THE EFFECTS OF INCREASING BODY WEIGHT SUPPORT ON GAIT KINEMATICS AND TIBIAL SHOCK DURING TREADMILL RUNNING

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Assentation of authorship

I hereby declare that this submission is my own work and that to the best of my knowledge and belief, it contains no material previously published or written by another person nor material which to a substantial extent has been accepted for the qualification of any degree or diploma of a university or other institution of higher learning, except where due acknowledgement is made.

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Ethical Approval

Ethical approval for this research was obtained from the AUT Ethics Committee (reference 13/198) on the 22 August 2013 (see Appendix 1).

Abstract

Lower body positive pressure (LBPP) treadmills have become a popular modality for both training and return to sport rehabilitation. The literature outlining the efficacy of LBPP treadmills for training and rehabilitation in elite athletic groups is limited. The purpose of this research was to examine the effects body weight support (BWS) treadmill running has on the kinematic variables of running gait; and on tibial shock during submaximal treadmill running in a cohort of elite athletes. Twenty elite (male =9, female = 11), international level athletes who represented New Zealand in a variety of team and individual Olympic sports volunteered to participate in this study. A repeated-measures design, using quantitative measures, was performed to determine changes in tibial accelerometery and sagittal plane gait kinematics while running on the Alter-G Trainer treadmill. Tibial acceleration was used to determine tibial shock and two-dimensional video analysis was performed to determine sagittal plane gait kinematics. All variables were assessed at 60, 70, 80 and 90% of body weight, and were measured against the control condition of 100% body weight. Tibial shock reduced as BWS increased, however not linearly. Very large and large effect sizes (ES) were observed for 60% condition (ES = 0.71), and 70% conditions (ES = 0.59) respectively. A moderate change for 80% condition (ES = 0.35) was observed while only a trivial change in mean was observed for the 90% condition (ES =0.09). Trivial to small changes were observed for changes in ankle angles across all conditions, with the exception of Initial Contact and Mid-stance angles at 60% conditions. Large and very large changes were observed for tROM at 60% (ES = 0.51) and 70% (ES = 0.79). Stride rate and foot strike patterns were also affected at the 60% condition. The change at 60% for foot strike was extremely large (ES = 1.29). The incidence of fore-foot strike (FFS) increased by 30% at 60% condition but not at lower levels of BWS. Running with BWS on a LBPP treadmill is effective for unloading the magnitude of tibial shock in the lower limb. LBPP treadmills may be an effective piece of equipment for rehabilitation purposes with appropriate consideration of kinetic and kinematic variables and how these interact, relative to the lower limb injury.

Chapter 1: Introduction

Running is used frequently in elite level sports, as a mode of physical conditioning. Large volume and high intensity running protocols are used across many sports to improve athlete fitness. One of the ongoing challenges for optimising sports performance is balancing training volume and intensity with recovery and adaptation to achieve the best possible adaptation (Smith, 2003). When training stress exceeds adaptation, the risk of the athlete obtaining a musculoskeletal overuse injury increases (Kibler, Chandler, & Stracener, 1992; Wilder & Sethi, 2004).

As running is a highly utilised modality of training for both physical conditioning and rehabilitation programmes, elite and professional sporting bodies continually seek world-leading technologies to support the need to push athletes to higher levels physically. Enabling the athlete to run at faster velocities, in a supported environment or with reduced impact forces offers variety to athletic programs. Body weight support (BWS) systems such as harness devices and lower body positive pressure (LBPP) treadmills enable these training variations. Multifunctional devices that can be implemented into the training environment to support efficient return to sport rehabilitation and performance plans are preferred. The effects of using advanced technical pieces of training equipment for rehabilitation and training purposes requires further exploration in the elite sporting population.

Equipment and technology designed to reduce impact forces during running, by supporting body weight, have become more common in sports training facilities in the last decade. Traditionally BWS modalities such as harness systems or aquatic jogging have been used in clinical populations for rehabilitation purposes (Aaslund, Helbostad, & Moe-Nilssen, 2011; Aaslund & Moe-Nilssen, 2008). Harness systems and other unloading devices have been shown to reduce effective ground forces experienced by the body in older patients and specific clinical groups (Aaslund et al., 2011; Hesse, Uhlenbrock, & Sarkodie-Gyan, 1999). However, less is known about the effect BWS devices have on the kinematic and kinetic variables of running in highly trained athletes.

Traditional rehabilitation strategies for return to running may be considered conservative in elite and professional sporting environments. Complete rest (Tolbert & Binkley, 2009), ergogenic aids (Wilder & Sethi, 2004) and graded running programmes (Winters et al., 2013) have been implemented in return to sport strategies following lower limb overuse injuries. Graded walking and running strategies have been shown to be more effective than complete rest (Moen et al., 2012). However, these strategies may not be the most effective option to stimulate central or peripheral physiological or neuromuscular adaptations required to retrain the athlete to an adequate level to return to play. Return to play strategies for many lower limb and sport related injuries remain inconclusive and conflicting in the literature (Galbraith & Lavallee, 2009; Winters et al., 2013). In order to effectively out-weigh participation risk with the advantages of return to training or play, decision-making models have been established to better determine and support the rationale for return to sport (Creighton, Shrier, Shultz, Meeuwisse, & Matheson, 2010). A crucial decision making process should be established, to minimise the risk of re-injury or regression, which is known to occur when rehabilitation strategies are fast tracked or poorly managed (Creighton et al., 2010).

1.1 Background to the Thesis

Lower body positive pressure treadmills have become a popular modality for both training and return to sport rehabilitation. The literature outlining the efficacy of LBPP treadmills for training and rehabilitation in elite athletic groups is limited. The purpose behind the LBPP treadmill is to provide the user with an environment where they can maintain normal locomotive patterns, while reducing impact forces on the body.

