 6) (e.g. history of stroke, did not bring signed form from general practitioner, dropped foot).

6.3.2 Participant Characteristics

The group's baseline characteristics are presented in table 6.3. There were no significant differences for age, stature, number of medications or number of medical conditions. The medical conditions reported included arthritis, visual impairment, back pain and hypertension. Group WBV/Placebo had a significantly larger body mass than group Placebo/WBV (p=0.009). Only five individuals indicated that they were taking any medications. All five of these individuals reported that they were currently on one medication only.

Table 6.3: Baseline Participant Characteristics (Mean ± SD)

Characteristic	Group Placebo/WBV (n = 12)	Group WBV/Placebo (n = 10)
Age (Yrs)	77.3 (6.2)	75.6 (7.9)
Mass (kg)	66.7 (13.5)	84.4 (14.6)
Stature (cm)	155.2 (7.6)	157.9 (6.2)
Number of Medical Conditions	1.9 (0.9)	2.4 (1.6)
Number of Medications	0.3 (0.5)	0.1 (0.3)

6.3.3 Compliance and Adverse Events

There were no adverse effects reported by participants in this study that were related to WBV. One participant reported itching on the lower legs and exhibited erythema during the first two WBV sessions but did not present with these during subsequent sessions. No difference in compliance was found between the first and second six weeks of the study for groups placebo/WBV or WBV/Placebo (Table 6.4). Explanations for not attending sessions included colds, viruses, social events, sporting events (lawn bowls) and holidays. Participants who failed to complete at least ten WBV and ten placebo intervention sessions were not included in the analyses.

Table 6.4: Compliance for exercise sessions (Mean ± SD)

Period	Group placebo/WBV (n = 12)	Group WBV/placebo (n = 10)
Weeks 1-6	13.5 (2.0)	14.5 (1.1)
Weeks 7-12	14.6 (1.8)	14.5 (2.3)

Note: Of the 27 participants who began the study, five participants did not complete at least ten sessions (two from group placebo/WBV and three from group WBV/Placebo) and were therefore excluded from the statistical analyses.

6.3.4.1 Stride Velocity for Group Placebo/WBV (Placebo Weeks 1-6 and WBV Weeks 7-12)

The analysis of stride velocity (Table 6.5) identified main effects for intervention ($F(1,11)=12.43$, $p=0.005$) and session ($F(2,22)=5.20$, $p=0.014$). An interaction effect for intervention*session was also observed ($F(2,22)=9.44$, $p=0.001$). For intervention*session interaction, contrast analysis calculated significant differences for sessions one vs. later ($p=0.025$) and session five vs. session ten ($p=0.004$), and respective effect sizes of 0.38 and 0.55 (Figure 6.3).

Table 6.5: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean stride velocity ($m.s^{-1}$) for sessions one, five and ten.

Source	F (df)	Sig.	Partial Eta Squared
Intervention	12.43 (1, 11)	0.005	0.53
Session	5.20 (2, 22)	0.014	0.32
Intervention*session	9.44 (2, 22)	0.001	0.46

Figure 6.3 demonstrates a 2 $cm.s^{-1}$ decrease in stride velocity from placebo sessions one to ten, and a 12 $cm.s^{-1}$ increase in stride velocity from WBV sessions one to ten. Stride velocity was 1.55 ± 0.26 $cm.s^{-1}$ for WBV one, 1.57 ± 0.25 $cm.s^{-1}$ for WBV five and 1.67 ± 0.28 $cm.s^{-1}$ for WBV ten. Stride velocity significantly increased from session's five to ten only ($p=0.007$).

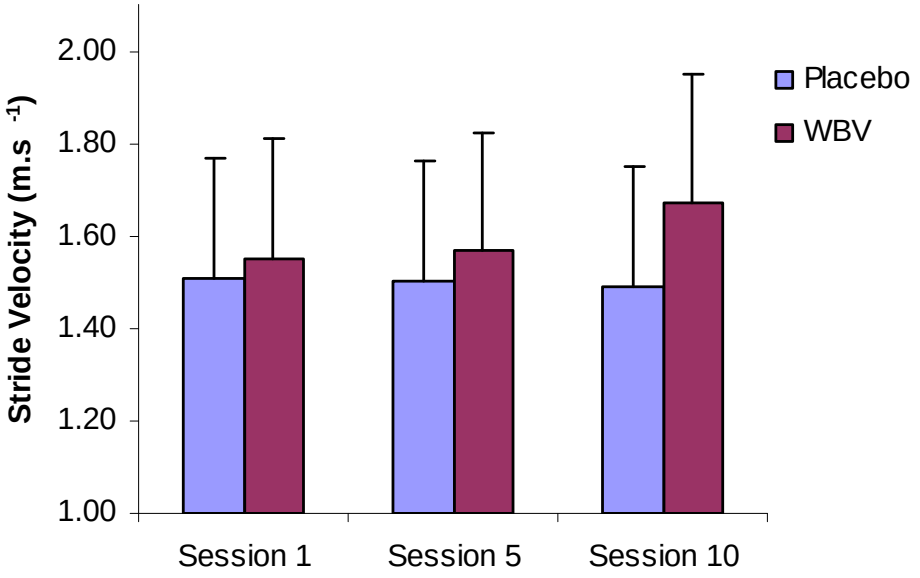
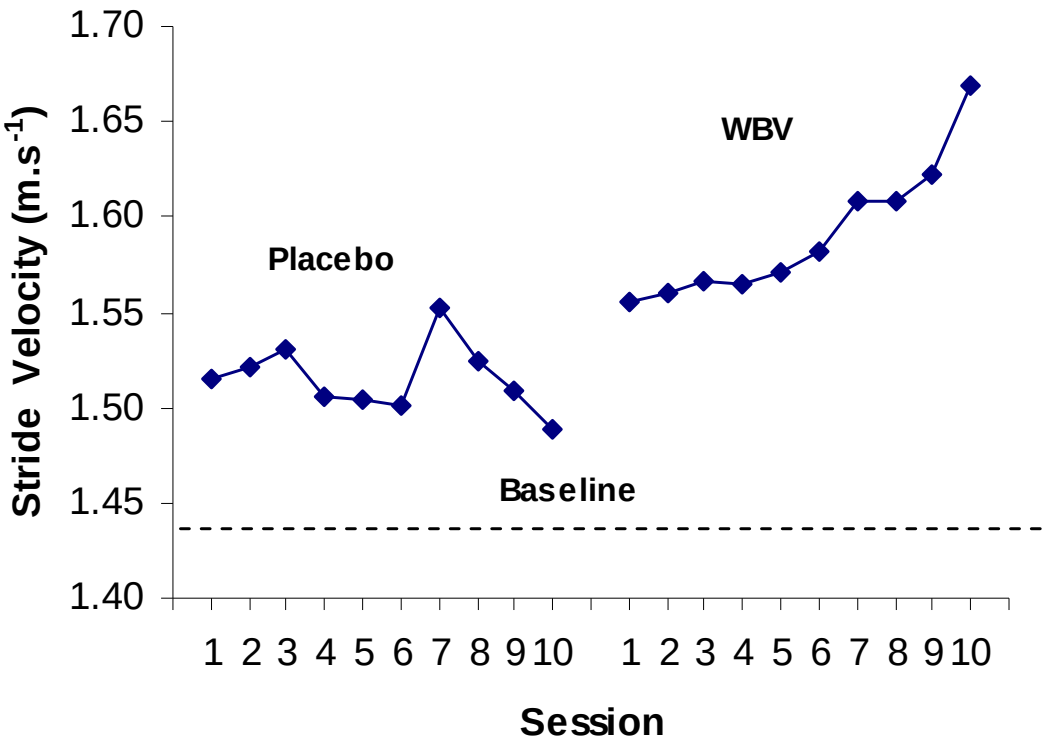


Figure 6.3: Stride velocity following sessions one, five and ten of placebo and WBV interventions.

Note:
- Significant intervention contrast effects (p=0.005)
  Effect size = 0.53
- Significant session contrast effects for session 5 vs. session 10 (p=0.007)
  Effect size = 0.50
- Significant intervention*session contrast effect for session one vs. later (p=0.025)
  Effect size = 0.38
- Significant intervention*session contrast effect for session five vs. ten (p=0.004)
  Effect size = 0.55

Figure 6.4 illustrates the stride velocity pattern recorded for the first ten placebo and WBV intervention sessions. There was no significant difference between the first placebo intervention and baseline walking velocity. Day-to-day stride velocity variability is observable for the placebo intervention. In contrast, stride velocity had an upward trend for the WBV intervention.



	Placebo		Whole Body Vibration	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	1.44	0.21		
1	1.51	0.26	1.55	0.26
2	1.52	0.24	1.56	0.25
3	1.53	0.24	1.57	0.25
4	1.51	0.26	1.56	0.24
5	1.50	0.26	1.57	0.25
6	1.50	0.25	1.58	0.25
7	1.55	0.28	1.61	0.27
8	1.53	0.24	1.61	0.25
9	1.51	0.25	1.62	0.28
10	1.49	0.26	1.67	0.28

Figure 6.4: Time-series graph and table for group placebo/WBV mean stride velocity following placebo and WBV interventions (n = 12)

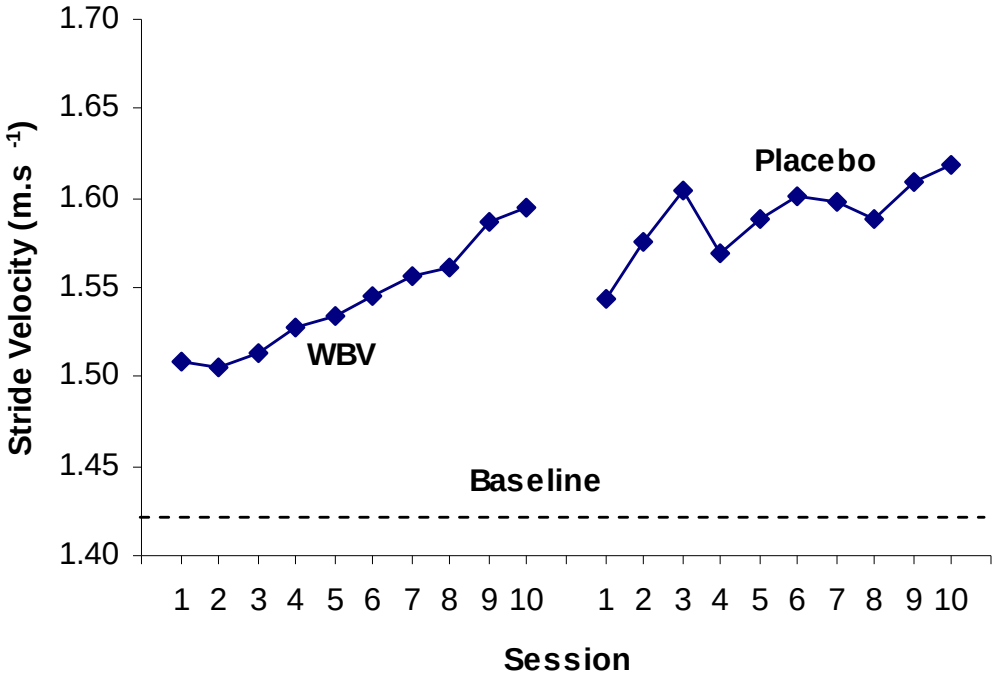
6.3.4.2 Stride Velocity for Group WBV/Placebo (WBV Weeks 1-6 and Placebo Weeks 7-12)

Group WBV/Placebo experienced the WBV in the first six weeks and the placebo intervention during the following six weeks. Following the first WBV session, stride velocity (1.51 ± 0.30 $cm.s^{-1}$) was significantly different (p = 0.001) to the baseline measures (1.42 ± 0.30 $cm.s^{-1}$).

A main effect (Table 6.6) was not found for intervention (p=0.184) or session (p=0.083). Furthermore, an intervention*session interaction effect was not found (p=0.548). Figure 6.5 illustrates a linear increase over the ten weeks of WBV and an undulating pattern during the placebo period.

Table 6.6: Two-way repeated measures ANOVA for the effect of WBV and placebo interventions on mean stride velocity ($m.s^{-1}$) for sessions one, five and ten.

Source	F (df)	Sig.	Partial Eta Squared
Intervention	2.07 (1, 9)	0.184	0.19
Session	3.57 (1.15, 10.36)	0.083	0.28
Intervention*session	0.62 (2, 18)	0.548	0.07



	Whole Body Vibration		Placebo	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	1.42	0.30		
1	1.51	0.30	1.54	0.28
2	1.51	0.28	1.58	0.29
3	1.51	0.29	1.60	0.30
4	1.53	0.29	1.57	0.30
5	1.53	0.29	1.59	0.29
6	1.54	0.27	1.60	0.30
7	1.56	0.28	1.60	0.32
8	1.56	0.30	1.59	0.31
9	1.59	0.28	1.61	0.33
10	1.59	0.31	1.62	0.36

Figure 6.5: Time-series graph for group WBV/Placebo mean stride velocity following WBV and placebo interventions (n = 10)

6.3.4.3 Stride Time for Group Placebo/WBV (Placebo Weeks 1-6 and WBV Weeks 7-12)

Table 6.7 indicates that a significant main effect was found for intervention (F(1,11)=10.70), p=0.007) and session (F(2,22)=4.26, p=0.027). For intervention, a moderate effect size of 0.49 was calculated, while for session a weak effect size of 0.28 was established. An interaction effect (F(2,22)=7.68, p = 0.009) was also found whereby stride time decreased with the number of sessions and according to the type of session. Forty-one per cent of the differences in stride time could be explained by the type of intervention and number of sessions exposed to WBV.

Table 6.7: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean stride velocity ($m.s^{-1}$) for sessions one, five and ten

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	10.70 (1,11)	0.007	0.49
Session	4.26 (2,22)	0.027	0.28
Intervention*session	15.13 (2,22)	0.009	0.41

Stride time was 0.86 seconds at the end of WBV one, 0.86 seconds at the end of WBV five and 0.81 seconds at the end of WBV ten (Figure 6.6). Contrast analysis identified significant differences for session five vs. session ten (p = 0.022) and an effect size of 0.39 (Figure 6.6). Therefore the largest changes in stride time occurred from WBV session's five to ten. It should be noted that

a considerable decrease occurred between sessions nine (0.84 seconds) to ten (0.81 seconds). Contrast analysis for intervention*session determined differences between session one vs. the mean of sessions five and ten (p=0.049) and session five vs. session ten (p=0.004).

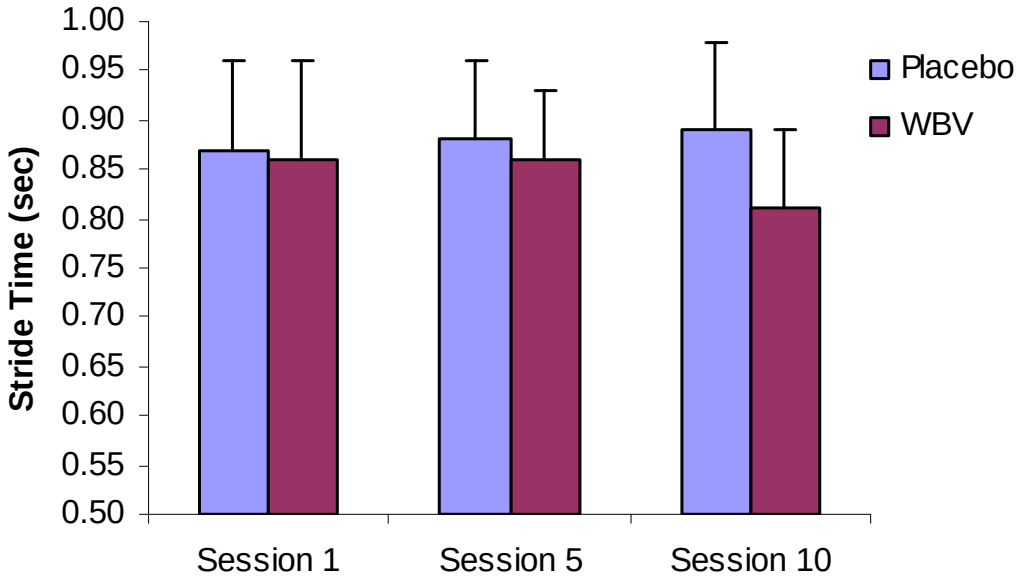
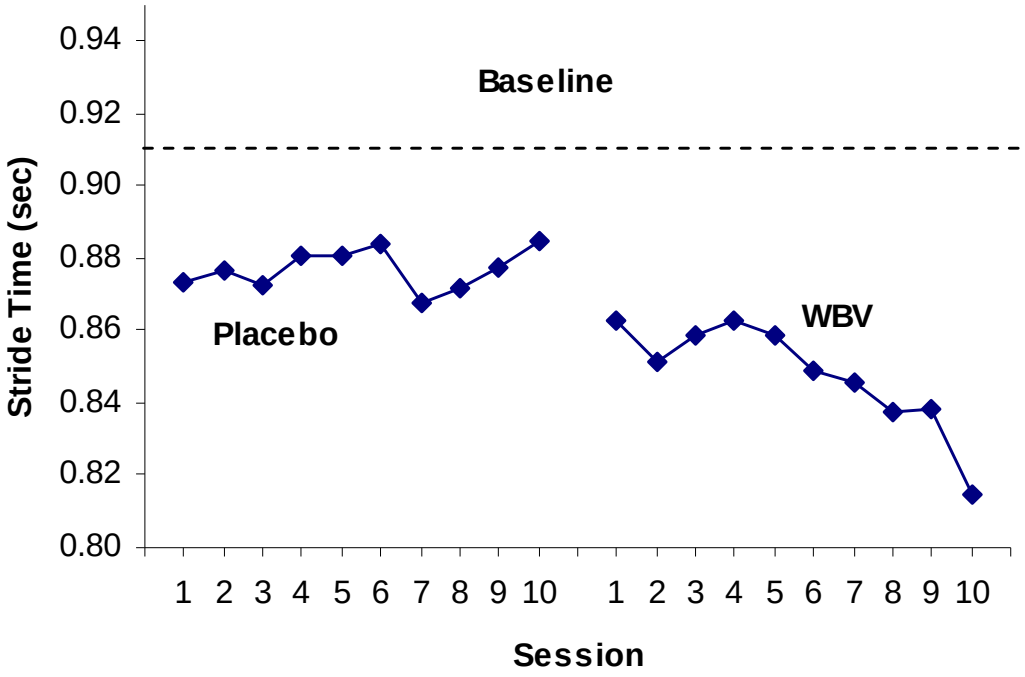


Figure 6.6: Stride time following sessions one, five and ten of placebo and WBV interventions

Note:
- Significant intervention contrast effects (p=0.007)
  Effect size = 0.49
- Significant session contrast effects for session 5 vs. session 10 (p=0.022)
  Effect size = 0.39
- Significant intervention*session contrast effect for session one vs. later (p=0.049)
  Effect size = 0.31
- Significant intervention*session contrast effect for session five vs. ten (p=0.004)
  Effect size = 0.54

For the placebo intervention, there was little change in stride time over the ten intervention sessions (Figure 6.7). During the WBV intervention, stride time decreased in a linear trend over the ten sessions (Figure 6.7).