Both vertical and horizontal ground reaction forces impact the body during walking and running at different magnitudes and vectors dependant on the stage of the gait cycle. Passive forces occur at the initial contact phase, while peak forces occur during midstance of the gait cycle. This is the impact and active peaks seen in Figure 1.1 (Fahlam, 2013)

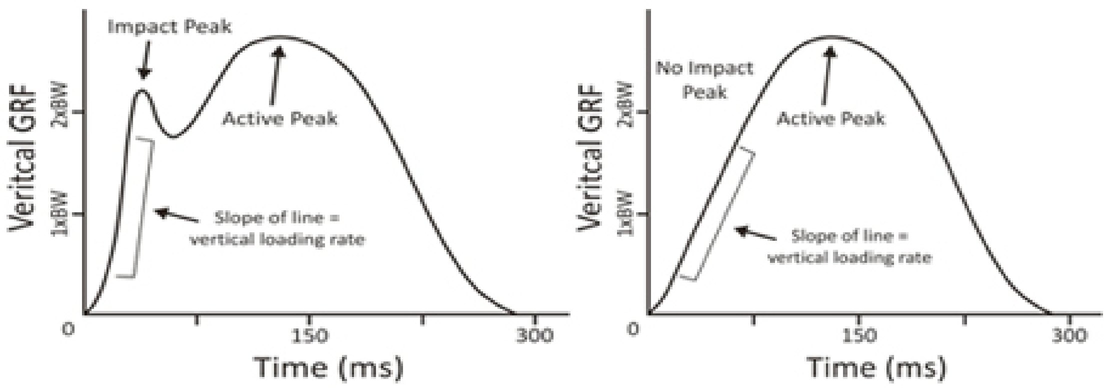

Figure 1.1 Foot strike force trace during rear (left) and fore (right) foot running

There are three main foot strike strategies observed during running which include fore-, mid- and rear foot strike. Rear foot or heel strike (RFS) patterns normally result in a more prominent impact peak when compared to a mid- or forefoot strike (FFS) patterns (Figure 1.1). Peak loading of the lower limb occurs during mid-stance, which is termed the active peak (Helseth, Hortobágyi, & DeVita, 2008; Støren, Helgerud, & Hoff, 2011). Impact forces acting on the body are distributed through the feet to the tibia. The transmitted force from the foot to the tibia attenuated only by shoe technology and surrounding active musculature, is known as tibial shock (Mizrahi, Verbitsky, & Isakov, 2000; Mizrahi, Verbitsky, Isakov, & Daily, 2000). Tibial shock has been linked to lower limb overuse injuries in elite athletes (Ferber, Hreljac, & Kendall, 2009b; Hreljac, 2004; Hreljac, Marshall, & Hume, 2000; Mizrahi, Verbitsky, & Isakov, 2000; Stanish, 1984; Wen, 2007). The distal tibia may experience compressive forces up to 14 times body weight (Sasimontonkul, Bay, & Pavol, 2007; Sasimontonkul & Puttapitukporn, 2007). With up to half of the reported stress fractures in elite athletes occurring in the distal end of the tibia, it is important to understand the factors which influence the magnitude of tibial shock (Mizrahi et al., 2000). Running velocity, body mass, foot-strike and fatigue are some of these factors (Laughton, Davis, & Hamill, 2003; Olin & Gutierrez, 2013; Rooney & Derrick, 2013; Stacoff, Nigg, Reinschmidt, van den Bogert, & Lundberg, 2000). At increasing running speeds as observed in sprint running, peak vertical and horizontal forces may increase by over 14% and 30% respectively (Brughelli, Cronin, & Chaouachi, 2011). In habituated FFS versus RFS runners, lower limb loading, impact peak and resultant peak forces from RFS patterns are greater than in FFS runners. Impact peaks are attenuated during FFS running, however active peaks tend to be greater (Lieberman et al., 2010; Rooney & Derrick, 2013). Having a better understanding of the effect BWS running has on foot strike patterns and tibial shock may be useful for the practitioner to more accurately prescribe rehabilitation programmes specific to injury and related injury mechanisms.

Impact forces experienced during human locomotion can be directly measured in sports science research using force platforms or force transducer technology. Accelerometery has also been deemed an accurate and reliable mode of detecting gait events, such as tibial acceleration to determine tibial shock (Fong & Chan, 2010; Sinclair, Hobbs, Protheroe, Edmundson, & Greenhalgh, 2013). Measured using non-invasive accelerometery technology, tibial shock has been correlated with vertical ground reaction forces during human locomotion (Elvin, Elvin, & Arnoczky, 2007; Lafortune, Lake, & Hennig, 1995; Olin & Gutierrez, 2013). Due to the importance of understanding the magnitude of tibial shock when prescribing training protocols for conditioning and rehabilitation purposes, it has also been used to measure impact during research addressing methods of attenuating shock during running (Greenhalgh, Sinclair, Leat, & Chockalingam, 2012; Olin & Gutierrez, 2013). Determining the magnitude of tibial shock and rate of attenuation during BWS running may assist the practitioner to more accurately prescribe training protocols and monitor progress for the recovering athlete. The resultant peak tibial accelerations, represented as the area under the curve and an indicator of loading rate through the lower limb, were used in this thesis as a measure of tibial shock.

There are three main phases to the running gait cycle, swing, stance and flight phase. Kinematic characteristics of stride during running include foot strike, ground contact time, stride rate, stride length, joint angles and velocities and flight time (Dugan & Bhat, 2005; Nicola & Jewison, 2012). Interactions of these may vary between athlete, sport and event demands. Throughout a marathon running event of 42.2 kilometres for example, athletes may increase stride frequency as stride length decreases with increasing levels of fatigue (Nicol, Komi, & Marconnet, 1991). Lower limb kinematic and kinetic characteristics observed in a 100m sprinter include faster hip angular velocity, and reduced braking forces upon ground contact respectively, when compared to the marathon runner (Maćkała, Fostiak, & Kowalski, 2015; Nagahara, Matsubayashi, Matsuo, & Zushi, 2014). In the distance runner, there is less knee flexion, and therefore a longer moment arm, during the recovery phase of the gait cycle. This contributes to the mechanical and metabolic efficiency required for long distance events, as compared to the sprint athlete (Bushnell & Hunter, 2007; Hunter, 2004). The inverse relationship between angular hip velocity and braking forces, combined with a shorter moment arm during the recovery phase allows for a higher stride rate in the sprint athlete (Bushnell & Hunter, 2007). These examples demonstrate the polarities in elite running events, however there is a diverse range of running styles and therefore kinetic and kinematic variability between elite level athletes across sporting codes.

1.2 Aim of the thesis

As LBPP treadmills such as the AlterG (Woodway Inc., CA) are becoming more commonly used in elite training and rehabilitation facilities, it is important to determine the effect supporting body weight has on lower limb running gait kinematics, and tibial shock. As running is a highly developed skill for elite athletes, particularly those who specialise in running events, it is important to determine if any technical changes occur with submaximal velocity, BWS treadmill running. It is known that non-preferred running technique and velocities can negatively influence impact load and potentially injurious soft tissue vibration (Enders, von Tscharner, & Nigg, 2014). Many return to running rehabilitation programs are often targeted at reducing impact forces following lower limb pathology so determining the kinematic effects in conjunction with shock attenuation is also important. A recent study outlined the accuracy of BWS provided using the AlterG anti-gravity treadmill (McNeill, de Heer, Bounds, & Coast, 2015). The device provided approximately 5% more support than set using the treadmill computer interface at 100% body weight, and at 40% or more BWS. This knowledge, in conjunction with determining other biomechanical changes as a result of unloading body mass, should be considered when designing elite level athletic programs. It is also known that differences exist in lower limb kinematics between male and female recreational athletes during running activities (Beaulieu, Lamontagne, & Xu, 2009; Ferber, McClay Davis, & Williams Iii, 2003; Maurer, Federolf, von Tscharner, Stirling, & Nigg, 2012); and in elite level soccer players during running and cutting manoeuvres (Beaulieu et al., 2009). Research suggests females to be at greater risk of tibial stress related injuries than their male counterparts (Milner, Ferber, Pollard, Hamill, & Davis, 2006). In addition to exploring the effects BWS might have on running technique and tibial shock, gender comparisons were also assessed. Further, athletes from different cohorts perform a variety of running modalities during training and competition. As there is a great amount of physiological and biomechanical variability in these running modalities, it may be beneficial to determine if any differences between team and individual sport athletes exists. Team sport athletes may require a combination of aerobic efficiency and repeat sprint ability dependant on their sport and positional characteristics (Bishop & Spencer, 2004). The kinematic and kinetic differences between sprint running and endurance running may be evident in team sport athletes due to the physical demands of their sport. As such, comparisons between team sport and individual athletes, was assessed in this thesis.

1.3 Purpose Statement

The purpose of this research was to (1) examine the effect BWS treadmill running has on kinematic variables of running gait in elite athletes; (2) to assess the effect of BWS on tibial shock during submaximal treadmill running; and (3) to determine if gender or athlete classification differences exist during BWS treadmill running.

1.4 Significance of research

Improved knowledge into the effect of body weight unloading on kinematic and kinetic variables in the elite athletic population, and across a variety of running based sporting codes may guide return to sport and strength and conditioning practices towards more effective rehabilitation and physical performance programming. Practitioners involved in the implementation of athletic rehabilitation training protocols are interested in identifying and improving factors that limit the efficacy and safety of return to sport programmes. This investigation aims to provide insight into the possible effects BWS treadmill running has on tibial shock and gait kinematics in elite athletes. Further, this study will also determine if any gender differences or athlete classification differences exist. The significance of this to the practitioner is to determine if BWS running is the most appropriate method of training to unloading and/or reloading recovering tissue. This has the potential for improving best practice protocols for return to sport rehabilitation and ultimately, athlete performance.

1.5 Study delimitations

Age, training experience and musculoskeletal health was delimited within this study so as to achieve an elite standard cohort.

- Age was limited to between 18-30 years of age.
- Training experience was limited to New Zealand national level athletes or above.
- Participants had to be engaged in full training 3 weeks prior to data collection,
- Participants with any illness or injury were excluded from the study

Chapter 2: Literature Review

2.1 Introduction

Running is a sport itself as well as one of the most common modalities of training for many elite sporting programmes. As an individual sport, running spans from high velocity, short distance sprint events such as the 100m sprint right through to the 42.2km marathon, and even elite level ultra-distance events which can cover up to and over 160km. As an effective mode of physical conditioning, running is not restricted solely to the running specialist but is used largely by athletes from many Olympic and Professional sports.

There is considerable biomechanical and physiological variability between running modalities, sports, and athletes (Bushnell & Hunter, 2007; Hamilton, Nevill, Brooks, & Williams, 1991). Running involves the complex integration of many body systems, from physiological interactions of the brain, central nervous system, cardiovascular and neuromuscular systems (Barnes & Kilding, 2015). The following chapter focuses on running biomechanics, epidemiology of lower limb overuse injuries commonly seen in elite level athletes and the effect of BWS mechanisms on running biomechanics.

2.2 Search strategies

The following literature review was conducted and written as a narrative review. The purpose of the following search strategies was to gather as much information regarding lower limb loading during running, related information regarding injury risk and mechanism and BWS running as possible. For the purpose of this review, research studies were identified via the following electronic databases: MEDLINE, SCOPUS, SPORTDiscus, and BiomechLibrary, AUTSummon. The following keywords were used: treadmill running, running, body weight support, BWS, unweighted ambulation, lower body positive pressure, Alter-G Trainer, gait, kinetics and kinematics, tibial shock, tibial acceleration, LBPP. Screening criteria initially was limited to (1) use of BWS devices; (2) non-injured; (3) elite or highly trained athletes and (4) biomechanical and/or neuromuscular characteristics of BWS running. Due to the scarcity of literature pertaining to highly trained athlete groups, the search criteria was expanded to include other participant groups. These cohorts included disabled, injured, recreationally trained and elderly populations. Body weight support systems were categorised into devices including harness systems, aquatic systems and positive pressure treadmills.

2.3 Biomechanics of running

The biomechanics of running are highly variable between athletes due to individual neuromuscular aspects and musculoskeletal structures (Hamill, van Emmerik, Heiderscheit, & Li, 1999). Running gait is a cyclic motor pattern and c be divided into two main phases. These are the stance and swing phases. The stance phase may be further divided into initial contact, loading response, mid- and terminal stances (Novacheck, 1998). The swing phase may also be further subdivided into toe-off, initial-, mid- and terminal- swing according to Novacheck (1998). Figure 2.1 shows the sub divisions of stance and swing phases of the gait cycle (Carr, 2010).