	Placebo		Whole Body Vibration	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.91	0.08		
1	0.87	0.09	0.86	0.10
2	0.88	0.08	0.85	0.08
3	0.87	0.07	0.86	0.07
4	0.88	0.08	0.86	0.08
5	0.88	0.08	0.86	0.07
6	0.88	0.08	0.85	0.07
7	0.87	0.08	0.85	0.07
8	0.87	0.08	0.84	0.07
9	0.88	0.08	0.84	0.07
10	0.89	0.09	0.81	0.08

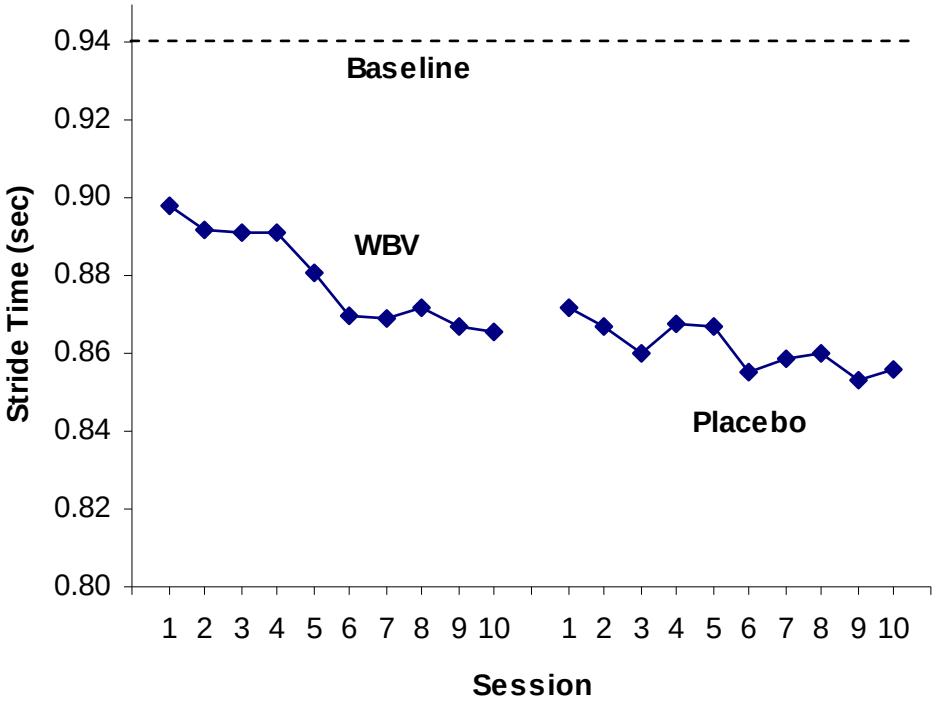
Figure 6.7: Time-series graph and table for group mean stride time following placebo and WBV interventions (n = 12)

6.3.4.4 Stride Time for Group WBV/Placebo (WBV Weeks 1-6 and Placebo Weeks 7-12)

Although stride time demonstrated trends of linear decreases for both WBV and placebo conditions, a main effect was not found for the intervention or session (Table 6.8). The intervention*session interaction also failed to present with an effect.

Table 6.8: Two-way repeated measures ANOVA for the effect of WBV and placebo interventions on mean stride velocity ($m.s^{-1}$) for sessions one, five and ten.

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	3.16 (1,9)	0.109	0.26
Session	3.04 (1.29,11.57)	0.101	0.25
Intervention*session	1.26 (2,18)	0.309	0.12



	Whole Body Vibration		Placebo	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.94	0.09		
1	0.90	0.09	0.87	0.09
2	089	0.08	0.87	0.08
3	0.89	0.08	0.86	0.08
4	0.89	0.09	0.87	0.09
5	0.88	0.09	0.87	0.08
6	0.87	0.09	0.86	0.09
7	0.87	0.08	0.86	0.08
8	0.87	0.09	0.86	0.09
9	0.87	0.8	0.85	0.09
10	0.87	0.10	0.86	0.10

Figure 6.8: Time-series graph and table for group mean stride time following placebo and WBV interventions (n = 12)

6.3.4.5 Stride Length for Group Placebo/WBV (Placebo Weeks 1-6 and WBV Weeks 7-12)

The average stride length was larger (F(1,11)=7.97,p=0.017) for the WBV intervention compared to the placebo intervention (Table 6.9; Figure 6.9). A main effect and interaction effect were not found for session and intervention*session respectively.

Table 6.9: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean stride velocity ($m.s^{-1}$) for sessions one, five and ten.

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	7.97 (1,11)	0.017	0.42
Session	0.61 (2,22)	0.551	0.05
Intervention*session	0.67 (2,22)	0.524	0.06

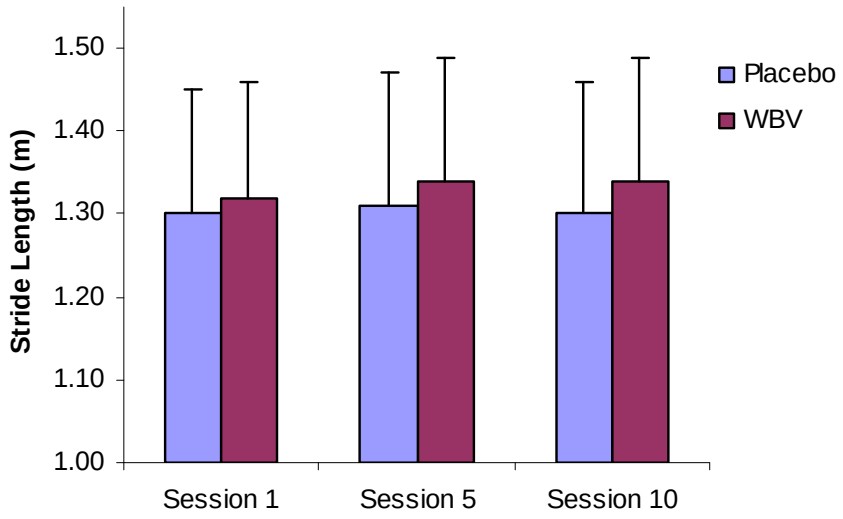
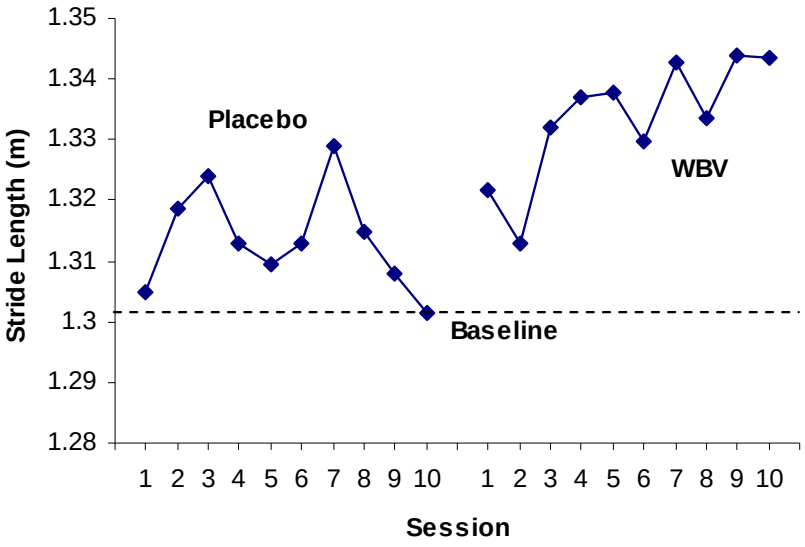


Figure 6.9: Stride length following sessions one, five and ten of placebo and WBV interventions for group placebo/WBV.

Note:
- Significant intervention contrast effects (p=0.017)
  Effect size = 0.42

Stride length exhibited an undulating pattern over the six weeks for the placebo group, returning to baseline levels by session ten (Figure 6.10). Similarly, during the WBV period, stride length displayed an undulating pattern; however, there was an upward trend. Notably, WBV stride length was higher than placebo stride length for the majority of sessions.



	Placebo		Whole Body Vibration	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	1.30	0.14		
1	1.30	0.15	1.32	0.14
2	1.32	0.16	1.31	0.14
3	1.32	0.16	1.33	0.15
4	1.31	0.16	1.34	0.15
5	1.31	0.16	1.34	0.15
6	1.31	0.16	1.33	0.15
7	1.33	0.16	1.34	0.14
8	1.31	0.15	1.33	0.14
9	1.31	0.14	1.34	0.15
10	1.30	0.16	1.34	0.15

Figure 6.10: Time-series graph and table for group mean stride length following placebo and WBV interventions (n = 12)

6.3.4.6 Stride Length for Group WBV/Placebo (WBV Weeks 1-6 and Placebo Weeks 7-12)

For participants who received the WBV intervention during the first six weeks, a main effect was found for session only ($F(2,18)=4.37, p=0.028$). An interaction effect was not found between intervention*session (Table 6.10).

Table 6.10: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean stride velocity ($m.s^{-1}$) for sessions one, five and ten.

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	0.39 (1,9)	0.550	0.04
Session	4.37 (2,18)	0.028	0.33
Intervention*session	1.31 (1.31, 11.76)	0.288	0.13

Although a main effect was found for session, a difference between WBV and placebo sessions could not be clearly discriminated. Contrast analysis calculated a significant difference ($p=0.05$) for session five vs. session ten (Figure 6.11) and a low effect size of 0.36 for number of sessions on stride length. Figure 6.11 illustrates that there was little to no session differences between the WBV and placebo interventions, ranging between 1.33 and 1.36 m for both groups.

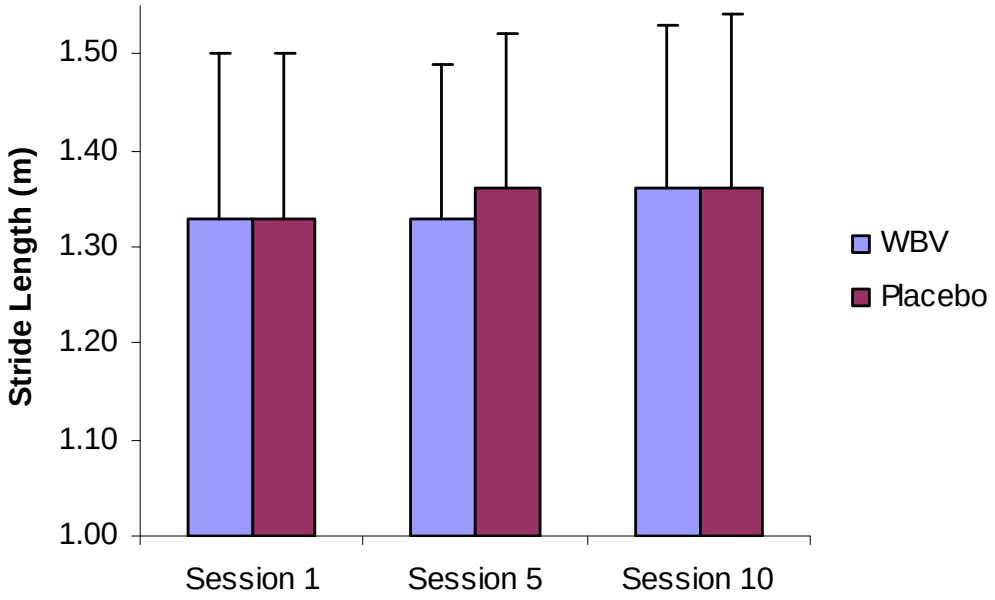
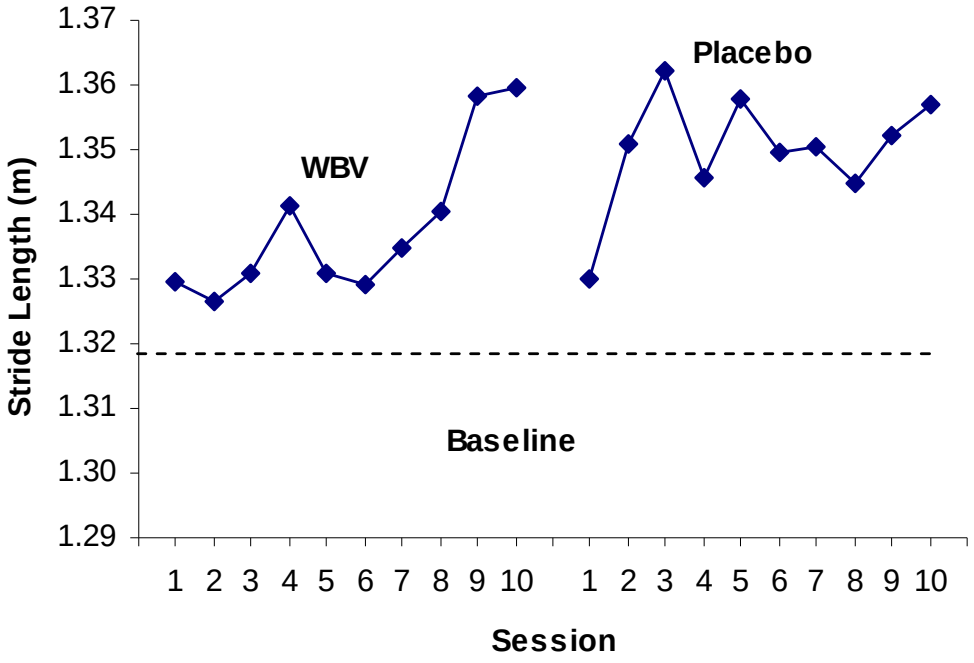


Figure 6.11: Stride length following sessions one, five and ten of WBV and placebo interventions for group WBV/Placebo

Note:
- Significant session contrast effects for session 5 vs. session 10 (p=0.05)
  Effect size = 0.36

Figure 6.12 displays evidently different time-series patterns for the WBV and placebo interventions. Stride length was relatively unchanged for the WBV period until session nine where there was a 2 cm increase. For the placebo, with the exception of the first session, stride length varied between 1.35 to 1.36 m.



	Whole Body Vibration		Placebo	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	1.32	0.18		
1	1.33	0.17	1.33	0.17
2	1.33	0.16	1.35	0.18
3	1.33	0.17	1.36	0.18
4	1.34	0.17	1.35	0.18
5	1.33	0.16	1.36	0.16
6	1.33	0.16	1.35	0.18
7	1.33	0.16	1.35	0.18
8	1.34	0.17	1.34	0.18
9	1.36	0.16	1.35	0.18
10	1.36	0.17	1.36	0.18

Figure 6.12: Time-series graph and table for group mean stride length following placebo and WBV interventions (n = 10)

6.3.4.7 Stance Time for Group Placebo/WBV (Placebo Weeks 1-6 and WBV Weeks 7-12)

Stance time was significantly lower (Table 6.11) for the WBV group ($F(1,11)=19.25, p=0.001$). A main effect was not found for session number. An interaction effect for intervention*session was calculated ($F(1.21,13.33)=6.37, p=0.021$) where stance time decreased for the WBV group as session number increased (Table 6.11). Partial eta squared indicated that the interaction effect was weak, with 37% of the difference being explainable by WBV and session number.

Table 6.11: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean stance time (seconds) for sessions one, five and ten.

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	19.25 (1,11)	0.001	0.64
Session	2.45 (2,22)	0.109	0.18
Intervention*session	6.37 (1.21,13.33)	0.021	0.37

Stance time decreased for the WBV group by 0.03 seconds from sessions one to ten, and was unchanged for the placebo group (Figure 6.13). Stance time was 0.51 ± 0.07 seconds for WBV one, 0.50 ± 0.05 seconds for WBV five and 0.48 ± 0.05 seconds for WBV ten. Contrast analysis identified a significant intervention*session effect (p=0.001) for session five vs. session ten only (Figure 6.13).

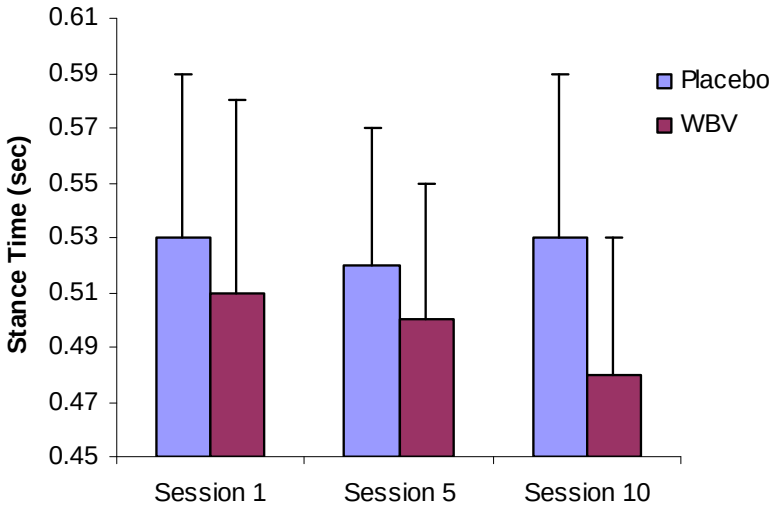
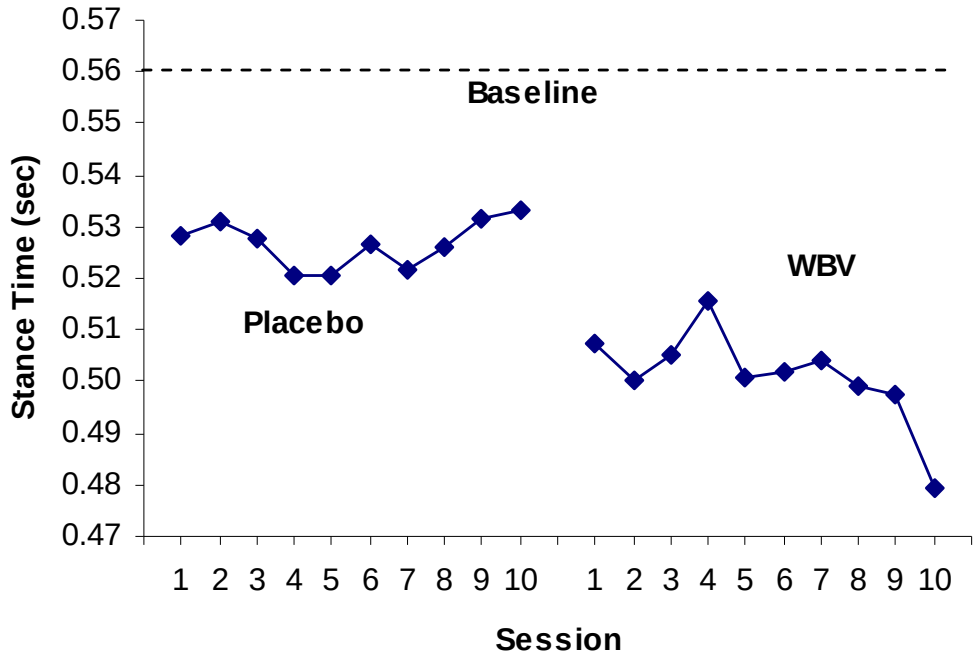


Figure 6.13: Stance time following sessions one, five and ten of WBV and placebo interventions for group WBV/Placebo

Note:
- Significant intervention contrast effects (p=0.001)
  Effect size = 0.64
- Significant intervention*session contrast effect for session five vs. ten (p=0.001)
  Effect size = 0.67

Figure 6.14 illustrates little change for stance time over the ten week placebo period. For WBV, although stance time was 0.03 seconds lower for session ten compared to session one it must be noted that a 0.02 second decrease occurred from sessions nine to ten. Analysis of individual sessions found that the decrease between sessions nine and ten was a common trend for most participants.