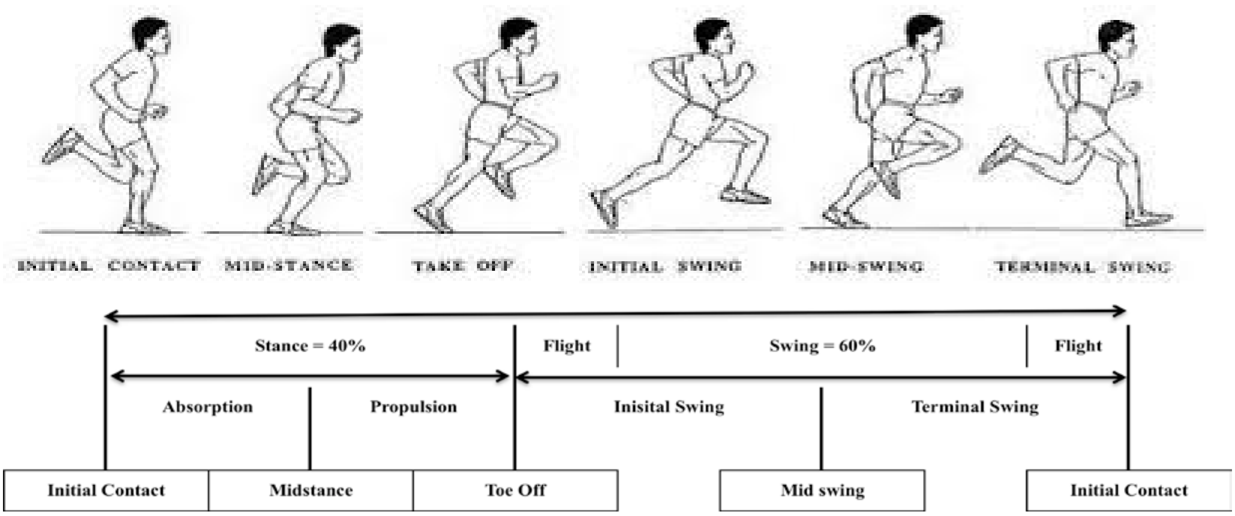

Figure 2.1 Phases of running gait. Adapted from Car (2010)

During initial contact, strike patterns are determined. Runners may be classified into one of three main types of foot strike. Rear-foot, mid-foot and fore-foot strikers (Novacheck, 1998). Running velocity, training experience and running distance may all influence foot strike patterns in athletes (Hatala, Dingwall, Wunderlich, & Richmond, 2013). A rear-foot strike (RFS) pattern is identified when the outside of the heel makes initial contact with the running surface. It is reported that approximately 75% of runners are RFS runners (Hasegawa, Yamauchi, & Kraemer, 2007). Impact peaks during rear-foot striking may range from 1.5 to 3 times body weight (Lieberman et al., 2010), with large horizontal breaking forces as a result of the foot making contact with the ground in front of the body’s centre of mass (COM) (Nicola & Jewison, 2012; Novacheck, 1998). Longer stance times have also been reported during RFS patterns as compared with mid- and fore-foot patterns (Mercer, Applequist, & Masumoto, 2013).

Forefoot strike occurs when the heads of the metatarsals make initial ground contact, as seen in sprint runners. The heel of the foot may slightly ‘kiss’ the ground during mid-stance, but often does not make contact with the ground during initial contact (Laughton, Davis, & Hamill, 2003) as the fore-foot tends to land beneath the athlete’s COM which is travelling forwards (Hatala et al., 2013; Novacheck, 1998). The impact peak seen in the vertical ground reaction force curve of a FFS runner is much smaller than that of a RFS runner due to the eccentric activity from the posterior leg musculature. Impact shock during FFS is attenuated with this eccentric activity due to the decrease of the vertical velocity of the athletes COM (Laughton et al., 2003). Mid-foot strike occurs approximately half way between the heel and metatarsal heads. Similar to FFS patterns, ground contact time is shorter than RFS, however without the large horizontal braking forces observed in RFS runners (Novacheck, 1998).

Electromyography (EMG) analysis of muscle activation shows lateral and medial gastrocnemius, soleus and peroneus longus are most active at initial contact, while tibialis anterior increases activity during the mid-swing phase (Cappellini, Ivanenko, Poppele, & Lacquaniti, 2006). At this point of the gait cycle, ground reaction forces, both horizontal and vertical, are experienced by the runner at varying magnitudes dependent upon a number of factors. These factors include body mass, running velocity, footwear and environmental factors such as running surface and foot-strike. Impact loading from vertical ground reaction forces have been associated with stress related, over-use injuries in athletes (Hreljac et al., 2000).

Impact peaks are passive forces, typically observed during the initial 10% of the stance phase. They result from the rapid deceleration of the foot when it meets the ground (Liu & Nigg, 2000) and reach magnitudes of ~1.5 times up to ~3.0 times body weight at slow (~2 $m.s^{-1}$) and faster velocities (~7 $m.s^{-1}$) respectively (Cavanagh & Lafortune, 1980; Hamill & Knutzen, 1995; Munro et al., 1987; Nilsson & Thorstensson, 1989). While impact peaks are passive, active peak ground reaction forces are due to the generation of force during eccentric activity of the lower limb musculature. At mid-stance, the body’s COM is at its lowest point. Active peaks typically reach magnitudes of ~2.5–2.8 times body weight and, like passive impact peaks, also increase with velocity (Cavanagh & Lafortune, 1980; Hamill & Knutzen, 1995; Munro et al., 1987; Nilsson et al., 1985; Grabowski & Kram, 2008).

It is well documented that impact and active ground reaction forces contribute to the risk of lower limb overuse injuries in athletes (Gaeta et al., 2006; Hreljac et al., 2000; Newman, Witchalls, Waddington, & Adams, 2013). Attenuating these forces for rehabilitative purposes, or during training, to increase physiological load without additional mechanical stress, may be performed using a number of methods such aquatic training harness or pneumatic suspended running, or specialised LBPP treadmills (Aaslund & Moe-Nilssen, 2008; Kilding, Scott, & Mullineaux, 2007).

2.3.1 Kinematics of running

Spatial and temporal variations of gait kinematics include foot strike, stride rate, stride length, flight time, and ground contact time. Interactions of these variables may vary throughout races (Nicol et al., 1991). This variation is observable in marathon runners where increases in stride frequency as stride length decreases with fatigue occurs over time (Nicol et al., 1991). Comparatively, as sprint athletes' transition through the sub phases of the shorter sprint events, acceleration through to maximum velocity phases for example, changes in running kinematics are also observed (Maćkała et al., 2015). For example, 100m sprint kinematics include faster hip angular velocity, and reduced braking forces upon ground contact during the maximal velocity phase as compared to the deceleration phase; and certainly when compared to the stride of a marathon runner (Maćkała et al., 2015; Nagahara et al., 2014). The distance runner has a longer moment arm due to less knee flexion during the recovery phase of the gait cycle, in order to enhance efficiency compared to the sprint athlete. The inverse relationship between angular hip velocity and braking forces, combined with a shorter moment arm during the recovery phase allows for a higher stride rate in the sprint athlete (Bushnell & Hunter, 2007).

Temporal measurements, such as ground contact time, differ between recreationally trained and highly trained runners. In a study of healthy recreational athletes, Grabowski et. al (2008), reported longer ground contact times with a concomitant increase of BWS during running (Grabowski & Kram, 2008). Conversely, in a study of elite and sub-elite Danish national runners, ground contact times reportedly decreased with an increase in BWS and an increase in running velocity (Raffalt, Hovgaard-Hansen, & Jensen, 2013). It is not clear however, if the increase in velocity or BWS alone, or the combination of both variables, was the cause for a decrease in ground contact times. These differences may be explained by the runners' level of experience. The highly trained athletes may have a better ability to apply force into the ground than their recreationally trained counterparts in response to changes in running velocity. Similarly, in a study of sixteen highly trained Australian Football league athletes Brughelli et al (2011) found that with increased running velocity ground contact times decrease and stride rate increased when running at 100% body weight. Further, when comparing stride length and flight time at changing velocities and BWS, trainability seems not to influence these characteristics. Stride length increased with both running velocity and BWS at 60, 70 and 80 %, in a recent study of active participants (Wen, 2007) in comparison to 0% BWS. Similarly, Raffalt et al (2013) also reported an increase in step length with increased BWS using highly trained, national runners. Furthermore, both flight and swing phases increased significantly with increased running velocity and increased BWS (Raffalt et al., 2013). These subtleties provide strong rationale to further investigate the kinematic effects of BWS running in highly trained athletes, across various sporting codes.

2.3.2 Kinetics of running

Some of the kinetic differences observed across the range of running modalities include, but are not limited to, vertical ground reaction forces, horizontal ground reaction forces and resultant peak forces (Helseth et al., 2008; Støren et al., 2011). At increasing running speeds, as observed in sprint running, kinetic variables such as horizontal and vertical impact forces are affected. In an observational study, Brughelli et al (2011) investigated the effect of increasing running velocity on vertical and horizontal forces using 16 semi-professional Australian Football League athletes. Authors found an increase in peak vertical forces from 40 to 60% of maximal running velocity, but no significant change from 60 to 100% of maximal running velocity. In contrast, horizontal force increased incrementally across all conditions from 40 through to 100% maximal running velocity. It is evident that alterations to running characteristics, such as velocity and body mass, affect kinetic forces acting upon the body (Brughelli et al., 2011).

2.4 Running and lower limb injuries

Improving the capability and capacity of athletes requires the planning and facilitation of high performance strength and conditioning programmes. Components of constructing elite level programs aimed at improving performance require a well-planned combination of volume, intensity, frequency, duration, density and recovery of mechanical, physiological and technical stimuli. Each variable is adjusted progressively in accordance with the training phase, and desired performance outcome required.

There is considerable research suggesting a rapid increase in training volume, with a concomitant increase in training intensity, and/or in conjunction with inadequate recovery, may contribute to lower limb overuse injuries in elite level athletes (Hreljac, 2004; Hreljac et al., 2000; Wen, 2007)**.** An overuse injury can be defined as the mechanical failure of a tissue or structure (Diehl, Best, & Kaeding, 2006) or as a cumulative trauma disorder, where tissue is damaged repetitively over time (Laker et al., 2015). Tissue tolerance is compromised when the stress stimulus applied to the musculoskeletal system is greater it's capable adaptive responses, accumulative micro-trauma is incurred at the sub injury-threshold level of the tissue (Ferber et al., 2009b). Bone and connective tissues are those primarily affected. When bone is exposed to abnormal repetitive stress, it commonly results in a spectrum of injuries ranging from asymptomatic osteopenia where osteoclast activity may overtake osteoblast activity under excessive strain (Winters et al., 2013) to reactive soft tissue (Newman et al., 2013). In the lower limb, this predominantly occurs in the tibia. The tarsal bones, metatarsals and fibula may also be effected, albeit less frequently than the tibia (Bresler, Mar, & Toman, 2012; Galbraith & Lavallee, 2009). When connective tissue is exposed to repetitive stress, overuse injuries such as tendinopathy may occur. In a clinical review Ferber et al (2009) report approximately 50% of the overuse injuries observed in 2000 recreational runners, occurred at the knee (Ferber et al., 2009b). These injuries include patella femoral pain, iliotibial band syndrome and patella tendonitis. With a further 40% of injuries observed below the knee, including Achilles tendonitis, plantar fasciitis and medial tibial stress syndrome (Taunton et al., 2002).

Stress related overuse injuries occur over time most commonly during repetitive athletic motions such as running and often without symptom until the later stages of injury (Diehl et al., 2006; Hreljac et al., 2000). Elite endurance runners spend hours every week completing up to and above 100km of running (Fricker et al., 2005). Team sport athletes may complete up to 4km in a single training session, including match simulated high intensity running. In addition, during weekly competition, team sport athletes can perform anywhere from 5.8km to 9.5km (Gabbett, 2010; White & MacFarlane, 2013) and up to 16.9km (Gray & Jenkins, 2010) of running. These accumulated weekly volumes of running are an example of training loads that may contribute to a high-risk environment for overuse, stress related injuries (Neely, 1998b; Newman et al., 2013). In addition to this, internal factors such as nutritional and hormonal variances and gait inefficiencies, often observed in athletes, may further contribute to the risk of lower limb stress injuries (Dallinga, Benjaminse, & Lemmink, 2012; Newman et al., 2013) (Table 2.1).