	Placebo		Whole Body Vibration	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.56	0.05		
1	0.53	0.06	0.51	0.07
2	0.53	0.06	0.50	0.06
3	0.53	0.05	0.51	0.05
4	0.52	0.05	0.52	0.05
5	0.52	0.05	0.50	0.05
6	0.53	0.05	0.50	0.05
7	0.52	0.06	0.50	0.06
8	0.53	0.06	0.50	0.06
9	0.53	0.06	0.50	0.06
10	0.53	0.06	0.48	0.05

Figure 6.14: Time-series graph and table for group mean stance time following placebo and WBV interventions (n = 12)

6.3.4.8 Stance Time for Group WBV/Placebo (WBV Weeks 1-6 and Placebo Weeks 7-12)

The 2x3 repeated measures ANOVA found placebo stance time to be significantly lower than WBV stance time (F(1,9)=5.44,p=0.045) and an effect size of 0.55 (Table 6.12). A main effect for session was also recorded (F(2,18)=4.47,p=0.027) with an effect size of 0.69. An interaction effect was not found for intervention*session.

Table 6.12: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean stance time (seconds) for sessions one, five and ten

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	5.44 (1,9)	0.045	0.55
Session	4.47 (2,18)	0.027	0.69
Intervention*session	2.06 (2,18)	0.157	0.37

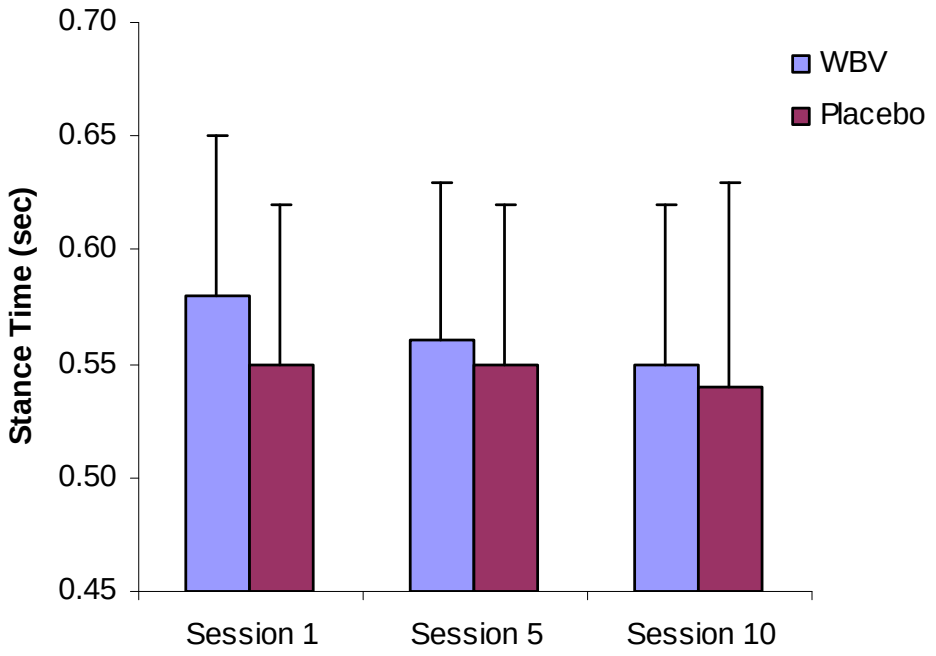
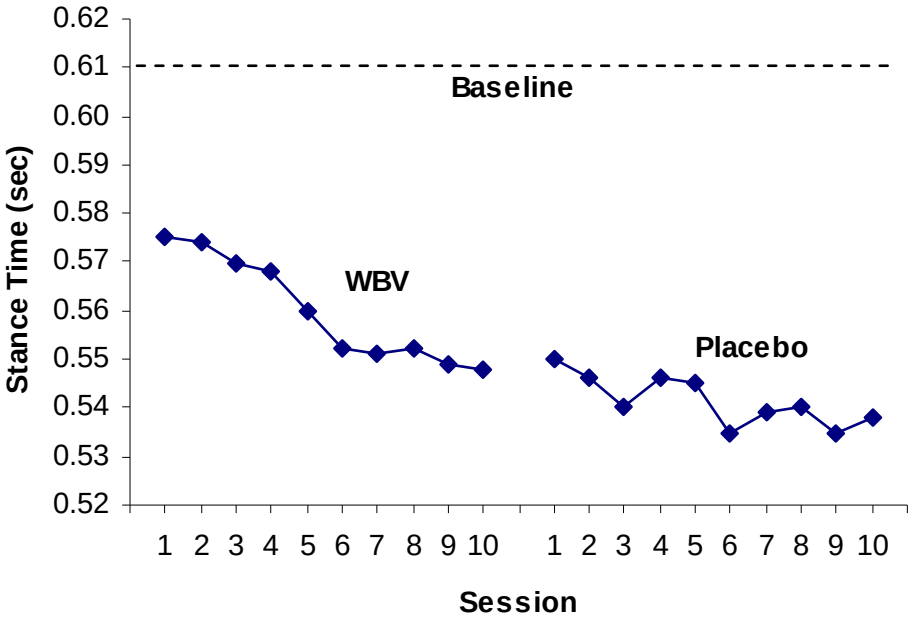


Figure 6.15: Stance time following sessions one, five and ten of WBV and placebo interventions

Note:
- Significant intervention contrast effects (p=0.045)
  Effect size = 0.38
- Significant session contrast effects for session 1 vs. later (p=0.048)
  Effect size = 0.37

Contrast analysis identified significant session differences (p=0.048) for session one vs. later and a low effect size of 0.38 (Figure 6.15). Stance time displayed a linear decrease over the ten WBV sessions (Figure 6.16). For WBV one it was 0.58 ± 0.07 seconds, 0.56 ± 0.07 seconds for WBV five and 0.55 ± 0.07 seconds for WBV ten. This demonstrates a 0.03 second decrease on stance time over this time period. Conversely, stance time displayed a relatively plateaued response during the placebo (Figure 6.16), decreasing by 0.01 seconds over the ten sessions. Placebo one was 0.55 ± 0.07 seconds, placebo five was 0.55 ± 0.07 seconds and placebo ten was 0.54 ± 0.09

seconds. Therefore, although the placebo stance time was significantly shorter for the placebo group, the largest change was for the WBV group.



	Whole Body Vibration		Placebo	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.61	0.07		
1	0.58	0.07	0.55	0.07
2	0.57	0.07	0.55	0.06
3	0.57	0.07	0.54	0.06
4	0.57	0.07	0.55	0.07
5	0.56	0.07	0.55	0.07
6	0.55	0.07	0.54	0.07
7	0.55	0.07	0.54	0.07
8	0.55	0.07	0.54	0.07
9	0.55	0.07	0.54	0.08
10	0.55	0.07	0.54	0.09

Figure 6.16: Time-series graph and table for group mean stance time following WBV and placebo interventions (n = 10)

6.3.4.9 Single Support Time for Group Placebo/WBV (Placebo Weeks 1-6 and WBV Weeks 7-12)

Statistical analysis failed to identify an intervention effect on single support time (Table 6.13). A main effect was found for session ($F(2,22)=8.84, p=0.002$) and an interaction effect was also calculated ($F(2,22)=4.63, p=0.021$).

Table 6.13: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean stance time (seconds) for sessions one, five and ten

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	0.51 (1,11)	0.462	0.05
Session	8.84 (2,22)	0.002	0.45
Intervention*session	4.63 (2,22)	0.021	0.30

For session, contrast analysis indicated that a significant difference existed for session five vs. session ten only ($p=0.002$), where stance time was lower for session ten compared to session five (Figure 6.17). For intervention*session, contrast analysis calculated a significant difference for session one vs. later ($p=0.022$). As session one and five are relatively similar, it appears that this interaction effect was produced at session ten where WBV single support time was lower than the placebo values, and the combined WBV and placebo values at this session were lower than for session one and five.

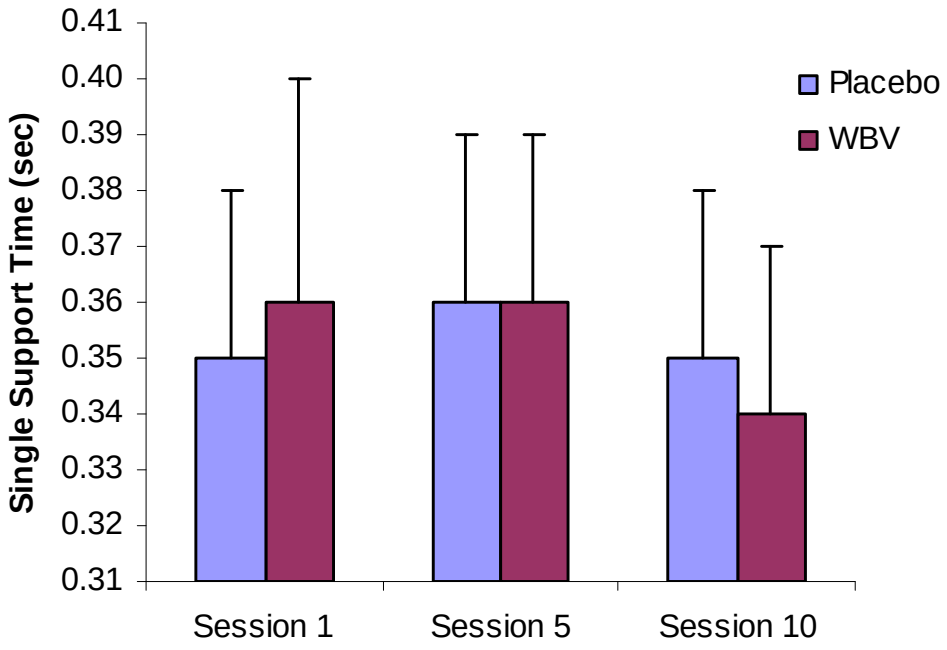
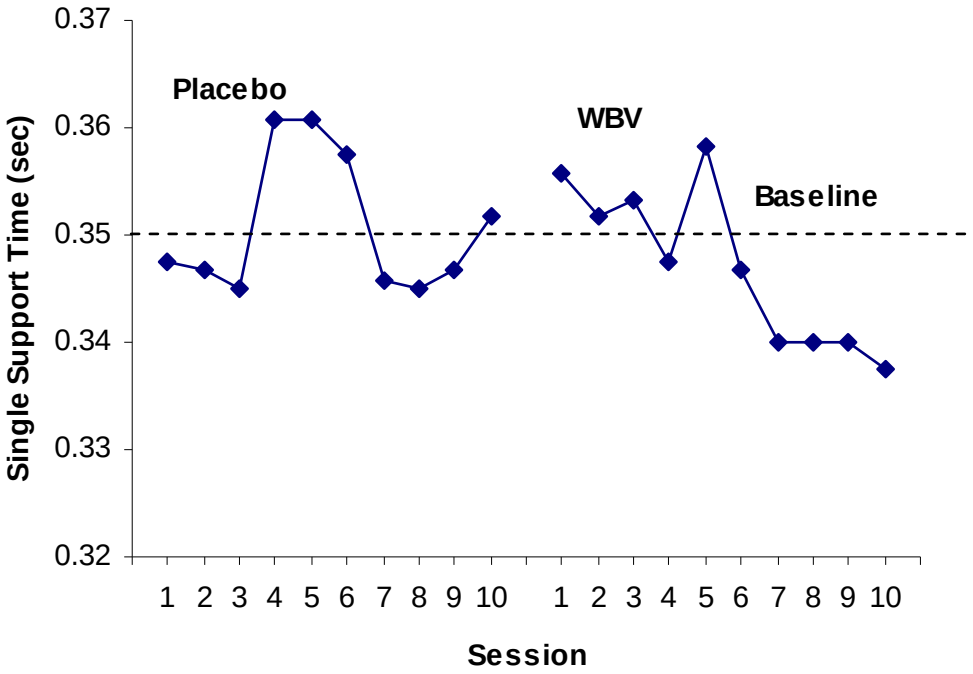


Figure 6.17: Single support time following sessions one, five and ten of WBV and placebo interventions for group placebo/WBV

Note:
- Significant session contrast effects for session 5 vs. session 10 (p=0.002) Effect size = 0.60
- Significant intervention*session contrast effect for session one vs. later (p=0.022) Effect size = 0.39

Visual inspection of the single support time time-series identifies an undulating pattern for the placebo period. Single support time exhibited WBV/Placebo horizontal periods separated by a horizontal peak at sessions four, five and six (Figure 3.9.1.2). WBV single support time presents with a different pattern over the ten sessions. For the first five sessions, single support time undulates within the vicinity of the baseline value, whereupon it begins to decrease linearly over the five subsequent sessions.



	Placebo		Whole Body Vibration	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.35	0.03		
1	0.35	0.03	0.36	0.04
2	0.35	0.03	0.35	0.03
3	0.35	0.03	0.35	0.03
4	0.36	0.03	0.35	0.03
5	0.36	0.03	0.36	0.03
6	0.36	0.03	0.35	0.02
7	0.35	0.03	0.34	0.03
8	0.35	0.03	0.34	0.03
9	0.35	0.03	0.34	0.03
10	0.35	0.03	0.34	0.03

Figure 6.18: Time-series graph and table for group mean single support time following placebo and WBV interventions (n = 12)

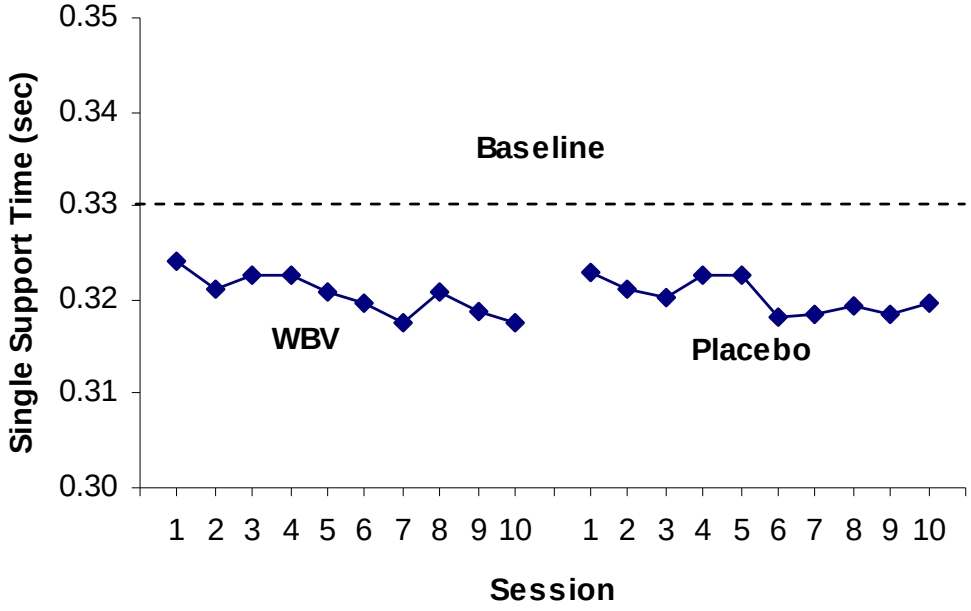
6.3.4.10 Single Support Time for Group WBV/Placebo (WBV Weeks 1-6 and Placebo Weeks 7-12)

Single support time for this group failed to demonstrate significant differences for intervention, session and intervention*session (Table 6.14).

Table 6.14: Two-way repeated measures ANOVA for the effect of WBV and placebo interventions on mean single support time (seconds) for sessions one, five and ten.

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	0.39 (1,9)	0.551	0.04
Session	1.31 (1.22,10.99)	0.293	0.13
Intervention*session	1.12 (2,18)	0.348	0.11

The time-series (Figure 6.19) displays single support time to remain relatively unchanged from sessions one to ten for both the WBV and placebo interventions.



	Whole Body Vibration		Placebo	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.33			
1	0.32	0.02	0.32	0.03
2	0.32	0.02	0.32	0.03
3	0.32	0.02	0.32	0.02
4	0.32	0.02	0.32	0.02
5	0.32	0.02	0.32	0.02
6	0.32	0.02	0.32	0.03
7	0.32	0.02	0.32	0.02
8	0.32	0.02	0.32	0.03
9	0.32	0.02	0.32	0.02
10	0.32	0.03	0.32	0.03

Figure 6.19: Time-series graph for group mean single support time following placebo and WBV interventions (n = 10)

6.3.4.11 Double Support Time for Group Placebo/WBV (Placebo Weeks 1-6 and WBV Weeks 7-12)

Statistical analysis (Table 6.15) demonstrated double support time to be significantly lower during the WBV period (F(1,11)=30.41,p=0.001). Compared to all other the gait parameters examined, this analysis indicated WBV to have the largest effect on double support time, where partial eta squared was calculated to be 0.73. A main effect (Table 6.15) was also found for session (F(2,22)=3.83,p=0.037) and an interaction effect was found for intervention*session (F(2,22)=5.18,p=0.014).

Table 6.15: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean double support time (seconds) for sessions one, five and ten

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	30.41 (1,11)	0.001	0.73
Session	3.83 (2,22)	0.037	0.26
Intervention*session	5.18 (2,22)	0.014	0.32

For session contrast effects (Figure 6.20), significance was found for session one vs. the mean of sessions five and ten (p=0.027). Visual inspection to determine which sessions produced this significant effect is somewhat unclear, as demonstrated by figure 6.20. The combined session means (i.e. the combined mean for WBV and placebo for each session) for session one, five

and ten were 0.17 seconds, 0.16 seconds and 16.5 seconds, respectively. The variability was constant for each session. Therefore, sessions five and ten appear to have positively impacted on this difference.

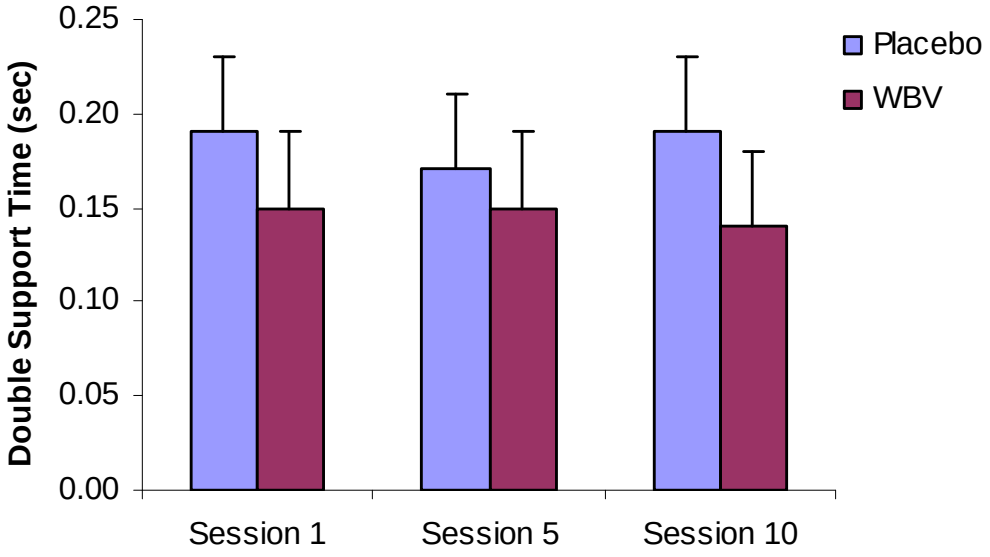
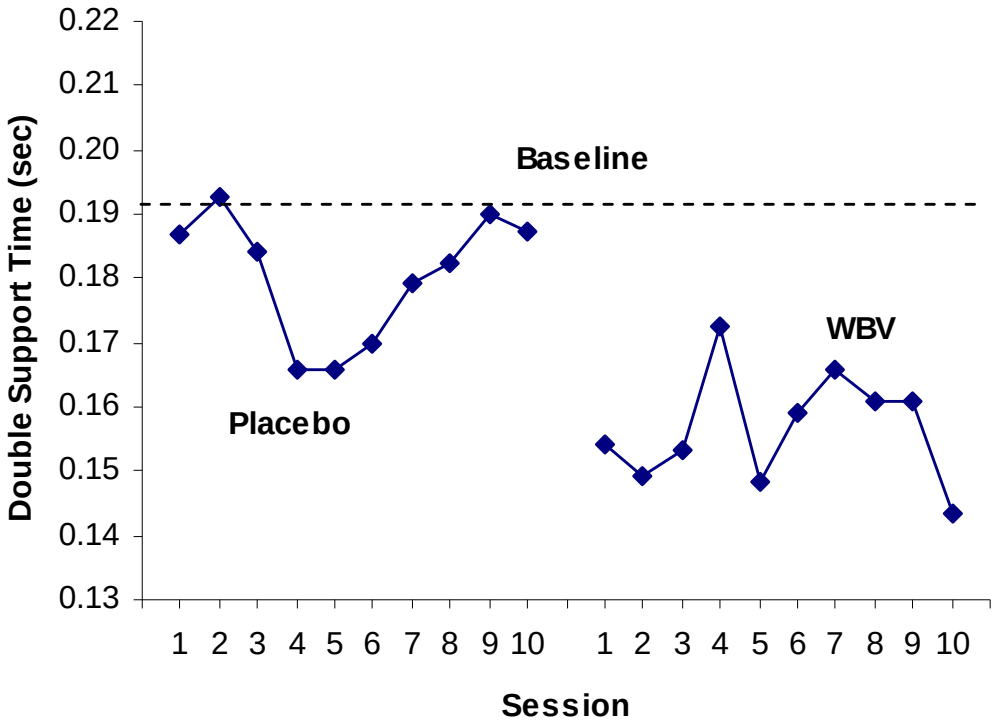


Figure 6.20: Double support time following sessions one, five and ten of WBV and placebo interventions for group placebo/WBV

Note:
- Significant intervention contrast effects (p=0.001)
  Effect size = 0.73
- Significant session contrast effects for session one vs. later (p=0.027)
  Effect size = 0.37
- Significant intervention*session contrast effect for session five vs. ten (p=0.004)
  Effect size = 0.55

Figure 6.20 show's double support time to have undulating patterns that show no distinct trends for both intervention periods. WBV was lower at most time points compared to the placebo. Double support time ranged between 0.17 to 0.19 seconds for the placebo and 0.14 to 0.17 seconds for WBV.



	Placebo		Whole Body Vibration	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.21	0.04		
1	0.19	0.04	0.15	0.04
2	0.19	0.04	0.15	0.04
3	0.18	0.04	0.15	0.05
4	0.17	0.04	0.17	0.04
5	0.17	0.04	0.15	0.04
6	0.17	0.03	0.16	0.04
7	0.18	0.05	0.17	0.05
8	0.18	0.04	0.16	0.05
9	0.19	0.04	0.16	0.05
10	0.19	0.04	0.14	0.04

Figure 6.21: Time-series graph and table for group mean double support time following placebo and WBV interventions (n = 12)

6.3.4.12 Double Support Time for Group WBV/Placebo (WBV Weeks 1-6 and Placebo Weeks 7-12)

The placebo group had significantly lower double support times (F(1,19)=17.71,p=0.002) and an effect size of 0.66 (Table 6.16). A main effect was also calculated for session (F(2,22)=5.22,p=0.016). Analysis of an intervention*session interaction effect identified no significant differences.

Table 6.16: Two-way repeated measures ANOVA for the effect of placebo and WBV interventions on mean double support time (seconds) for sessions one, five and ten

Source	F (d.f)	Sig.	Partial Eta Squared
Intervention	17.71 (1,9)	0.002	0.66
Session	5.22 (2,18)	0.016	0.37
Intervention*session	1.39 (2,18)	0.276	0.13

Figure 6.22 demonstrates the double support time sessions analysed by the two-way repeated measures ANOVA. It can be seen that double support time was higher at all time points for the WBV period. For session one, WBV double support time was 0.25 seconds, for session five it was 0.24 seconds and 0.24 seconds for session ten. For the placebo period, double support time was 0.23 seconds, 0.22 seconds and 0.22 seconds for placebo sessions one, five and ten respectively. Contrast analysis identified session one to be significantly different to the combined means of session five and ten (p=0.027).

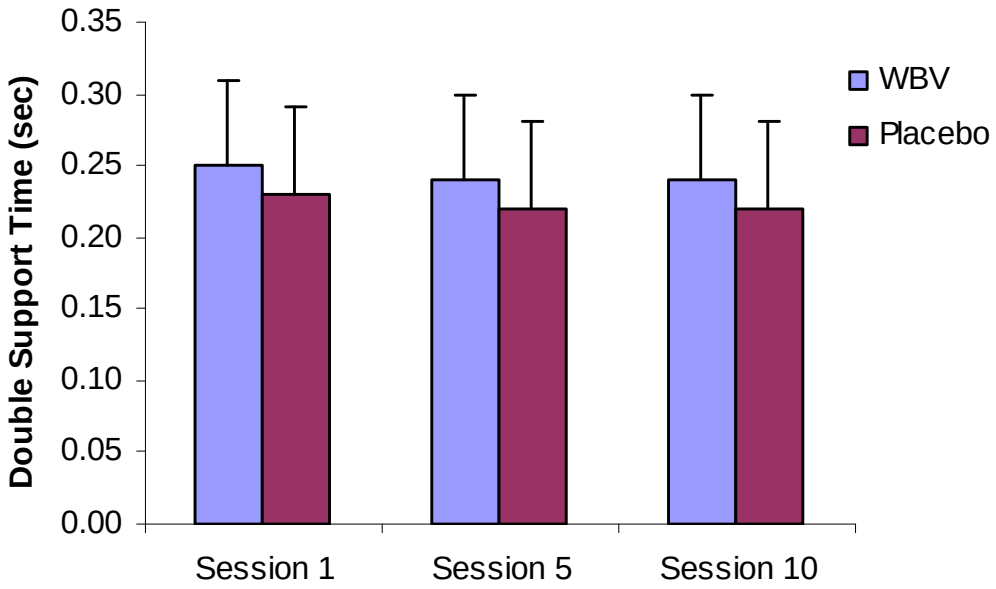
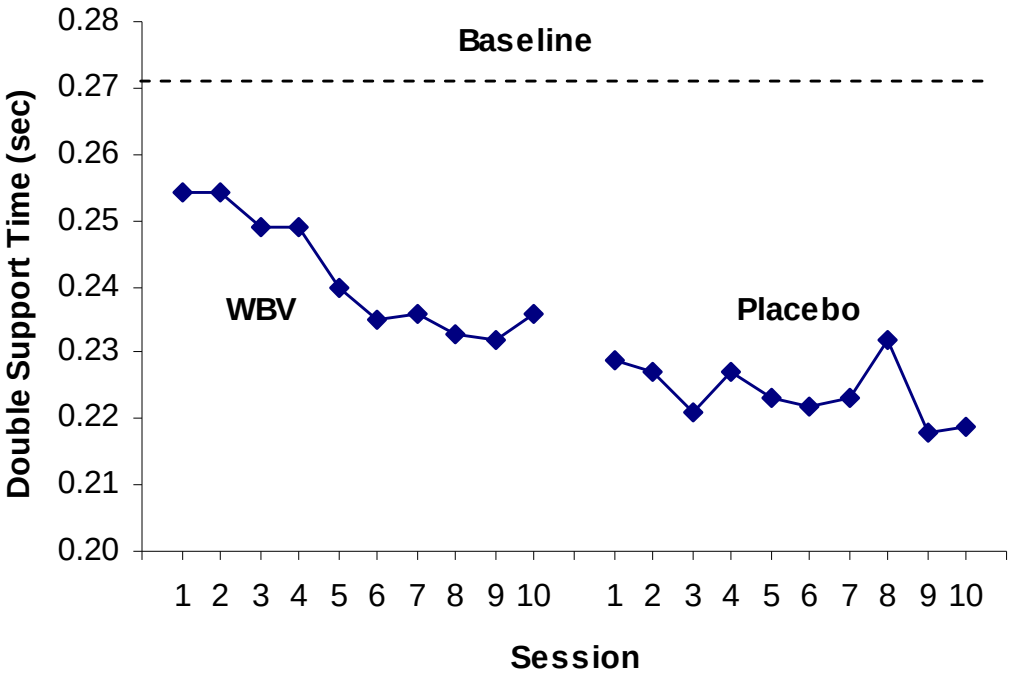


Figure 6.22: Double support time following sessions one, five and ten of WBV and placebo interventions for group WBV/Placebo

Note:
- Significant intervention contrast effects (p=0.002)
  Effect size = 0.63
- Significant session contrast effects for session one vs. session later (p=0.027)
  Effect size = 0.44

Visual time-series analysis of double support time illustrates this parameter to have a linear trend that descended with time for both intervention types (Figure 6.23). WBV has a slightly larger decrease with time of 0.02 seconds compared to the placebo which decreases by 0.01 seconds. The 0.02 second decrease was not distinguished in the two-way repeated measures ANOVA as there was a slight increase for the final WBV intervention session.



	Whole Body Vibration		Placebo	
Session	**Mean**	**SD**	**Mean**	**SD**
Baseline	0.27	0.06		
1	0.25	0.06	0.23	0.06
2	0.25	0.06	0.23	0.05
3	0.25	0.06	0.22	0.05
4	0.25	0.06	0.23	0.05
5	0.24	0.06	0.22	0.06
6	0.24	0.06	0.22	0.05
7	0.24	0.05	0.22	0.05
8	0.23	0.06	0.23	0.06
9	0.23	0.06	0.22	0.06
10	0.24	0.06	0.22	0.06

Figure 6.23: Time-series graph and table for group mean double support time following placebo and WBV interventions (n = 12)

6.3.4.13 Activity-Specific and Balance Confidence (ABC) Scale for Group Placebo/WBV

For the placebo/WBV group the ABC scale was incorrectly completed by six of the 12 participants. Only those questionnaires that were correctly completed were included in the statistical analysis. At baseline, post-placebo and post-WBV the ABC scores were 78.3±21.1%, 81.4±16.0% and 76.7±17.4% respectively (Figure 6.26). Statistical analysis failed to demonstrate a significant difference for the ABC scale at any stage of the intervention.

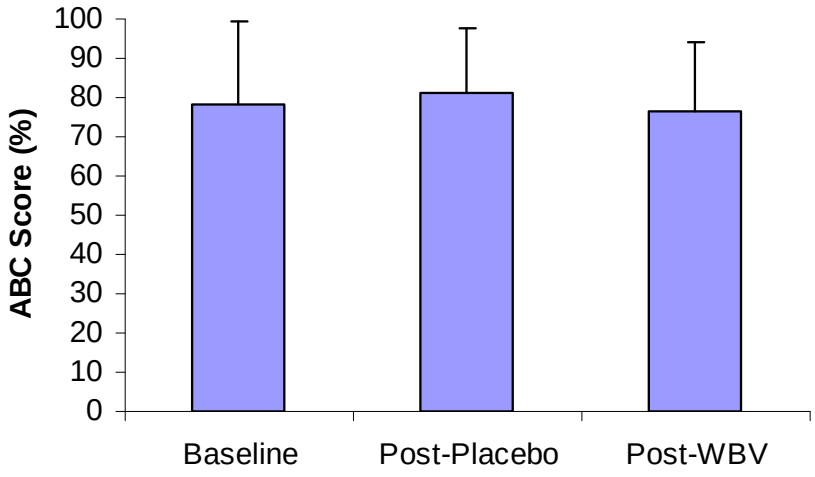


Figure 6.24: Percentage ABC scores for: (1) baseline; (2) at the completion of six weeks of the placebo intervention (weeks one to six); and (3) at the completion of six weeks of the WBV intervention (weeks seven to twelve).

6.3.4.14 Activity-Specific and Balance Confidence (ABC) Scale for Group WBV/Placebo

Only six of the 10 participants correctly completed the ABC scale for the WBV/placebo group. Only those questionnaires that were correctly completed were included in the statistical analysis. At baseline, post-WBV and post-placebo, the ABC scores were 74.7±9.3%, 80.7±15.3% and 75.9±15.3% respectively (Figure 6.28). Statistical analyses demonstrated that there was no significant difference for the ABC scale at any stage of the intervention.

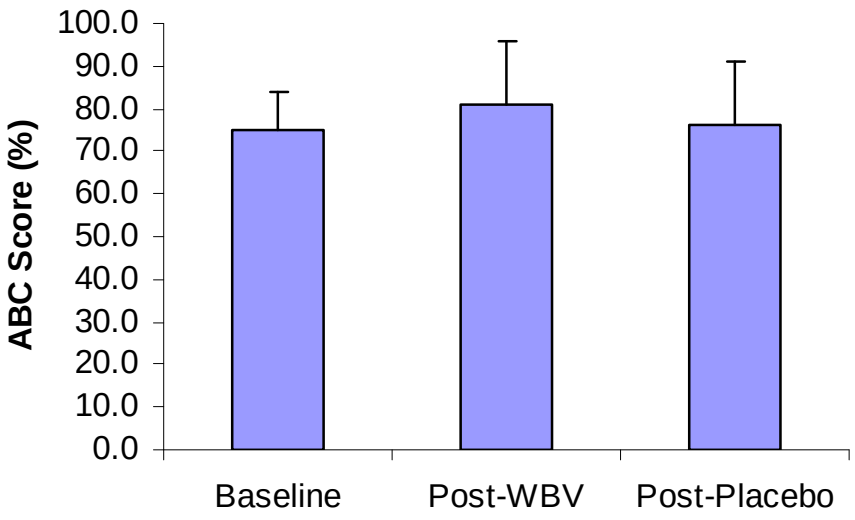


Figure 6.25: Percentage ABC scores for: (1) baseline; (2) at the completion of six weeks of the WBV intervention (weeks one to six); and (3) at the completion of six weeks of the placebo intervention (weeks seven to twelve).

6.4 Major Study Discussion

The current study performed cross-over design, time-series analyses to examine the response of elderly female gait to WBV and placebo interventions. Participants were placed into one of two groups for this study. Group Placebo/WBV was exposed to a placebo intervention for six weeks followed by a second 6 weeks where they were exposed to WBV. Group WBV/Placebo received the same intervention but in the reverse order whereby they were exposed to the WBV for the first 6 weeks and the placebo during the following six weeks.

Compliance should be considered whenever investigating the utility of an intervention. Previous WBV work with institutionalised elderly has reported compliance rates ranging between 72.7 and 96 per cent (Bautmans et al., 2005; Bruyere et al., 2005). In community dwelling older adults, the only study to have reported compliance indicated that participants completed 77 per cent of all sessions (Russo et al., 2003). For this study, compliance was poor. Although no participants dropped out of the study, five out of 27 participants did not complete at least ten sessions in one of the interventions and were therefore excluded from the analyses. Of the 22 participants who completed the study, 82% of the sessions were completed in the first six weeks and 85% in the final six weeks. Several explanations were given for non-attendance including viruses, holidays and sporting events (lawn bowls). Another explanation may be the rigid structure of the program where intervention sessions were held on only three days of the week, providing no flexibility for the participants. Attendance was probably also impacted upon by travel time

as a number of participants had to travel across several suburbs to attend the sessions. The data showed that only ten WBV sessions were necessary to demonstrate improvements in stride velocity. This valuable information indicates that despite the 'real-world' barriers to regular attendance, the participants still experienced significant benefits from sessions that they did attend.

Figures 6.4 and 6.5 demonstrate that when participants were exposed to WBV, their stride velocity increased in a linear fashion over ten sessions. Following the first intervention session, increases were evident for both the placebo (0.7 $cm.s^{-1}$) and WBV interventions (0.9 $cm.s^{-1}$). Since both groups reported similar increases in stride velocity, these improvements were considered to be placebo responses and were therefore excluded from statistical analyses.

Older adults performing 24 weeks of combined strength and explosive resistance training have demonstrated an 11 per cent increase in stride velocity (Häkkinen, Alen, Kallinen, Newton & Kramer, 2000). In the current study, statistical analyses only identified a significant intervention effect for group placebo/WBV which increased by 0.12 $m.s^{-1}$ or 7.2 per cent. Despite group WBV/placebo displaying a similar linear pattern and a 0.8 $m.s^{-1}$ increase (5 per cent) an intervention effect was not found for this group. It is probable that the effect was masked by the undulating pattern of data for group WBV/placebo during the placebo period. This group's placebo stride velocity for sessions one and five dropped below the stride velocity recorded in the

final WBV session. The large decrease in placebo stride velocity of 0.5 $m.s^{-1}$ for session one of the placebo period is somewhat difficult to explain. Several participants demonstrated disappointment in the reduced intensity during this period indicating that they found the higher intensity more pleasant. It may be that this disappointment led to a reduced motivation to walk fast during this first placebo session. Participants returned towards the stride velocity recorded at the final WBV over subsequent sessions and were able to maintain this velocity for at least ten sessions equating to approximately three weeks. This demonstrates that the participants were able to maintain the benefits received from the WBV for at least this period. Previous research that identified improvements in walking velocity following resistance training has demonstrated elderly men and women to maintain these changes for 48 weeks after a final resistance training session (Häkkinen et al., 2000). Prior work has also indicated that a 12 week fast walking program does not improve fast walking (Paillard et al., 2004). This is consistent with the current study which found no changes in stride velocity during the placebo periods.

Interestingly, no participants, for either group, indicated that they felt that the low intensity may have been a placebo, although they did say that they preferred the higher intensities. Only one other study has used a placebo WBV intervention with older adults (Bautmans et al., 2005). In contrast to this work, participants stood on a WBV plate that was switched off and had a tape recording of the sound of the WBV motor playing below the device. As a placebo, we believe that this method compared to that used by other research (Delecluse et al., 2003), is more effective as participants could hear the motor

running and could also feel a mild tingling in their feet from the very low plate accelerations.

To increase gait speed an individual must increase their stride length and/or decrease their stride time. Based on the pilot study, it was hypothesised that if stride velocity increased, stride time would be the component that produced this change. Although the placebo/WBV group showed intervention effects for stride length, this was not observed in the WBV/placebo group. During WBV, both groups' data undulated within a range of 3 $cm.s^{-1}$. A 3 cm increase in stride length would contribute to a 3 $cm.s^{-1}$ increase in stride velocity. As discussed previously, stride velocity increased by 8 $cm.s^{-1}$ and 12 $cm.s^{-1}$ for both groups. Assuming stride length increased significantly, based upon a 3 cm increase, stride length would have contributed to approximately 25-30 per cent of the changes. Evidently, the reduction in stride time was the major contributor to the changes in stride velocity which supports the stated hypothesis (p. 137).

Figures 6.26 and 6.27 illustrate the relatively inverse relationship between stride velocity and stride time for both groups, where as stride time decreased, stride velocity increased. This relationship replicates data previously recorded in the pilot study. For group placebo/WBV, stride time decreased by 0.05 seconds from sessions one to ten, contributing 0.10 seconds to the 0.12 $m.s^{-1}$ increase in stride velocity.