Table 2.1: Intrinsic and Extrinsic Risk Factors for Overuse Injury

Author	**Injury Mechanism**	**Overuse Injury**
(Neely, 1998a)	Ankle Dorsi Flexion <10o	Metatarsal fractures, plantar faciitis
	Femoral Anteversion	Patellofemoral Pain
	Hip Ext Rotations >65o	Tibial Stress Fracture (2x); Femoral stress Fracture (2.4x)
	Genu Varum/Valgum	Medial knee compartment pathology, osteoarthritis, tibial stress fractures
	Leg length discrepancies	Plantar fasciitis, lower back pain
	Foot architechture, Hyperpronation, navicular drop, > subtalar ROM	MTSS, tibia, fibular and tarsal stress fractures
(D. Wen, 2007)	Q-angle Previous Injury	Patellofemoral pain Injury reoccurrence, MTSS
(Messier & Pittala, 1988)	Pronation, ankle ROM, Q-angle, hamstring, flexibility, leg length	Shin splints, plantar fasciitis, iliotibial band syndrome
(Dallinga et al., 2012; Neely, 1998c; D. Y. Wen, 2007)	BMI & Body Fat %, Hormonal and Nutritional Status	Stress reaction, fractures
(Thomson, 2014)	Low tendon stiffness Leg stiffness (bilateral differences or > stiffness)	Tendinopathy Non-contact injury
(Reed Ferber, Alan Hreljac, & Karen D. Kendall, 2009a; Neely, 1998b; Newman et al., 2013; D. Y. Wen, 2007)	Exposure and intensity of activity (volume)	Overuse stress injury
(Neely, 1998c; D. Wen, 2007)	Environmental factors (surface, terrain, weather, time of day) Equipment (shoes)	PFP, MTSS, Plantar Fasciitis

Poor hip musculature stabilisation (Ferber et al., 2009b) previous injury (Zifchock, Davis, & Hamill, 2006) and impact forces (Hreljac, 2004) have also been associated with an increased risk of mechanical stress-induced overuse injury in athletes. Yang et al (2012) used a descriptive epidemiology study to track incidence of injury of 573 NCAA division one male and female athletes. Athletes from 14 sports codes and 16 teams participated. Authors reported 1317 injuries over the three-year period with 29.3% of all injuries being deemed overuse injuries. Female athletes presented with a higher proportion of total overuse injuries as compared to males (61.7% versus 38.3%). Although this study reports gender comparisons to be statistically insignificant (48.3% Female versus 51.7% Male) across ten gender-comparable sporting codes, other research suggests female gender to be a risk factor for lower limb overuse injury (Milner et al., 2006; Zifchock et al., 2006). The most common types of overuse injuries reported were general stress reactions (n = 103, 26.7%), followed by inflammation (n = 80, 20.7%) and tendonitis (n = 60, 15.5%). Overuse injuries were located in the lower extremity in 49% or 189 incidences. Similarly to Ferber et al (2009), the majority of these injuries were observed in the lower portion of the lower extremity (21.5% knee, 16.8% lower leg/ankle, 6.2% foot/toe) (Yang et al., 2012). Further, Bresler et al (2012) report the majority of running related injuries observed in athletes and treated in sports medicine clinics are stress related lower limb injuries. These include musculotendinous hamstring injury, Achilles tendinopathy and stress injury to the bone with tibial stress injuries being the most prominent (Bresler et al., 2012). Yang et al (2012) report 70.5 overuse injuries per 10000 exposures to training, practice games or competition occurred in athlete's representing women's field hockey, soccer, softball and volleyball. Lower limb overuse related injuries often result in the cessation of training and competition for the athlete for numerous weeks, and up to months. Comparing the biomechanical characteristics of team sport athletes who are involved in running, cutting and jumping actions; with individual running athletes, who experience repetitive loads, may provide insight into technical running variations that contribute to overuse injuries in the two groups. Assessing the gender differences during BWS running, with respect to both kinematic and tibial shock data, remains and important area of research for the elite athlete cohort also.

Athlete's performing high training and competition loads as previously mentioned, should also be mindful of the external factors known to contribute to stress related injury mechanism. These include footwear and running surface (Raasch & Hergan, 2006). Including strategies to improve these external factors in the daily training environment may assist to reduce mechanical stress for higher risk athletes, such as those returning from injury or with a previous injury history.

Medial tibial stress syndrome (MTSS) is one of the more common of the stress-induced lower limb injuries observed in athletic populations. MTSS has been believed to be caused as a result of a stress reaction of the fascia or underlying periostitis of the tibia (Gaeta et al., 2006). More recently, tibial stress injuries such as periosteal remodelling and the dysfunction of tibialis posterior and anterior, and soleus, have been implicated in MTSS (Beck, 1998; Galbraith & Lavallee, 2009; Newman et al., 2013). Medial tibial stress syndrome, similar to other mechanical stress-induced injuries, sits along the continuum from bone remodelling injuries such as osteopenia (Newman et al., 2013) through to stress fracture (Mizrahi et al., 2000; Wilder & Sethi, 2004). Typically occurring in cortical bones, such as the tibia, in runners the fracture or bone crack itself is propagated following persistent repetitive and accumulated micro trauma during loading (Pepper, Akuthota, & McCarty, 2006). As cortical bones are usually long bones, stress occurs most commonly through torsional forces. Torsion provides tension to the bone circumferentially, this combined with tensional forces, which occur along the convex aspect of the bone during bending, have been shown to disrupt the architecture of the bone. The effect of this is a de-bonding along the bones cement lines, leading to micro fractures (Pepper et al., 2006). Combined with the dysfunction of lower limb musculature and facia, continued loading, or in cases where inappropriate rehabilitation and training is implemented, stress fractures may occur. Similarly, ineffective treatment of stress reactions may lead to a stress fracture of the affected bone (Beck, 1998; Ferber et al., 2009b; Galbraith & Lavallee, 2009; Van Middelkoop, Kolkman, Van Ochten, Bierma-Zeinstra, & Koes, 2008). In a recent meta-analysis by Newman et al (2013) reviewed 10 journal articles with sufficient evidence of factors that were statistically significant with MTSS. A large number of participants across studies was reviewed (N=1924) and ranged from military and recreational groups to athletes representing tennis, volleyball, track and field, and soccer. Authors reported a number of qualities that may contribute to the risk of MTSS in runners. These risk factors include female biomechanical characteristics such as increased knee abduction at heel contact, decreased knee flexion, and increased peak internal rotation of the hip and increased femoral adduction as compared to that observed in the male athlete. The increase of body mass index, and therefore greater impact forces, lower limb anatomy, specifically femoral rotations in males, navicular drop and pronation also is considered risk factors for the incidence of MTSS in runners. Further, post tibialis tendinopathy has also been implicated in the complexity of MTSS (Maffulli, Wong, & Almekinders, 2003). No conclusive evidence for the most effective treatment of MTSS was reported. These factors highlight the need to assess the effects body weight unloading has during running, in both male and female athletes, on tibial shock and technique.