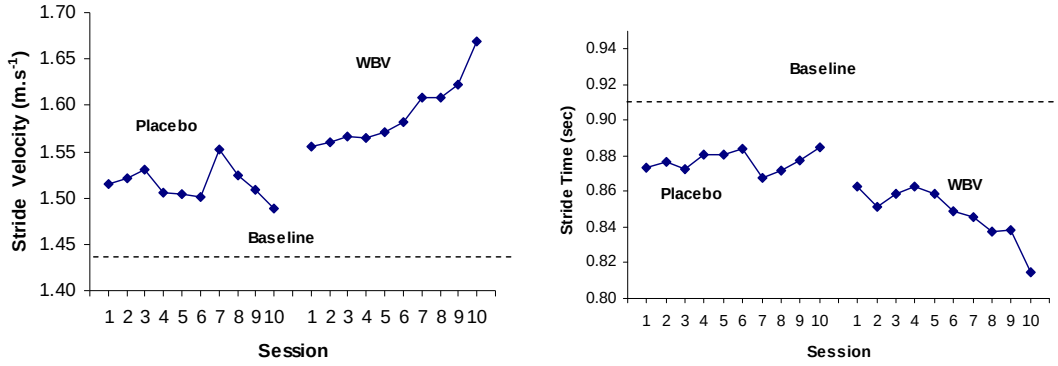


Figure 6.26: The inverse relationship between stride velocity and stride time for group placebo/WBV

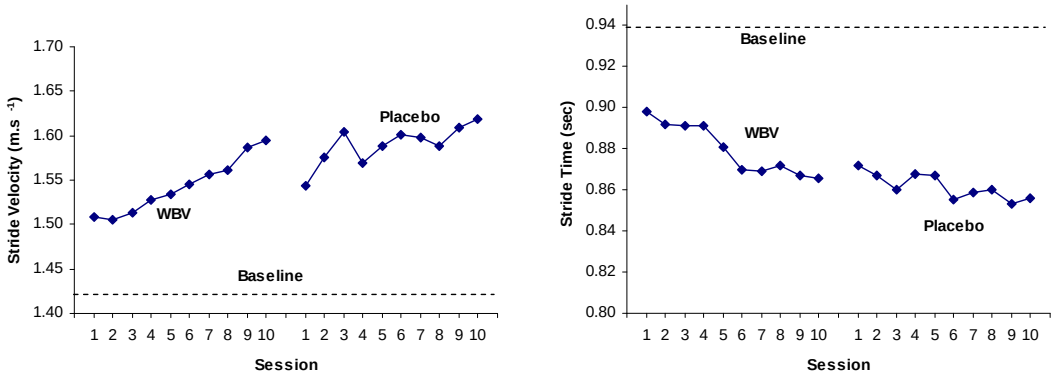


Figure 6.27: The inverse relationship between stride velocity and stride time for group WBV/placebo

Stride time is comprised of two major components, that is, stance time and swing time. The swing time was not directly analysed in this study as it is proportional to single support time, which is to be discussed further at a later stage. Further analyses identified stance duration to be related to the changes in stride time and therefore stride velocity.

Stance duration increases with age, most probably serving to stabilise the gait pattern (Lord et al., 1996; McGibbon & Krebs, 1999). An intervention effect

was found for stance time for both groups. For group Placebo/WBV, stance time was lower during the WBV periods for most time points, while for group WBV/Placebo, stance time was lower during the placebo period. Despite both groups decreasing stance time by 0.03 seconds over their respective WBV periods, their stance time appeared to respond differently during the WBV period. Although group Placebo/WBV had a shorter stance time during the WBV period compared to the placebo period, it was relatively stable for the first nine sessions. At the completion of session ten there was a large decrease, where stance time was 0.02 seconds faster than session nine. In contrast, WBV decreased in a linear fashion over the ten sessions for group WBV/Placebo. At this point it is unclear why the groups responded differently. It is possible that the change was due to the order of stimulus where one group received WBV first and the other experienced the placebo. It must be noted, however, that group WBV/Placebo produced a similar stance time pattern to the pilot study which was conducted over nine sessions of WBV. There is therefore a precedent for such a pattern.

Stance time can be further divided into single support time and double support time. The pilot study indicated that single support time remained relatively unchanged, while a reduced double support time was associated with the increased stride velocity. For this study, group WBV/placebo replicated this pattern; however, for group Placebo/WBV the contribution of these two parameters is less clear. For this group, single support time undulated within 0.01 seconds for the placebo, and the WBV period displayed a 0.02 second decrease. The GAITRite is unlikely to have the capacity to measure such

small changes in single support time as it records at 80 Hz, or every 0.0125 seconds. It is plausible that the measurement errors that were produced during the single support phase impacted upon the double support phase as the latter parameter is the phasic temporal opposite.

Based upon the perceived benefits reported by participants during the pilot research, the activities-specific balance confidence scale was included in this study. Unfortunately only six out of twelve participants in group placebo/WBV and six out of ten participants in group WBV/placebo correctly completed the form. Participants were requested to indicate their level of self-confidence from the corresponding rating scale that was provided. It was explained to them that 0% means that they have no confidence and 100% means they feel completely confident with that particular task. Table 6.17 illustrates how participants incorrectly and correctly answered the questions. It was unclear why the participants answered the questions incorrectly by circling the text rather than the percentage. It was evident by their behaviours, however, that at the time of completing the form the participants were in a hurry and found it "annoying".

1. How confident are you that you can maintain your balance and remain steady when you walk around the house?

0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
Not at all confident					Somewhat confident					Absolutely confident

Table 6.17: Example of participants incorrectly (red circle) and correctly (blue circle) answered questions on the activities-based confidence scale.

The study demonstrated that older adults can receive benefits from WBV with relatively lower accelerations (2.0 $m.s^{-2}$ to 24.7 $m.s^{-2}$) than have been reported by previous studies (5.9 $m.s^{-2}$ to 241.9 $m.s^{-2}$) (Bautmans et al., 2005; Bruyere et al., 2005). Currently there are no published scientifically based guidelines for the most appropriate frequencies and amplitudes for different groups and desired outcomes. Accelerations produced by WBV are the product of frequency and amplitude. An increase in either of these variables will increase the resulting acceleration. The frequencies and amplitudes selected by studies have varied considerably with little explanation as to why they have chosen these parameters. For athletes and healthy young adults, frequencies have typically ranged between 20 to 35 Hz (Bosco et al., 1998; Bosco et al., 2000; Rittweger et al., 2001). The most significant impact on the accelerative forces is the amplitude, which has varied significantly between studies, reportedly ranging between 2.5 to 8 mm (Bosco et al., 2000; Cochrane et al., 2004; Cronin et al., 2004; de Ruiter, van Raak et al., 2003; Rittweger et al., 2001; Roelants et al., 2004). For older adults, frequencies have ranged between 10 to 40 Hz and amplitudes 1.5 to 5 mm (Bryuere et al., 2005; Roelants et al., 2004). A single session WBV study performed at a high amplitude (6 mm) reported young adult participants to complain of pain in the jaw, neck and lower extremity which required physiotherapy treatment (Cronin et al., 2004). For older adults participants have dropped out, due to knee pain, in research that used amplitudes of 2.5 and 5 mm (Roelants et al., 2004).

At the extreme end of these reported parameters, plate accelerations at the point of contact with the feet of the individual are as high as 177.7 $m.s^{-2}$

(18.1 g) at a frequency of 30Hz and amplitude of 5mm. The maximal accelerations participants were exposed to in the current study was a cautious 35.5 $m.s^{-2}$ (3.6 g) due to the belief that there may be an inherent injury risk such as an osteoporotic related femoral neck fracture associated with the large accelerations reported in previous elderly WBV research. Concerns for WBV exceeding 1 g have been previously indicated within the literature (Rubin et al., 2004). This is the first study to incorporate such low accelerations and it has demonstrated that the gait of elderly participants will benefit from these loads without injury or major side effects. In this research, one participant reported lower leg itching and erythema, a response that has been reported for both the young (Cronin et al., 2004) and old (Roelants et al., 2004; Russo et al., 2003). Similar to that reported by other studies (Roelants et al., 2004; Russo et al., 2003), these side effects were temporary and were not experienced again following the first two sessions. The most significant event to occur, which has not been reported in WBV previous literature, was an incontinence episode by one participant, post-WBV. This occurred on one occasion only and occurred during the first week of WBV. This individual, who later indicated that urinary incontinence had been a major problem since the birth of her children, continued with the study and reported that by weeks 3-4, the urinary incontinence she had previously experienced with her daily activities of living was no longer a concern. Although previous research has demonstrated WBV to improve urinary incontinence (von der Heide, Emons, Hilgers & Viereck, 2003), there have been no reports of acute episodes such as this.

The study incorporated a similar exercise protocol to previous elderly research where participants stood in a stationary position on the platform while exposed to the vibration (Bruyere et al., 2005; Russo et al., 2003). Other work has required older adults to perform static (Bautmans et al., 2005) and dynamic (Roelants et al., 2004) exercises while simultaneously experiencing the WBV. Although each of these studies has demonstrated improvements in their measured outcome, it is unclear whether one method is superior to the other for this age group.

Overall, this study has demonstrated benefits to fast walking gait for elderly females at low amplitudes of WBV. The underlying mechanisms that are responsible for these improvements are unclear. Scientific inquiry does not currently have a comprehensive explanation for the neuromuscular adaptations that occur following WBV. Although physical activity studies have demonstrated neural system plasticity (Barbeau, Ladouceur, Mirbagheri & Kearney, 2002; Lagerquist, Zehr & Docherty, 2006), the exact source of these changes is difficult to isolate since alpha motor neurons receive input from several locations including the motor cortex, brain stem, spinal cord and afferent pathways (Bawa, 2002). One explanation, specific to the present study, is the possibility of an enhanced stretch reflex function. WBV, combined with simultaneous squats, has been demonstrated to have an acute stimulatory effect on the stretch reflex (Rittweger et al., 2003). Direct application of vibration to the tendon stimulates the primary and secondary muscle spindle endings (Burke et al., 1976a; Burke et al., 1976b) producing a reflexive muscular contraction termed the “Tonic Vibration Reflex”. Vibration’s

mechanical action creates short, rapid changes in the muscle-tendon complex, activating muscle spindles and thereby stimulating the stretch reflex (Cardinale & Bosco, 2003).

Although it is understood that mechanical vibration can stimulate muscular contraction by monosynaptic and polysynaptic pathways, it is difficult to make a generalisation pointing to which element of the stretch reflex pathway is altered by vibration as there are several components involved. These include, but are not limited to, the muscle spindles, Ia afferent fibres, motor end plates, inter-neurons and alpha motor neurons (Mynark & Koceja, 2001). The degeneration of one or more of these components will negatively impact on the stretch reflex, and are therefore likely to impact upon activities that are reliant upon this mechanism.

Stretch reflexes contribute to human locomotion and are believed to play stabilising and propulsive roles during the stance phase (Chalmers & Knutzen, 2000; Simonsen & Dyhre-Poulsen, 1999; Sinkjær, Andersen & Larsen, 1996; Yakovenko, Gritsenko & Prochazka, 2004; Zehr & Stein, 1999). In humans and decerebrate cats, afferent pathways have been demonstrated to stimulate activity of the ankle extensor motor neurons during the stance phase of walking (Chalmers & Knutzen, 2000; De Serres, Bennett & Stein, 2002; Hiebert & Pearson, 1999; Sinkjær et al., 1996). In humans, this reflexive activity is smaller during the mid- to late-stance for the elderly compared to the young, indicating that the former recruit a smaller number of muscle fibres

from their motor pool and therefore potentially have reduced stability and propulsion (Chalmers & Knutzen, 2000).

The small changes in stride length during the current study suggest it is unlikely that WBV enhanced the propulsive abilities of the triceps surae. It is therefore more likely that improvements were due to improved stability. To optimise balance, older adults spend more time in the stance and double support phases of the gait cycle (Judge et al., 1996; Murray et al., 1969; Winter et al., 1990). Both of these gait parameters decreased during WBV for group WBV/placebo, while group placebo/WBV displayed a shortened stance time during the WBV period. The first double support phase occurs at 0-10 per cent of the step cycle and is the period of time where energy absorption occurs at the knee extensors and ankle flexors (Judge et al., 1996; Winter et al., 1990). A reduction in musculotendinous stiffness at both or either of these joints may reduce the capacity of older adults to absorb energy during the loading response and may threaten balance (Chalmers & Knutzen, 2000; Judge et al., 1996; McGibbon & Krebs, 1999). Since the stretch reflex contributes to enhanced musculotendinous stiffness (Chalmers & Knutzen, 2000), it is possible that the stimulated stretch reflex activity that occurs during WBV in both the knee extensors and plantar flexors had a positive impact upon the stretch reflex and musculotendinous stiffness, and therefore stability.

Although it has been postulated in several studies (Bosco et al., 1998) and reviews (Cardinale & Bosco, 2003) that vibration increases muscle stiffness, to date there has only been one study to have investigated the response of

musculotendinous stiffness to WBV (Cronin et al., 2004). In young healthy adults, an observable but not statistically significant acute increase in muscle stiffness has been observed after a single WBV session (Cronin et al., 2004). Failing to find a significant difference was probably due to an inadequate sample size. There have been no studies that have examined the musculotendinous stiffness of the elderly following WBV.

Since somatosensory reaction time tasks have been demonstrated to activate the motor cortex (Naito, Kinomura, Geyer, Kawashima, Roland, & Zilles, 2000) it has been suggested that the motor cortex should also be considered as a possible site that is stimulated by vibration and leads to performance enhancement (Cardinale & Bosco, 2003). The sensitivity of the corticospinal pathway to muscle tendon vibration (Steyvers et al., 2003) suggests that this is a possibility. The optimal frequency for muscle-tendon vibration to increase corticospinal excitability has been approximated to be 75 Hz for the flexor carpi radialis, while no observed changes have been found at 20 Hz (Steyvers et al., 2003). This would suggest that the frequency typically used in WBV research would not stimulate the corticospinal pathway. However, conclusions such as this should be made cautiously as only one muscle was stimulated by Steyvers et al., (2003). It is possible that WBV at the lower frequencies stimulates the corticospinal pathway via the stimulation of multiple muscle groups. Further research is required in this area before generalisations can be made.

Although cutaneous receptors are sensitive to vibration (Ribot-Ciscar, Vedel & Roll, 1989) it is unlikely that they have contributed to the enhanced stride velocity. The cutaneous reflex is relatively inactive over even surfaces during the stance phase (Zehr, Komiyama & Stein, 1997). This pathway is most active during the swing phase when perturbations occur to the swinging limb; or the stance limb during uneven terrain (Zehr, Stein & Komiyama, 1998; Zehr & Stein, 1999).

There is clearly an opportunity for future research to identify the mechanisms contributing to the enhanced neuromuscular function in older adults. Future research should examine:

(1) The stretch reflex mechanism of older adults before and after whole body vibration;

(2) The muscle stiffness of older adults before and after WBV; and

(3) The strength and power of the older adult triceps surae before and after WBV.

This research has contributed to the understanding of the human response to WBV in several key areas. The major contributions are: (1) the development of a unique WBV platform (ACUWBV) that allows the researcher to increase WBV amplitude and the participant to maintain a consistent stance width, thereby eliminating a possible confound; (2) evidence that low amplitude/low frequency WBV increases the walking velocity of elderly females; (3) insight into the gait parameters and mechanisms that contribute to the increased walking velocity of older adults; (4) evidence that the improvement in walking velocity exhibited by older adults following a six-week WBV may be maintained for at least six weeks after the intervention period; and (5) further supportive evidence for the low risk of major adverse effects from low frequency/low amplitude WBV in the older adult population. The following discussion summarises each of these key contributions.

(1) Development of a Unique WBV Platform

A unique WBV platform (ACUWBV) was developed for this research. The ACUWBV produced tilting oscillations similar to that of commercial devices such as the Galileo 2000 (NovotTec, Pforzheim, Germany). A limitation of current commercial WBV devices that produce vibration from tilting oscillations, is that to increase the vibration amplitude, the individual must change their standing posture by increasing their stance width on each side of a tilting axis. Different standing postures impact upon the transmissibility of WBV as it moves proximally from the feet (Harazin & Grzesik, 1998; Matsumoto & Griffin, 1998). The eccentric cam developed for the ACUWBV does not require the participant to change their stance width to increase

vibration amplitude, therefore allowing for the standardisation of this portion of the testing protocol. The ACUWBV eccentric cam was demonstrated to be a reliable method for adjusting platform amplitude. Although the intra-class coefficient calculated was considered low (ICC = 0.4), the typical error for each amplitude setting indicated small error ranges that had little impact upon the final plate acceleration outcome. For example, for an amplitude of 0.5 mm and a frequency of 20 Hz, the acceleration error range was 7.58-8.85 $m.s^{-2}$.

The ACUWBV eccentric cam also provides a placebo that is superior to that reported by previous research. A study with elderly nursing home residents used a tape recorded sound of the WBV motor running while participants stood on the device (Bautmans et al., 2005). The ACUWBV creates the placebo effect when the drive shaft is manually centred on the eccentric cam (amplitude = 0 mm), producing a mild vibration feeling at the base of the feet. The participant can also hear the motor running, further contributing to the placebo experience. There was no indication by any of the participants that they were aware of being exposed to a placebo suggesting that this is a viable method for future research.

(2) Low Amplitude/ Low Frequency WBV Increases Elderly Female Gait Velocity

Both the pilot and major studies demonstrated that low amplitude/low frequency WBV is effective for increasing the walking velocity of elderly females. Although the frequency range (10 to 30 Hz) was similar to previous research examining the impact of WBV on the functional mobility of older

adults, comparatively lower amplitudes (0.5 to 1.0 mm versus 1.5 to 5.0 mm) were used for the present research (Bautmans et al., 2005; Bruyere et al., 2005). The minimum and maximum accelerations that the participants were exposed to were therefore approximately 2.0 $m.s^{-2}$ ($0.2\,g$) and 24.7 $m.s^{-2}$ ($2.5\,g$). These accelerations are considerably lower than those reported by previous researchers which has ranged between 96.7 $m.s^{-2}$ ($9.9\,g$) to 241.9 $m.s^{-2}$ ($24.7\,g$) (Bautmans et al., 2005) and 5.9 $m.s^{-2}$ ($0.6\,g$) to 93.4 $m.s^{-2}$ ($9.5\,g$) (Bruyere et al., 2005). Concerns of potential harm using high accelerations have previously been indicated (Rubin et al., 2003; Rubin et al., 2004). Studies to have reported adverse effects showed healthy young adults to experience hip (Crewther et al., 2004) and knee pain (Roelents et al., 2004) and older adults to have experienced knee (Roelants et al., 2004; Russo et al., 2003), back (Roelants et al., 2004) and groin pain (Bautmans et al., 2005). At an acceleration of 160.2 $m.s^{-2}$ (26 Hz; 6 mm; $16.3\,g$), untrained participants reported pain in the lower limbs, neck and jaw up to 7-10 days after a single WBV session (Cronin et al., 2004). Although higher accelerations (241.9 $m.s^{-2}$; 35 Hz; 5 mm; $24.7\,g$) have been used with older adults (Bautmans et al., 2005), the potential of injury at such high intensities may be even greater for older adults, and especially for those with predisposing conditions such as osteoporosis. The low accelerations used by this research were shown to be safe and effective for improving the fast walking velocity of elderly females. Furthermore, the high accelerations applied by previous researchers appear to be unnecessary for enhancing the fast walking velocity of elderly females.

This is also the first research to measure gait changes after each intervention session. This unique format provided additional information about the effectiveness of each training intensity to improve the fast walking velocity of elderly females. Previous research has only performed pre- and post-intervention testing to examine changes in the mobility of elderly females. The time-series method demonstrated that the low intensities that were used in the first sessions of WBV had an immediate, although small, impact upon the fast walking capacity of elderly females. This provided further evidence for the effectiveness of low frequency/low amplitude WBV for elderly females.

(3) Gait Parameters and Mechanisms Contributing to Increases in Elderly Female Walking Velocity

Previous research has used clinical measures, such as the timed-up-and-go and the Tinetti test, to identify the functional mobility changes of older adults from WBV. This research is the first to objectively measure the changes in gait parameters. It was predicted that walking velocity would increase due to increases in stride length and decreases in stride time. This assumption was based on previous research in the elderly which had demonstrated improved leg power. It was assumed that improvements in leg power would contribute to the capacity to improve stride length and reduce the time for each stride. Stride length, however, demonstrated very little change over the six week period and stride time was the significant contributor to increased walking velocity. Further analyses were conducted on the gait parameters that were involved in the faster stride time. For the pilot study, stance time decreased while there was no change in swing time. Stance time is composed of double

support time and single support time. Single support time is considered the temporal equivalent of swing time as they occur in-phase to each other. The propulsive phase occurs during single support, therefore if the improvements in walking velocity occurred during the propulsive phase, expressed as an increase in stride length and/or a decrease in stride time, the changes would be evident during the single support period. For the pilot and the WBV/Placebo group, there were no changes in single support time; the key contributor to the improved stride time was a shorter double support time. Unfortunately these results were not replicated in the placebo/WBV group who did not show any clear pattern for either single support or double support time. It was suggested that this discrepancy may be due to measurement error of the GAITRite system used for placebo/WBV. This system was used for the pilot study and for group placebo/WBV. The GAITRite walkways have a measurement accuracy of ±0.0125 seconds and therefore a variable such as single support time, which showed undulating changes of 0.01 seconds, is more likely to be affected upon by measurement error compared to stride time which is a larger time component of the gait cycle.

Two studies to have examined the impact of WBV on the functional mobility of older adults have implied enhanced strength and power to be the likely mechanism to contribute to the improved performance (Bautmans et al., 2005; Bruyere et al., 2005). Only one study suggested the improvements in functional mobility were due to tonic vibration reflex stimulation (Bautmans et al., 2005). The triceps surae is the prime contributor to forward propulsion during human walking gait (Kepple et al., 1997) and is therefore the most likely

muscle group to contribute to the faster walking velocity following WBV. This research indicates that enhanced propulsion via the triceps surae is unlikely. If enhanced strength and power of this muscle group contributed to the improved walking velocity, it is reasonable to assume that single support time would decrease since this phase of the gait cycle propels the body forward. As highlighted above, the reduction in double support time, and not single support time appears to be the significant contributor to the faster walking velocity. This research therefore provides evidence for enhanced stability as an alternative explanation for the improvements in functional performance.

To counteract reduced lower extremity function and enhance stability, older adults spend a greater proportion of time with both feet in contact with the ground (Judge et al., 1996; Murray et al., 1969; Winter et al., 1990). The participants had a reduced reliance upon this strategy following the six week WBV intervention. This presented as decreases in stance and double support time, and may explain the enhanced fast walking speed. The ankle and knee muscle extensors absorb energy during initial ground contact (i.e. heel contact) (Judge et al., 1996; Winter et al., 1990). The capacity for this energy absorption diminishes with age, and is associated to some extent with deterioration in stretch reflex activity which subsequently produces a reduction in both motor unit recruitment and muscle stiffness (Chalmers & Knutzen, 2000). Since it is known that vibration stimulates the tonic vibration reflex, it is therefore proposed that enhanced stretch reflex activity of the ankle and knee extensors, and a consequent enhanced capacity to absorb energy, contributed to the reduction in stance and double support time.

(4) Maintenance of Fast Walking Velocity during Six-Week Detraining Period

For the main study, participants were assigned to one of two intervention groups: (1) WBV for first six weeks followed by a WBV placebo for six weeks; or (2) a WBV placebo for six weeks and WBV for the final six weeks. This is the first WBV research to have implemented such a design. The benefits from this cross-over design are that all participants are exposed to the intervention for a period of time. The ethical dilemma of using a placebo and thereby excluding participants from the potential benefits of the WBV intervention was therefore eliminated.

Following six weeks of exposure to WBV, participants were able to maintain the benefits from WBV. It is possible that participants maintained the ability to walk fast because they continued to perform fast walks at each placebo intervention session. An alternative method may have been to have participants not attend placebo sessions following the WBV period and then measure their walking velocity after six weeks. However, using the non-attendance method would not have provided valuable information regarding what was happening to walking velocity over the six weeks. This research has demonstrated that participants are able to maintain a fast walking velocity for six weeks after being exposed to WBV.

(5) Adverse Effects to WBV

There were no lasting adverse effects due to the WBV. The most significant adverse effect was a urinary incontinence episode. There have been no reports of acute episodes such as this in the literature. Participants of previous research may not have experienced such episodes or may not have told the researcher due to embarrassment. Future research should identify whether participants suffer from urinary incontinence. This recommendation has not been provided as a suggestion to exempt participants from research but to help the researcher with participant management. For example, it may be wise for the researcher to advise all participants to void before WBV exercise.

A minor side effect reported by participants, which has been reported in previous literature, was itching. It is unclear why this response occurs. As with the previous research this side effect presented as an acute response and was not long lasting, only occurring during the first one to two WBV sessions.

Limitations

Although calibration was performed on the platform, it was not performed with someone standing on the platform with a weight similar to the mean of the participants involved with the study. It is possible that a mass may have disrupted the calibration of the device. The use of an accelerometer would have been useful to determine whether a mass changed the acceleration of the platform.

Compliance by the community dwelling elderly females to the WBV sessions in this study was considered low. Participants failed to attend for several sessions for reasons such as viruses, holidays and sporting events. Five out of the 27 participants did not complete the minimum required 10 sessions, for each intervention period, and were therefore excluded from the analyses. The structure of this study required participants to attend the WBV sessions on Mondays, Wednesdays and Fridays. This structure was rigid and did not allow for flexibility of appointments. A more flexible schedule may have resulted in a higher compliance. However, this limitation may also be considered a strength of the research as it demonstrated that participants were still able to improve their fast walking velocity even though they did not attend all of the sessions.

The GAITRite electronic walkway provides information about the spatio-temporal parameters of an individual's gait. The parameters that were examined for the current study were walking velocity, stride length, stride time, stance time, single support time and double support time. Although examination of these parameters provided insight into the parameters associated with the faster walking velocity, they did not provide clear explanations why these changes occurred. For example, double support time decreased over time but it is unclear what mechanisms may have contributed to the change. This limitation provides the opportunity for future research in WBV and elderly females such as examining joint kinetics.

Based upon comments from participants during the pilot study, the activities-based confidence scale was included in the major study but failed to demonstrate any perceived improvements in balance for either group. Approximately 50 per cent of participants incorrectly answered the surveys. Many participants were disinterested in this part of the research and described it as “annoying”. The surveys were completed by participants at the end of testing sessions and were hastily completed by participants who were keen to leave. The surveys may have been more successful if they were completed at the start of the session or if the participants were allowed to take them home to complete them.

Although there were no reports of adverse effects during this study, conclusions of the safety of WBV exercise as an intervention for the elderly cannot be made. For example, it is possible that the vibration intensities currently reported in the literature (e.g. Roelants et al., 2004) and used in this research are creating micro-damage to the participant’s bones. The impact may take several years before it is identified. Longer term studies are needed to determine the safety of this form of exercise.

For control purposes a placebo with faint vibrations was compared with the WBV intervention. It would have been valuable to have a control group that was not exposed to the WBV or placebo. Although the placebo group did not demonstrate a significant improvement when compared to the WBV group, a “true” control would have indicated whether the subtle vibration used during the placebo had any impact on the gait of the participants.

Although the concept of progression is not new to exercise programming and has been used by previous researchers examining WBV exercise (e.g. Russo et al., 2003), the progressive increases of WBV intensity in this study is a limitation. Participants may have continued to improve their walking velocity without increasing the frequency and/or amplitude. Currently, there are no guidelines for the most appropriate intensities and progressions for WBV.

Finally, the WBV platform used for this research produced vibration via tilting oscillations. It is unclear whether vertical or tilting oscillations are the most effective for improving the walking velocity of elderly females. Compared to vertical oscillations, the tilting platform may enhance or hinder transmissibility impacting negatively or positively upon its efficacy. Furthermore, the pelvic rotations produced by the tilting platform may dampen the vibration impacting upon the efficacy of this form of vibration compared to vertical vibration.

Future Directions for WBV Research

Future research should attempt to identify the physiological mechanisms that contribute to the enhanced walking velocity. It is speculated that the enhanced walking velocity was the consequence of enhanced reflexive activity and muscle stiffness of the knee and ankle extensors. Research examining changes in reflex activity and stiffness of these muscle groups should therefore be performed to support or refute this hypothesis.

The current research was limited to elderly females since they are more susceptible to losing neuromuscular function due to the ageing process. Future research should not be limited to elderly females, and it is recommended that further work be performed examining the response of elderly males. Although older men also walk slower with age (e.g. Bohannon, 1997), it is possible that the intensities used during this research are not sufficient to stimulate improvements in the male elderly population.

Future research should also further examine the detraining response of older adults. As previously highlighted, the current study required participants to continue fast walks during each placebo session to identify the detraining response. It was expected that walking velocity would gradually decline and not remain. A future protocol could include a control group that experiences WBV during six weeks, followed by a six week placebo period without fast walks. This will provide further information pertaining to the fast walking maintenance of elderly females following a WBV program.

The inability of current literature to recommend appropriate protocols for WBV application to all human populations is a rich opportunity for further research. For older adults, researchers have not indicated the reasons why they have implemented the intensities used in their studies. Future research should examine a range of options to explore the most beneficial frequency and amplitude combinations. Although this research found that frequencies ranging between 10-30 Hz and amplitudes of 0.5-1.0 mm were sufficient to improve fast walking velocity, it did not show whether these relatively low

accelerations produce similar benefits for other functional tasks such as getting in and out of a chair? Other methodological questions include:

- Is it optimal to use one minute per set, as was used in the current research, or is it better to use longer or shorter durations?
- How many sets of WBV should the older adult experience in one session and how much recovery time should they have between sets?
- How many WBV sessions does the older adult need be exposed to each week to receive optimal benefits?
- Does elderly gait function respond differently to vertical and tilting WBV oscillations?

Conclusions

As a result of this research, a number of conclusions are made. The unique WBV platform (ACUWBV) that was developed provides reliable adjustments to WBV amplitude, without requiring the participant to change their stance posture. A distinguishing characteristic of this platform, compared to current commercially available platforms, is the ability to adjust the amplitude setting to zero and create a placebo effect. This method of placebo was shown to be effective as participants provided no indication that they were suspicious of being exposed to a placebo.

The WBV accelerations used in this research effectively improved the faster walking velocity of elderly females - despite being of lower intensities compared to previous research. A six week detraining period demonstrated these benefits to be maintained for at least this period of time. The improvements in walking velocity were primarily related to decreased stride time. Further analyses surprisingly indicated double support time, rather than single support time, to be the most likely contributor to the reduced stride time and therefore faster walking velocity. Over six week periods of progressively increasing WBV intensity, no major adverse effects were reported by participants, further supporting the low frequency/low amplitude intensities that were used for this research.

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Appendix A
Human Research Ethics Committee Approval Form

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National

Human Research Ethics Committee

Committee Approval Form

Principal Investigator/Supervisor: Dr Wayne Maschette Melbourne Campus
Co-Investigators: Melbourne Campus
Student Researcher: Christian Lorenzen Melbourne Campus

Ethics approval has been granted for the following project:
A study investigating the effect of whole body vibration on the gait of females aged 65-74 yrs of age.

for the period: 13.05.04 to 30.06.05

Human Research Ethics Committee (HREC) Register Number: V200304 82

The following standard conditions as stipulated in the *National Statement on Ethical Conduct in Research Involving Humans* (1999) apply:

(i) that Principal Investigators / Supervisors provide, on the form supplied by the Human Research Ethics Committee, annual reports on matters such as:
- security of records
- compliance with approved consent procedures and documentation
- compliance with special conditions, and

(ii) that researchers report to the HREC immediately any matter that might affect the ethical acceptability of the protocol, such as:
- proposed changes to the protocol
- unforeseen circumstances or events
- adverse effects on participants

The HREC will conduct an audit each year of all projects deemed to be of more than minimum risk. There will also be random audits of a sample of projects considered to be of minimum risk on all campuses each year.

Within one month of the conclusion of the project, researchers are required to complete a *Final Report Form* and submit it to the local Research Services Officer.

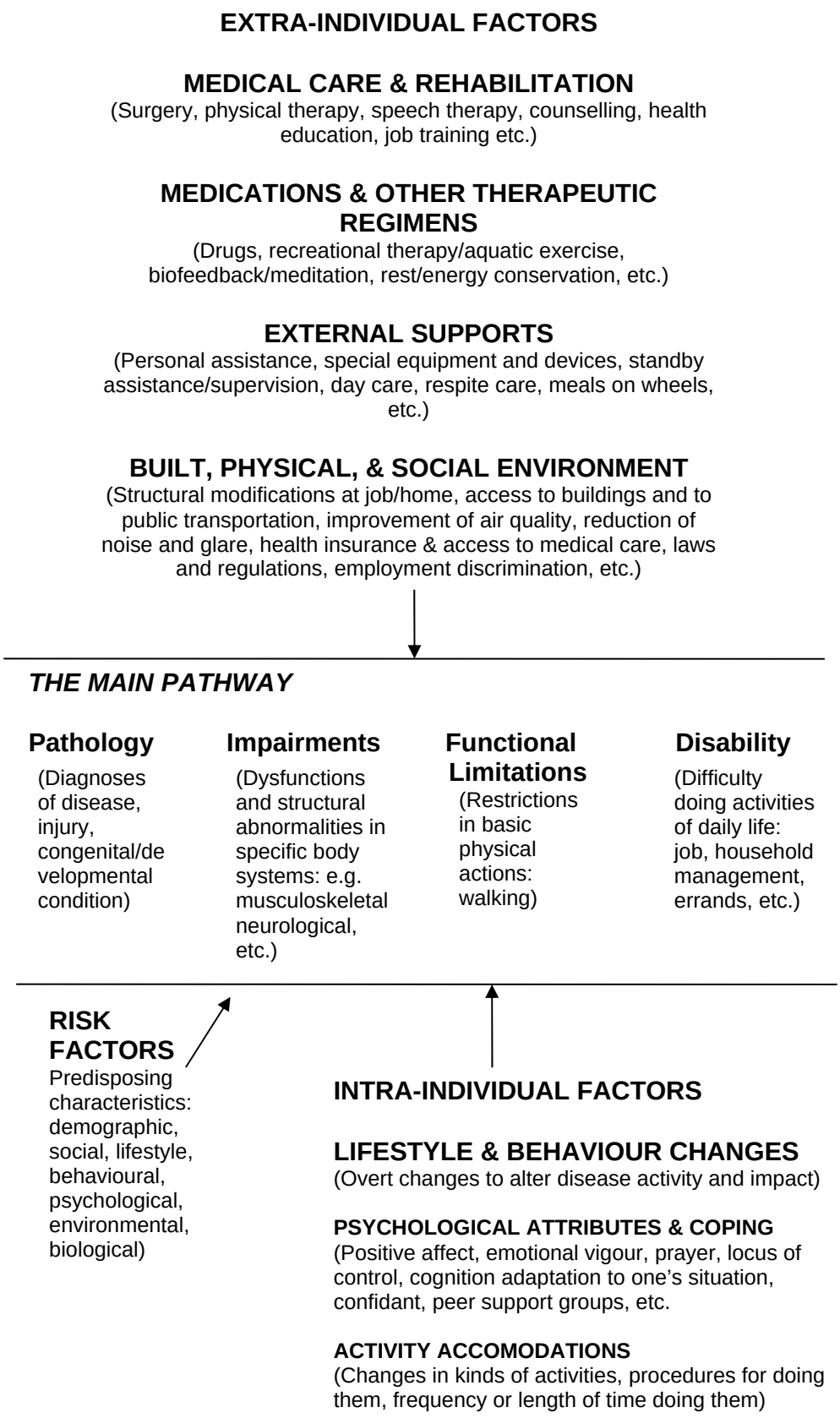
If the project continues for more than one year, researchers are required to complete an *Annual Progress Report Form* and submit it to the local Research Services Officer within one month of the anniversary date of the ethics approval.

Signed: ..... Date: 13.5.04
(Research Services Officer, Melbourne Campus)

(Committee Approval.dot @ 31/10/06) Page 1 of 1

Appendix B

The complete model of the Disablement Process (Verbrugge & Jette, 1994).



Appendix C

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National

INFORMATION LETTER TO PARTICIPANTS (Pilot Study)

TITLE OF PROJECT: A study investigating the effect of whole-body vibration on the gait of females aged 65-74 years of age.

NAMES OF STAFF SUPERVISORS: Dr. Wayne Maschette & Mr. Noel Lythgo.

NAME OF STUDENT RESEARCHERS: Mr. Christian Lorenzen (Ph.D. Candidate)

You are invited to be a participant in a study at the Australian Catholic University, Fitzroy, investigating the effect of whole body vibration on the walking pattern of older women. Whole body vibration (WBV), where the individual stands on a vibrating plate, is a recent technique that has been used to improve athletic performance (Figure 1). This study aims to identify whether WBV may improve the walking pattern of women aged 65-74 years of age.

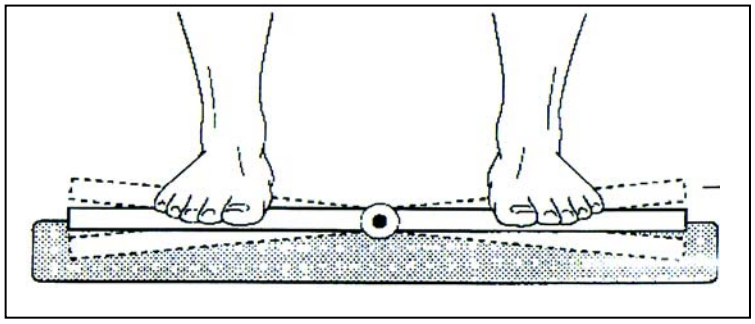


Fig 1: Representation of a typical whole-body vibration device.

If you choose to take part in this study, your participation will be required for five weeks in total. Figure 2 (page 2) outlines the sessions that you will be required to attend. You will be requested to attend a single session one week prior and one week following the WBV program. These sessions will be used to identify changes in your walking pattern after a WBV program. For these sessions, participants will be requested to walk over an electronic mat at a fast walking speed. The electronic mat will assess your walking pattern.

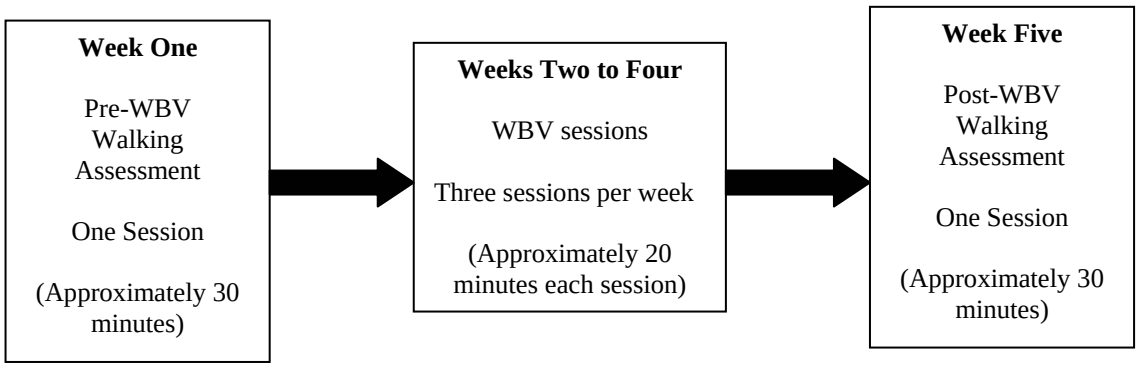


Figure 2: Outline of time you will be requested to invest if you choose to participate in this study.