Tendinopathy is another of the more common forms of overuse injuries experienced by athletes. In the United States of America patellar tendinopathy represents 45% and 32% of injuries in volleyball and basketball players respectively (Crowell, Milner, Hamill, & Davis, 2010). While up to 29% of running based injuries involve Achilles tendinopathy (Crowell et al., 2010). Extrinsic factors such as excessive mechanical load, a rapid increase in progression, poor running technique and tasks that involve repetitive movement (Magnan, Bondi, Pierantoni, & Samaila, 2014; van Dijk, van Sterkenburg, Wiegerinck, Karlsson, & Maffulli, 2011) are some of the main contributors to tendinopathy. Intrinsic factors such age, body mass index and a high percentage of mechanically weak type III collagen (Kulig, Lederhaus, Reischl, Arya, & Bashford, 2009), also contribute to the pathology of tendinopathy.

There are a number of classifications of tendinopathy. Tendinosis is the term commonly used to indicate a degenerative non-inflammatory process with a disorganised collagen structure (van Dijk, van Sterkenburg, Wiegerinck, Karlsson, & Maffulli, 2011). Tendonitis, refers to the degeneration of the tendon accompanied with vascular disruption and inflammatory response (Timpka et al., 2014)**.** Para-tendonitis with and without tendinosis refers to the inflammation of the para-tendon, with or without associated degeneration (Maffulli et al., 2003). The physical components of most elite level sports include running, jumping or cutting manoeuvres either for training or competition, or both.

The tendon plays an active role in the musculotendinous unit in producing the forces that produce these movements. The Achilles tendon during running stores elastic energy, approximately 10% of the mechanical energy used each step (Brayne, Barnes, Heller, & Wheat, 2015), while during hopping tasks up to 40% of the mechanical energy required is produced by the Achilles tendon (Maffulli et al., 2003). The patella tendon is also a common site of tendinopathy for athletes. Commonly known as ‘jumper’s knee, it is evident in athletes pursuing sports where jumping and landing is a large component of the game (Wen, 2007). Proximal hamstring tendinopathy is also a common running related overuse injury, experienced by athletes performing primarily in the sagittal plane such as during distance running, hurdles and sprinting. Proximal hamstring tendinopathy is also common in athletes who perform changes of direction, concomitant with large ranges of hip flexion as experienced in sports such as soccer and field hockey (Knobloch, Yoon, & Vogt, 2008; Maffulli et al., 2003). The tibialis posterior tendon, which has been implicated in complex tibial stress related injuries in athletes (Newman et al., 2013) is also a common site of pain and dysfunction for athletes performing large volumes of impact based movements such as running.

The paradox of tendinopathy overuse injury is that reloading the injured tendon using eccentric strength training (Alfredson, Pietila, Jonsson, & Lorentzon, 1998; Crowell et al., 2010; Henriksen, Lund, Moe-Nilssen, Bliddal, & Danneskiod-Samsøe, 2004; Kulig et al., 2009) has been shown to improve function and symptom in athletes with chronic Achilles tendinopathy. Alfredson et al (1998) pioneered eccentric training for Achilles tendon rehabilitation in a study of 15 recreational athletes. The intervention consisted of three sets of 15 repetitions of heavy eccentric straight and bent leg heel lowers, performed twice daily, every day for 12 weeks. Concentric and eccentric calf strength improved, pain was reduced and athletes were able to return to pre-injury running following a 12-week intervention. Further, Saxena et al (2011) found LBPP treadmill based exercises to be effective for reloading and rehabilitating the Achilles following surgery. This research is discussed in more detail in the following paragraphs.

The recovery of stress related injuries is determined by the magnitude and distribution of mechanical load to the affected area. The magnitude and distribution of load can be controlled using stabilising aids, combined with progressive loading parameters, to stimulate bone remodelling. A recent study by Vijayakumar et al (Vijayakumar, Marks, Bremmer-Smith, Hardy, & Gardner, 2006) shows the force variation during tibial axial loading, at different stages of the gait cycle (heel strike, passive peak force; mid-stance, resultant peak force; and toe-off, active peak force) during a 16-week rehabilitation protocol using three dimensional bone modelling (Figure 2.2).

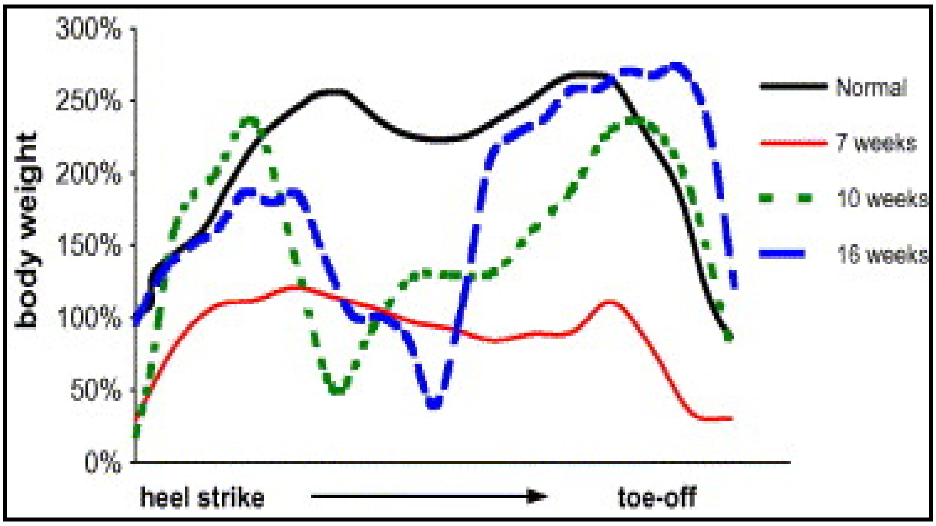

Figure 2.2 Tibial axial forces during stance phase of gait (Vijayakumar et al., 2006)

It is apparent that as bone remodelling progresses, resultant and peak forces increase. With this in mind it is important to understand lower limb loading rates during supported running to better understand the potential loading effect on the recovering tissue (Vijayakumar et al., 2006)

2.5 Body Weight Support Systems

Body weight support ambulation is a method of rehabilitation used for post-operative and post injury recovery interventions (Draovitch, Maschi, & Hettler, 2012; Saxena & Granot, 2011), gait retraining (Hesse et al., 1999), stroke rehabilitation (Aaslund et al., 2011; Chen, Patten, Kothari, & Zajac, 2005) and in patients with other orthopaedic and neurological conditions (Cyarto, Brown, Marshall, & Trost, 2008).