Week's two to four will consist of the WBV program. Figure 3 outlines the procedure for each WBV session. You will experience five one minute WBV periods during each session. Each vibration period will be immediately followed by you walking over the electronic mat at a fast walking speed. During these three weeks, you will be required to attend three WBV sessions each week. Each session will last for approximately 20 minutes. The WBV will be specifically designed to strengthen the muscles of your legs.

The risks associated with this study are minimal. For your safety you will be requested to have a medical screening prior to your involvement. The cost of the medical examination will be met by the School of Exercise Science at the Australian Catholic University. If you have further queries on this issue, it should be directed to Dr. Wayne Maschette or Mr. Christian Lorenzen as listed below.

The benefits that have been associated with WBV for athletes include: (1) increased strength; (2) increased explosive power; (3) improved balance; and (4) improved flexibility. As yet, there have been no studies that have investigated the effect of WBV on older adults. It is speculated that older adults will experience similar benefits. Of these benefits this study will solely investigate changes in your functional performance (i.e. changes in walking). It is anticipated that the results of this study will be published. At all times the information you give will remain confidential. A coding system will be used to identify you and this will be destroyed at the conclusion of the study. It will not be possible to identify any participant in any report as only group data will be reported.

If during the project you feel uncomfortable in any way and no longer wish to continue you are free to withdraw at any time. Upon completion of the study tasks, or if you choose to withdraw from the study, you will be given the opportunity to ask questions regarding the project.

Dr. Rob Sands, a registered psychologist, is available for consultation in the unlikely event that you experience any emotional discomfort as a result of taking part in this study. Dr. Sands can be contacted on 9953 3034 (W) or 0418 459 857 (M).

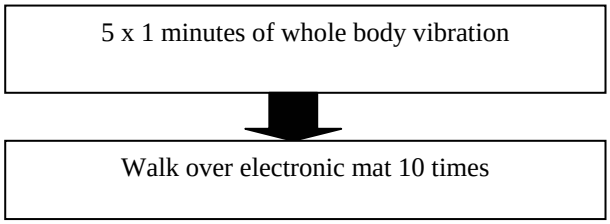


Figure 3: Outline of whole body vibration sessions.

Any questions regarding this project should be directed to the research supervisor.

Dr Wayne Maschette

on 9953 3040

in the School of Exercise Science

115 Victoria Parade Fitzroy Victoria 3065

OR

Mr Christian Lorenzen

on 9953 3419

in the School of Exercise Science

115 Victoria Parade Fitzroy Victoria 3065

At the conclusion of the study an information session will be held where group results will be produced. You will be cordially invited to attend that session and ask questions about the study (date to be advised).

This study has been approved by the Human Research Ethics Committee at Australian Catholic University, any complaints should be directed to:

Chair, HREC
C/o Research Services
Australian Catholic University
Locked Bag 4115
FITZROY VIC 3065
Tel: 03 9953 3157
Fax: 03 9953 3305

Any complaint made will be treated in confidence, investigated fully and the participant informed of the outcome.

Appendix D

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National

INFORMATION LETTER TO PHYSICIAN (Pilot Study)

TITLE OF PROJECT: A study investigating the effect of whole-body vibration on the gait of females aged 65-74 years of age.

NAMES OF STAFF SUPERVISORS: Dr. Wayne Maschette & Mr. Noel Lythgo.

NAME OF STUDENT RESEARCHERS: Mr. Christian Lorenzen (Ph.D. Candidate)

Whole body vibration (WBV) is a recent technique that has been used to improve athletic performance (Figure 1). The benefits that have been associated with WBV for athletes include: (1) increased strength; (2) increased explosive power; (3) improved balance; and (4) improved flexibility. As yet, there have been no studies to have investigated the effect of WBV on older adults. It is speculated that older adults will experience similar benefits. Of these benefits this study will solely investigate changes in functional performance (i.e. changes in gait).

This study aims to identify whether WBV may improve the gait of women aged 65-74 years of age. To participate in this study it is a requirement for all participants to complete a medical examination.

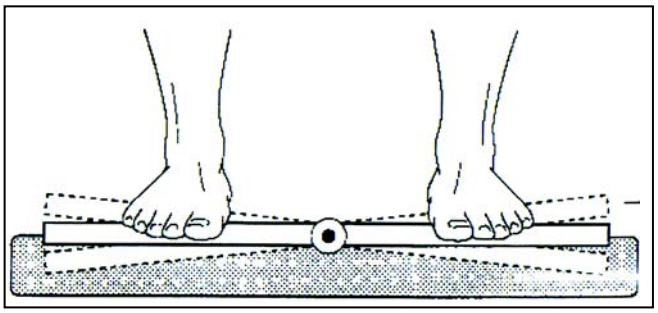


Fig 1: Representation of a typical whole-body vibration device.

Participant's will be required for five weeks in total. Figure 2 outlines the sessions that the participant's will be required to attend. They will be requested to attend a single session one week prior and one week following the WBV program. These sessions will be used to identify changes in walking pattern after a WBV program. For these sessions, participants will be requested to walk over an electronic mat at a fast walking speed. The electronic mat will assess walking pattern.

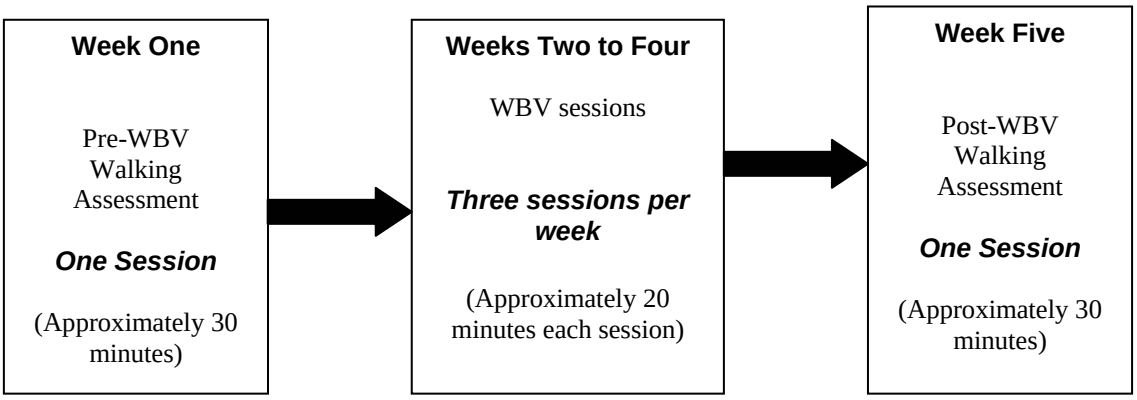


Figure 2: Outline of time invested by participants in this study.

Week's two to four will consist of the WBV program. Figure 3 outlines the procedure for each WBV session. Participants will experience five one minute WBV periods during each session. Each vibration period will be immediately followed by the participant walking over the electronic mat at a fast walking speed. During these three weeks, participants will be required to attend three WBV sessions each week. Each session will last for approximately 20 minutes. The WBV is specifically designed to strengthen the muscles of the participant's legs.

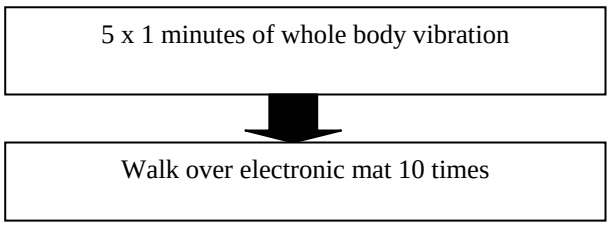


Figure 3: Outline of whole body vibration sessions.

The risks associated with this study are minimal. For the participant's safety it is at our request that they have a medical screening prior to their involvement. The cost of the medical examination will be met by the School of Exercise Science at the Australian Catholic University.

The participant has been requested to complete a health screening form which they have been asked to show to you during their medical screening. Space is provided at the end of the health screening form for you indicate any additional comments as well as your clearance of the participant to be involved in this study.

Any questions regarding this project should be directed to the research supervisor.

Dr Wayne Maschette

on 9953 3040

in the School of Exercise Science

115 Victoria Parade Fitzroy Victoria 3065

OR

Mr Christian Lorenzen

on 9953 3419

in the School of Exercise Science

115 Victoria Parade Fitzroy Victoria 3065

This study has been approved by the Human Research Ethics Committee at Australian Catholic University, any complaints should be directed to:

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C/o Research Services
Australian Catholic University
Locked Bag 4115
FITZROY VIC 3065
Tel: 03 9953 3157
Fax: 03 9953 3305

Thank you for your cooperation with this important research.

Yours sincerely

Dr. Wayne Maschette
Supervisor

Mr. Christian Lorenzen
Student Researcher

Appendix E

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National

Health Screening Form
(To be completed by participant & shown to physician)

Student Researcher: Christian Lorenzen

1. Personal Details

Name: _____

Address: _____

Telephone: _____ (AH) _____ (W)

Date of Birth: _____

Height: _____ cm

Weight: _____ kg

Have you ever been hospitalised?

If yes, when and why?

_____

_____

_____

Have you ever had any fractures? □ No □ Yes If yes, please specify:

_____

_____

Have you had any falls in the last 12 months ? □ No □ Yes

If yes, how many falls? ☐

If yes, specify the environment in which the fall occurred (e.g. stair, footpath etc.) and how it occurred (e.g. as a result of a trip or loss of balance)

_____

_____

_____

Do you currently experience pain in any of the following areas?

	No	Yes	If yes, specify frequency and type of pain
Lower back	□	□	_____
Hips	□	□	_____
Legs	□	□	_____
Knees	□	□	_____
Ankles	□	□	_____
Feet	□	□	_____

Do you experience pain when walking? □ No □Yes

Do you consider yourself to be:
- □ Inactive (no exercise)
- □ Slightly active (exercise 1-2 times per week)
- □ Active (exercise 3-4 times per week)
- □ Very active (exercise 5-7 times per week)

Do you have any of the following conditions?

	No	Yes	If yes, please specify
Musculo-skeletal dysfunction	□	□	_____
Neuromuscular dysfunction	□	□	_____
Overuse injuries	□	□	_____
Vascular disorders	□	□	_____
Traumatic injuries/surgeries	□	□	_____
Diabetes	□	□	_____
Arthritis	□	□	_____
Visual impairment	□	□	_____
Persistent vertigo	□	□	_____
Light-headedness	□	□	_____
Thrombosis	□	□	_____
Cardiovascular Conditions	□	□	_____
Epilepsy	□	□	_____
Other	□	□	_____

Are you currently on medication which affects your balance (e.g. hypnotics, sedatives such as benzodiepines)? □ No □Yes

If yes, please specify.

_____

Participant's Signature:_____ ***Date:_____***

Please show this form to your doctor and ask him/her to include a recommendation that it is safe for you to participate in this study as well as any additional comments in the space provided below.

_____

_____

_____

_____

_____

_____

_____

_____

Physician's Signature:_____ Date:_____

N.B. The following sections will be completed by the investigator.

2. Height (cm): ☐

3. Vision

3.1 Visual Acuity Test distance 6 m

Last line read ☐

3.2 Melbourne Edge Test ☐

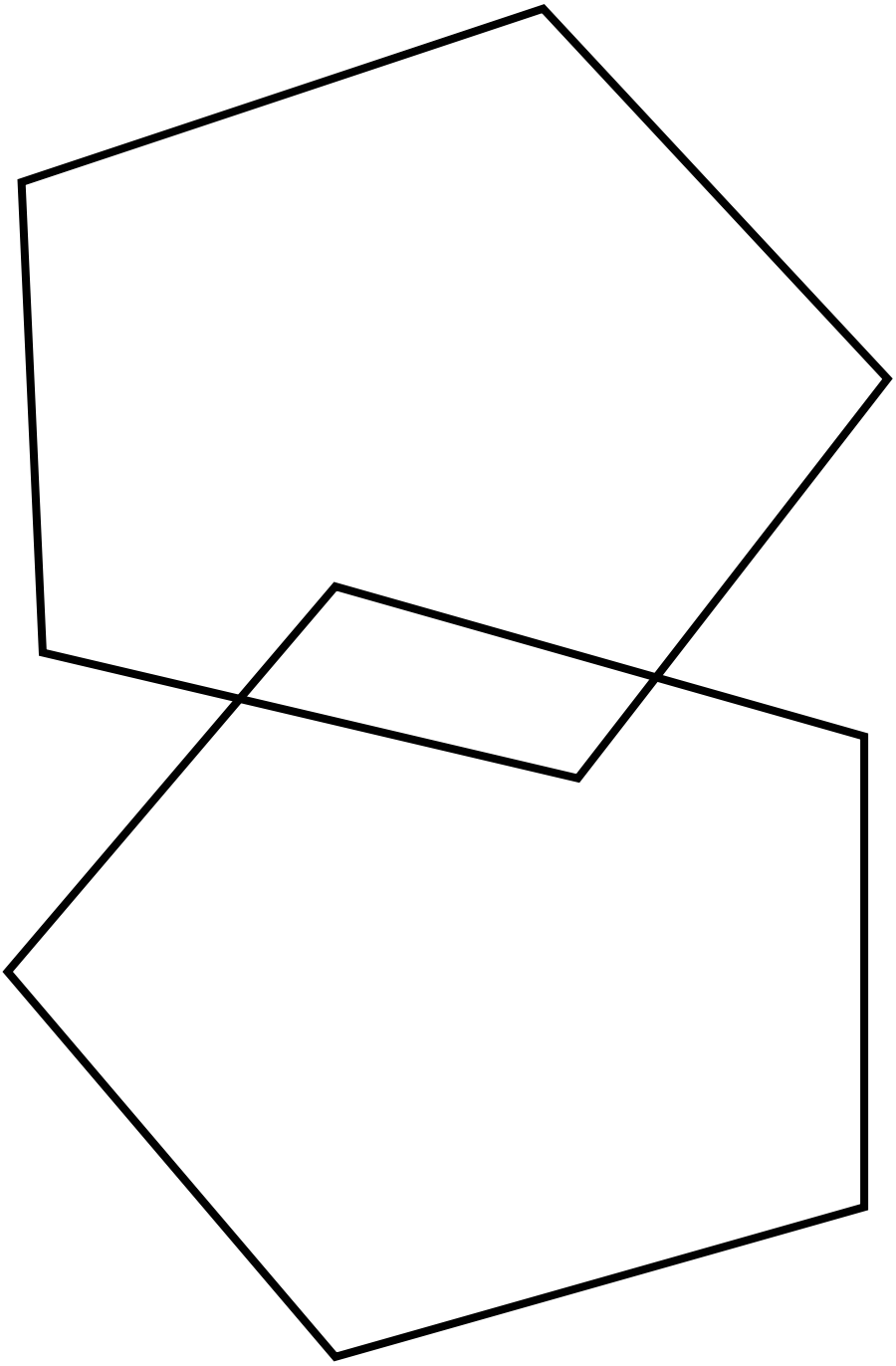
4. Romberg test – positive ☐ No ☐ Yes

Appendix F

Mini-Mental State Examination (MMSE)

Patient Code: _____
Instructions:
Score one point for each correct response within each question or activity.

Maximum Score	Patient's Score	Questions
5		What year is it? What season is it? What is the date? What day is it? What month is it?
5		Where are we now? Country? State? Suburb/City? Floor?
3		Name 3 objects: 1 second to say each. Then ask the patient all three after you have said them. Give 1 point for each correct answer. Then repeat them until he/she learns all 3. Count trial and record.
5		Count backwards by 7's from 100. Stop after 5 answers. Alternatively spell "world" backwards
3		Ask for the three objects repeated above. Give 1 point for each correct answer.
2		Name a pencil and a watch.
1		Repeat the following "No ifs , ands, or buts"
3		Follow a 3-stage command: "Take a paper in your hand, fold it in half, and put it on the floor"
1		Read and obey the following: Close your eyes
1		"Make up and write a sentence about anything" Sentence must contain a noun and a verb.
1		"Please copy this picture" (The examiner gives the patient a blank piece of paper and asks him/her to draw the symbol below. All 10 angles must be present and two must intersect.)
30		TOTAL



Appendix G

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National

INFORMATION LETTER TO PARTICIPANTS (Major Study)

TITLE OF PROJECT: A study investigating the effect of whole-body vibration on the gait of females aged 65 years and older.

NAMES OF STAFF SUPERVISOR: Dr. Wayne Maschette

NAME OF STUDENT RESEARCHERS: Mr. Christian Lorenzen (Ph.D. Candidate)

You are invited to be a participant in a study at the Australian Catholic University, Fitzroy, investigating the effect of whole body vibration on the walking pattern of older women. Whole body vibration (WBV), where the individual stands on a vibrating plate, is a recent technique that has been used to improve athletic performance (Figure 1). This study aims to identify whether WBV may improve the walking pattern of women aged 65 years and older.

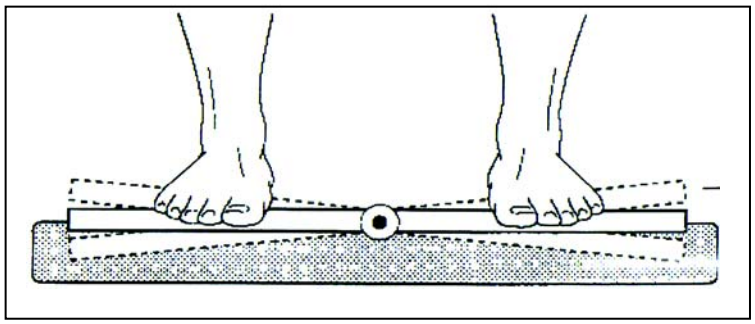


Fig 1: Representation of a typical whole-body vibration device.

If you choose to take part in this study, your participation will be required for fourteen weeks in total. Figure 2 (page 2) outlines the sessions that you will be required to attend. You will be requested to attend a single session one week prior and one week following the WBV program. These sessions will be used to identify changes in your walking pattern after a WBV program. For these sessions, participants will be requested to walk over an electronic mat at a fast walking speed. The electronic mat will assess your walking pattern.

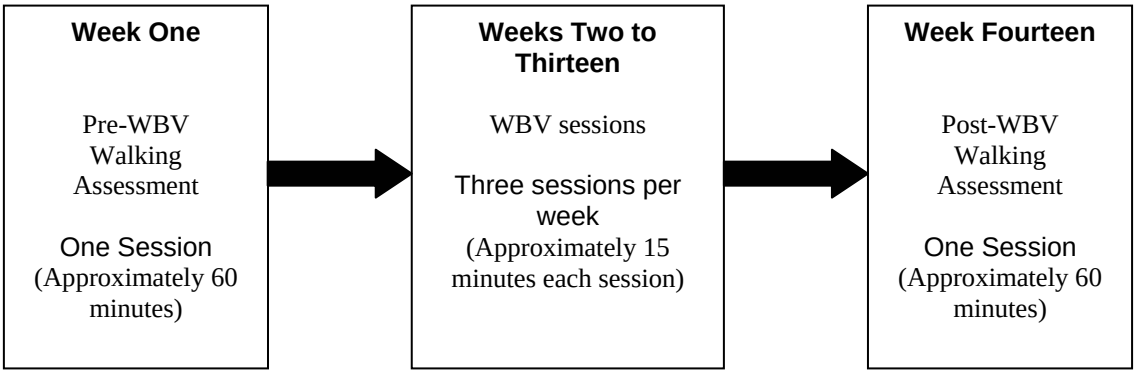


Figure 2: Outline of time you will be requested to invest if you choose to participate in this study.

Weeks two to thirteen will consist of the WBV program. Figure 3 outlines the procedure for each WBV session. You will experience five one minute WBV periods during each session. At the end of each WBV session, you will walk across the electronic mat at a fast speed. During these 12 weeks, you will be required to attend three WBV sessions each week. Each session will last for approximately 15 minutes. The WBV will be specifically designed to strengthen the muscles of your legs.

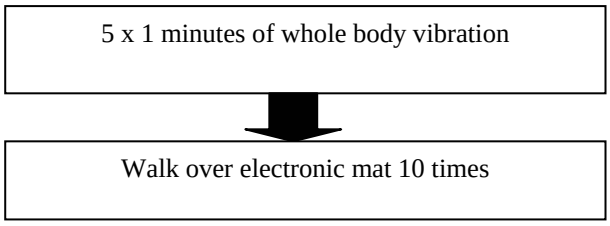


Figure 3: Outline of whole body vibration sessions.

The risks associated with this study are minimal. For your safety you will be requested to have a medical screening prior to your involvement. The cost of the medical examination will be met by the School of Exercise Science at the Australian Catholic University. If you have further queries on this issue, it should be directed to Dr. Wayne Maschette or Mr. Christian Lorenzen as listed below.

The benefits that have been associated with WBV for athletes include: (1) increased strength; (2) increased explosive power; (3) improved balance; and (4) improved flexibility. As yet, there have been no studies that have investigated the effect of WBV on older adults. It is speculated that older adults will experience similar benefits. Of these benefits this study will solely investigate changes in your functional performance (i.e. changes in walking). It is anticipated that the results of this study will be published. At all times the information you give will remain confidential. A coding system will be used to identify you and this will be destroyed at the conclusion of the study. It will not be possible to identify any participant in any report as only group data will be reported.

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in the School of Exercise Science

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on 9953 3419

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Appendix H

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National

INFORMATION LETTER TO PHYSICIAN (Major Study)

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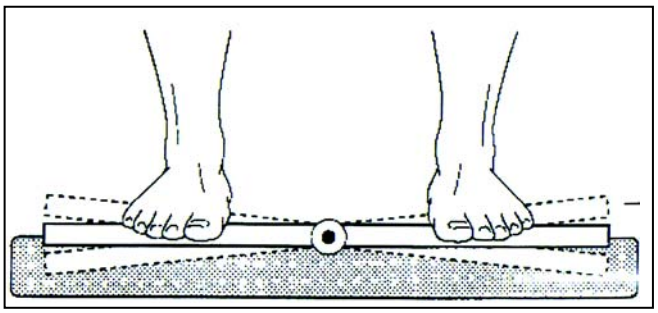

Fig 1: Representation of a typical whole-body vibration device.

Participants will be required for fourteen weeks in total. Figure 2 outlines the sessions that the participants will be required to attend. They will be requested to attend a single session one week prior and one week following the WBV program. These sessions will be used to identify changes in walking pattern after a WBV program. For these sessions, participants will be requested to walk over an electronic mat at a fast walking speed. The electronic mat will assess their walking pattern.

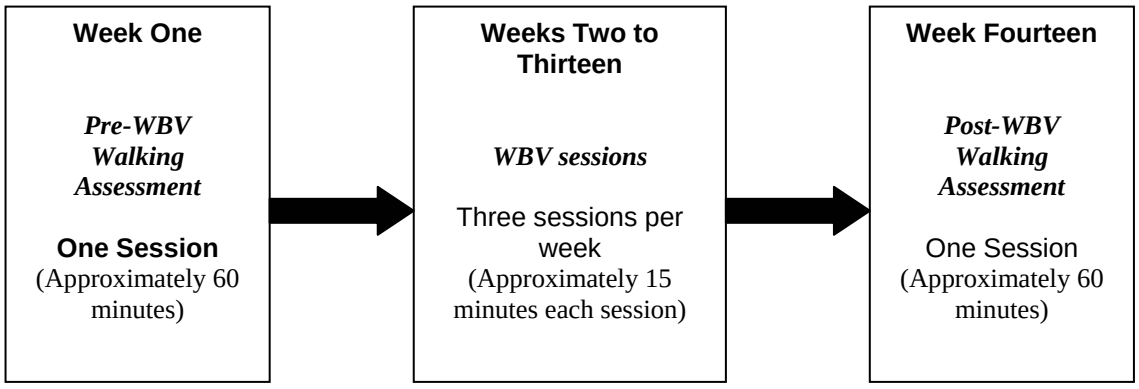

Figure 2: Outline of time invested by participants in this study.

Weeks two to thirteen will consist of the WBV program. Figure 3 outlines the procedure for each WBV session. Participants will experience five one minute WBV periods during each session. At the end of each WBV session, the participant will walk across the electronic mat at a fast walking speed. During these twelve weeks, participants will be required to attend three WBV sessions each week. Each session will last for approximately 15 minutes. The WBV is specifically designed to strengthen the muscles of the participant's legs.

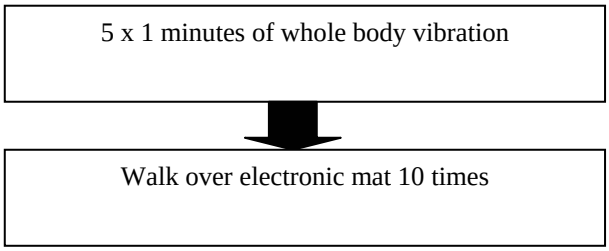

Figure 3: Outline of whole body vibration sessions.

The risks associated with this study are minimal. For the participant's safety it is at our request that they have a medical screening prior to their involvement. The cost of the medical examination will be met by the School of Exercise Science at the Australian Catholic University.

The participant has been requested to complete a health screening form which they have been asked to show to you during their medical screening. Space is provided at the end of the health screening form for you indicate any additional comments as well as your clearance of the participant to be involved in this study.

Any questions regarding this project should be directed to the research supervisor.

Dr Wayne Maschette

on 9953 3040

in the School of Exercise Science

115 Victoria Parade Fitzroy Victoria 3065

OR

Mr Christian Lorenzen

on 9953 3419

in the School of Exercise Science

115 Victoria Parade Fitzroy Victoria 3065

This study has been approved by the Human Research Ethics Committee at Australian Catholic University, any complaints should be directed to:

Chair, HREC
C/o Research Services
Australian Catholic University
Locked Bag 4115
FITZROY VIC 3065
Tel: 03 9953 3157
Fax: 03 9953 3305

Thank you for your cooperation with this important research.

Yours sincerely

Dr. Wayne Maschette
Supervisor

Mr. Christian Lorenzen
Student Researcher

Appendix I

Output Summary for Within-Subjects Effects for Stride Velocity

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Stride Velocity	Sphericity Assumed	.043	2	.022	3.548	.062	.372	7.095	.545
	Greenhouse-Geisser	.043	1.117	.039	3.548	.102	.372	3.962	.380
	Huynh-Feldt	.043	1.192	.036	3.548	.097	.372	4.227	.395
	Lower-bound	.043	1.000	.043	3.548	.109	.372	3.548	.355
Error (Stride Velocity)	Sphericity Assumed	.073	12	.006					
	Greenhouse-Geisser	.073	6.701	.011					
	Huynh-Feldt	.073	7.149	.010					
	Lower-bound	.073	6.000	.012					

a Computed using alpha = .05

Calculation of Effect Size Index (f) :

$$f = \sqrt{\frac{SS_b}{SS_e}}$$

$$f = \sqrt{\frac{0.043}{0.073}}$$

$$f = 0.77$$

Sample size was calculated using the effect size index ($f = 0.77$) with sample estimation table in Portney & Watkins (2000). The minimum number of participants required, according to this table was 8.

Appendix J

Output Summary for Within-Subjects Effects for Double Support Time

Source		Type III Sum of Squares	df	Mean Square	F	Sig.	Partial Eta Squared	Noncent. Parameter	Observed Power(a)
Stride Velocity	Sphericity Assumed	.002	2	.001	3.983	.047	.399	7.965	.597
	Greenhouse-Geisser	.002	1.159	.002	3.983	.083	.399	4.615	.427
	Huynh-Feldt	.002	1.262	.002	3.983	.078	.399	5.027	.451
	Lower-bound	.002	1.000	.002	3.983	.093	.399	3.983	.390
Error (Stride Velocity)	Sphericity Assumed	.003	12	.000					
	Greenhouse-Geisser	.003	6.952	.000					
	Huynh-Feldt	.003	7.574	.000					
	Lower-bound	.003	6.000	.001					

a Computed using alpha = .05

Calculation of Effect Size Index (f) :

$$f = \sqrt{\frac{SS_b}{SS_e}}$$

$$f = \sqrt{\frac{0.002}{0.003}}$$

$$f = 0.82$$

Sample size was calculated using the effect size index ($f = 0.82$) with sample estimation table in Portney & Watkins (2000). The minimum number of participants required, according to this table was 8.

Appendix K – Plate Accelerations for an amplitude of 0.25 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
0.25	10	1.0	0.1
0.25	11	1.2	0.1
0.25	12	1.4	0.1
0.25	13	1.7	0.2
0.25	14	1.9	0.2
0.25	15	2.2	0.2
0.25	16	2.5	0.3
0.25	17	2.9	0.3
0.25	18	3.2	0.3
0.25	19	3.6	0.4
0.25	20	3.9	0.4
0.25	21	4.4	0.4
0.25	22	4.8	0.5
0.25	23	5.2	0.5
0.25	24	5.7	0.6
0.25	25	6.2	0.6
0.25	26	6.7	0.7
0.25	27	7.2	0.7
0.25	28	7.7	0.8
0.25	29	8.3	0.8
0.25	30	8.9	0.9
0.25	31	9.5	1.0
0.25	32	10.1	1.0
0.25	33	10.8	1.1
0.25	34	11.4	1.2
0.25	35	12.1	1.2
0.25	36	12.8	1.3
0.25	37	13.5	1.4
0.25	38	14.3	1.5
0.25	39	15.0	1.5
0.25	40	15.8	1.6
0.25	41	16.6	1.7
0.25	42	17.4	1.8
0.25	43	18.3	1.9
0.25	44	19.1	1.9
0.25	45	20.0	2.0
0.25	46	20.9	2.1
0.25	47	21.8	2.2
0.25	48	22.7	2.3
0.25	49	23.7	2.4
0.25	50	24.7	2.5

Appendix L – Plate Accelerations for an amplitude of 0.5 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
0.5	10	2.0	0.2
0.5	11	2.4	0.2
0.5	12	2.8	0.3
0.5	13	3.3	0.3
0.5	14	3.9	0.4
0.5	15	4.4	0.5
0.5	16	5.1	0.5
0.5	17	5.7	0.6
0.5	18	6.4	0.7
0.5	19	7.1	0.7
0.5	20	7.9	0.8
0.5	21	8.7	0.9
0.5	22	9.6	1.0
0.5	23	10.4	1.1
0.5	24	11.4	1.2
0.5	25	12.3	1.3
0.5	26	13.3	1.4
0.5	27	14.4	1.5
0.5	28	15.5	1.6
0.5	29	16.6	1.7
0.5	30	17.8	1.8
0.5	31	19.0	1.9
0.5	32	20.2	2.1
0.5	33	21.5	2.2
0.5	34	22.8	2.3
0.5	35	24.2	2.5
0.5	36	25.6	2.6
0.5	37	27.0	2.8
0.5	38	28.5	2.9
0.5	39	30.0	3.1
0.5	40	31.6	3.2
0.5	41	33.2	3.4
0.5	42	34.8	3.6
0.5	43	36.5	3.7
0.5	44	38.2	3.9
0.5	45	40.0	4.1
0.5	46	41.8	4.3
0.5	47	43.6	4.4
0.5	48	45.5	4.6
0.5	49	47.4	4.8
0.5	50	49.4	5.0

Appendix M – Plate Accelerations for an amplitude of 0.75 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
0.75	10	3.0	0.3
0.75	11	3.6	0.4
0.75	12	4.3	0.4
0.75	13	5.0	0.5
0.75	14	5.8	0.6
0.75	15	6.7	0.7
0.75	16	7.6	0.8
0.75	17	8.6	0.9
0.75	18	9.6	1.0
0.75	19	10.7	1.1
0.75	20	11.8	1.2
0.75	21	13.1	1.3
0.75	22	14.3	1.5
0.75	23	15.7	1.6
0.75	24	17.1	1.7
0.75	25	18.5	1.9
0.75	26	20.0	2.0
0.75	27	21.6	2.2
0.75	28	23.2	2.4
0.75	29	24.9	2.5
0.75	30	26.7	2.7
0.75	31	28.5	2.9
0.75	32	30.3	3.1
0.75	33	32.3	3.3
0.75	34	34.2	3.5
0.75	35	36.3	3.7
0.75	36	38.4	3.9
0.75	37	40.5	4.1
0.75	38	42.8	4.4
0.75	39	45.0	4.6
0.75	40	47.4	4.8
0.75	41	49.8	5.1
0.75	42	52.2	5.3
0.75	43	54.8	5.6
0.75	44	57.3	5.8
0.75	45	60.0	6.1
0.75	46	62.7	6.4
0.75	47	65.4	6.7
0.75	48	68.2	7.0
0.75	49	71.1	7.2
0.75	50	74.0	7.5

Appendix N – Plate Accelerations for an amplitude of 1.0 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
1	10	3.9	0.4
1	11	4.8	0.5
1	12	5.7	0.6
1	13	6.7	0.7
1	14	7.7	0.8
1	15	8.9	0.9
1	16	10.1	1.0
1	17	11.4	1.2
1	18	12.8	1.3
1	19	14.3	1.5
1	20	15.8	1.6
1	21	17.4	1.8
1	22	19.1	1.9
1	23	20.9	2.1
1	24	22.7	2.3
1	25	24.7	2.5
1	26	26.7	2.7
1	27	28.8	2.9
1	28	31.0	3.2
1	29	33.2	3.4
1	30	35.5	3.6
1	31	37.9	3.9
1	32	40.4	4.1
1	33	43.0	4.4
1	34	45.6	4.7
1	35	48.4	4.9
1	36	51.2	5.2
1	37	54.1	5.5
1	38	57.0	5.8
1	39	60.1	6.1
1	40	63.2	6.4
1	41	66.4	6.8
1	42	69.7	7.1
1	43	73.0	7.4
1	44	76.5	7.8
1	45	80.0	8.2
1	46	83.6	8.5
1	47	87.2	8.9
1	48	91.0	9.3
1	49	94.8	9.7
1	50	98.7	10.1

Appendix O – Plate Accelerations for an amplitude of 2.0 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
2	10	7.9	0.8
2	11	9.6	1.0
2	12	11.4	1.2
2	13	13.3	1.4
2	14	15.5	1.6
2	15	17.8	1.8
2	16	20.2	2.1
2	17	22.8	2.3
2	18	25.6	2.6
2	19	28.5	2.9
2	20	31.6	3.2
2	21	34.8	3.6
2	22	38.2	3.9
2	23	41.8	4.3
2	24	45.5	4.6
2	25	49.4	5.0
2	26	53.4	5.4
2	27	57.6	5.9
2	28	61.9	6.3
2	29	66.4	6.8
2	30	71.1	7.2
2	31	75.9	7.7
2	32	80.9	8.2
2	33	86.0	8.8
2	34	91.3	9.3
2	35	96.7	9.9
2	36	102.4	10.4
2	37	108.1	11.0
2	38	114.0	11.6
2	39	120.1	12.2
2	40	126.4	12.9
2	41	132.8	13.5
2	42	139.3	14.2
2	43	146.0	14.9
2	44	152.9	15.6
2	45	159.9	16.3
2	46	167.1	17.0
2	47	174.5	17.8
2	48	182.0	18.5
2	49	189.6	19.3
2	50	197.4	20.1

Appendix P – Plate Accelerations for an amplitude of 3.0 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
3	10	11.8	1.2
3	11	14.3	1.5
3	12	17.1	1.7
3	13	20.0	2.0
3	14	23.2	2.4
3	15	26.7	2.7
3	16	30.3	3.1
3	17	34.2	3.5
3	18	38.4	3.9
3	19	42.8	4.4
3	20	47.4	4.8
3	21	52.2	5.3
3	22	57.3	5.8
3	23	62.7	6.4
3	24	68.2	7.0
3	25	74.0	7.5
3	26	80.1	8.2
3	27	86.4	8.8
3	28	92.9	9.5
3	29	99.6	10.2
3	30	106.6	10.9
3	31	113.8	11.6
3	32	121.3	12.4
3	33	129.0	13.2
3	34	136.9	14.0
3	35	145.1	14.8
3	36	153.5	15.7
3	37	162.2	16.5
3	38	171.1	17.4
3	39	180.2	18.4
3	40	189.5	19.3
3	41	199.1	20.3
3	42	209.0	21.3
3	43	219.0	22.3
3	44	229.4	23.4
3	45	239.9	24.5
3	46	250.7	25.6
3	47	261.7	26.7
3	48	272.9	27.8
3	49	284.4	29.0
3	50	296.2	30.2

Appendix Q – Plate Accelerations for an amplitude of 4.0 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
4	10	15.8	1.6
4	11	19.1	1.9
4	12	22.7	2.3
4	13	26.7	2.7
4	14	31.0	3.2
4	15	35.5	3.6
4	16	40.4	4.1
4	17	45.6	4.7
4	18	51.2	5.2
4	19	57.0	5.8
4	20	63.2	6.4
4	21	69.7	7.1
4	22	76.5	7.8
4	23	83.6	8.5
4	24	91.0	9.3
4	25	98.7	10.1
4	26	106.8	10.9
4	27	115.1	11.7
4	28	123.8	12.6
4	29	132.8	13.5
4	30	142.2	14.5
4	31	151.8	15.5
4	32	161.7	16.5
4	33	172.0	17.5
4	34	182.6	18.6
4	35	193.5	19.7
4	36	204.7	20.9
4	37	216.2	22.0
4	38	228.1	23.3
4	39	240.2	24.5
4	40	252.7	25.8
4	41	265.5	27.1
4	42	278.6	28.4
4	43	292.1	29.8
4	44	305.8	31.2
4	45	319.9	32.6
4	46	334.2	34.1
4	47	348.9	35.6
4	48	363.9	37.1
4	49	379.2	38.7
4	50	394.9	40.3

Appendix R – Plate Accelerations for an amplitude of 5.0 mm

Amplitude (mm)	Frequency (Hz)	Acceleration Max. ($m.s^{-2}$)	Gravitational Force (g)
5	10	19.7	2.0
5	11	23.9	2.4
5	12	28.4	2.9
5	13	33.4	3.4
5	14	38.7	3.9
5	15	44.4	4.5
5	16	50.5	5.2
5	17	57.1	5.8
5	18	64.0	6.5
5	19	71.3	7.3
5	20	79.0	8.1
5	21	87.1	8.9
5	22	95.6	9.7
5	23	104.4	10.6
5	24	113.7	11.6
5	25	123.4	12.6
5	26	133.5	13.6
5	27	143.9	14.7
5	28	154.8	15.8
5	29	166.0	16.9
5	30	177.7	18.1
5	31	189.7	19.3
5	32	202.2	20.6
5	33	215.0	21.9
5	34	228.2	23.3
5	35	241.9	24.7
5	36	255.9	26.1
5	37	270.3	27.6
5	38	285.1	29.1
5	39	300.3	30.6
5	40	315.9	32.2
5	41	331.9	33.8
5	42	348.3	35.5
5	43	365.1	37.2
5	44	382.3	39.0
5	45	399.8	40.8
5	46	417.8	42.6
5	47	436.2	44.5
5	48	454.9	46.4
5	49	474.1	48.3
5	50	493.6	50.3