A number of systems are currently being used to support body weight, thereby reducing effective vertical and horizontal ground reaction forces incurred during normal over ground ambulation (Chang, Huang, Hamerski, & Kram, 2000). BWS systems currently employed include harness systems, aquatic training and LBPP treadmills. Currently the literature addressing the topic of BWS training, outlines biomechanical and physiological advantages and disadvantages for injured and/or disabled populations (Aaslund et al., 2011; Aaslund & Moe-Nilssen, 2008; Franz et al., 2007; Hesse et al., 1999). Highly trained athletes may benefit from such BWS systems for more concise rehabilitation and specialised training. Research addressing BWS training protocols in this cohort is currently limited to a small number of research articles and case study reports (Aaslund et al., 2011; Aaslund & Moe-

Nilssen, 2008; Franz et al., 2007; Hesse et al., 1999; Moore, Vandenakker-Albanese, & Hoffman, 2010; Patil et al., 2013; Saxena & Granot, 2011; Tenforde, Watanabe, Moreno, & Fredericson, 2012). Harness and pneumatic systems and aqueous training are some of the most common forms of BWS systems employed in both medical and elite training environments.

2.5.1 Harness body weight support

Harness systems comprise of polypropylene mesh over shoulder straps, lumbar or waist supports and in some, such as the Pneumex Inc. or Maine Anti-Gravity System harness, may also have individual leg straps (Millslagle, Levy, & Matack, 2006). Harness systems lift the participant at harness contact points to aid in reducing vertical ground reaction forces during treadmill ambulation. While pneumatic support systems such as the Pneu-Lift (Pneu-Lift, Pneumex, Sandpoint, ID, USA) are positioned directly above a standard treadmill and support the participant under the arms rather than around the pelvis (Thomas, DeVito, & Macaluso, 2007).

Extensive research in the areas of biomechanics (Aaslund & Moe-Nilssen, 2008), gait retraining and physical rehabilitation (Draovitch et al., 2012; Saxena & Granot, 2011) have suggested harness support systems to be valuable for patients commencing rehabilitation plans following injury, surgery, hemiparesis and stroke. Millslagle et al (2006) assessed running gait kinematics and kinetics in ten male runners using three different harness systems. The authors identified significant increases in stride length during harness supported running. A decrease in stride frequency and ground contact time was also reported with an increase in BWS. These kinematic changes may be reflective of an increase in flight time at a set running velocity. Shorter ground contact times and the reduction in stride frequency may be advantageous to the recovering athlete due to the reduced quantity of ground collisions and total volume of shock accumulation. In addition to these kinematic changes, there was a significant change in hip angle and increase in torso rotation suggesting many crucial kinematic changes occurred with increased BWS, albeit with a concomitant decrease in vertical ground reaction forces and vertical displacement of the COM (Millslagle et al., 2006). These kinematic changes in gait may be disadvantageous to the recovering athlete due to alterations in motor patterns which may later need re-training. Further, the possibility that kinetic forces are being re-distributed through other areas of the body such as the lumbar spine, is evident from the additional trunk rotation mentioned previously (Millslagle et al., 2006). Similarly, Aaslund et al (2007) investigated twenty-eight healthy participants walking at different velocities under varying supported conditions (floor and treadmill alone, with a harness, and with a harness and BWS). A decrease in vertical acceleration of the trunk was observed during harness support as compared to normal treadmill walking as might be expected when body mass is supported. Further, anterio-posterior, medio-lateral and vertical accelerations of the trunk decreased with 30% BWS during treadmill walking, and variability of vertical acceleration was reduced as compared to unsupported treadmill walking. These results suggest the variability of trunk movements may be reduced with the use of harnesses for support, which may also reduce vertical accelerations of the lower limb when ground contact is made (Aaslund et al., 2011). Conflicting observations were reported between Millslagle et al (2006) and Asslund et al (2007) with respect to stride frequency. Stride rate, reportedly decreased as BWS increased in the Millslagle (2006) study, but increased in Asslund (2007) study, as BWS increased. Participant health and physical capabilities, as well as study methodology and outcome measures may have contributed to this discrepancy.

In support of earlier research highlighting the benefit of harness systems for reducing impact forces during treadmill ambulation, Teunissen (2007) reported significant decreases in active peak vertical ground reaction forces experienced during mid-stance phase. In addition to this, there was a decrease in vertical impulse, peak horizontal breaking forces and peak horizontal propulsive forces incurred during treadmill running. These changes were statistically significant and occurred in linear fashion with the increase in BWS. Interestingly, decreases were non-proportionate with the magnitude of increased BWS. This suggests there may be a level of support at which kinetic changes deflect from the proportional linear relationship and become either more or less significant as BWS increases.

Harness systems may be effective for reducing body weight for rehabilitation purposes in patients following stroke (Aaslund et al., 2011; Danielsson & Sunnerhagen, 2000), injury (Draovitch et al., 2012; Saxena & Granot, 2011) and in elderly populations (Thomas et al., 2007) however have been described by research participants and anecdotally by patients as ‘uncomfortable’. Additionally, certain harnesses have also been shown to limit circulation in the lower extremities due to blood constriction around the groin (Millslagle et al., 2006; Thomas, De Vito, & Macaluso, 2007). Although these studies demonstrate a reduction in impact and ground forces during walking, harness systems may be limited to walking based ambulation rather than running. Access, cost and ease of use are limitations of both harness and pneumatic systems outside a clinical or laboratory setting, in which the patient or athlete would require permission and assistance to set up, calibrate and use the technology. Further, due to the circulatory and comfort restrictions, for the elite population these may be problematic from both a kinematic and compliance perspective.

2.5.2 Aquatic body weight support systems

An alternative approach to reducing body weight and associated forces is aquatic training. Advances in technology have led to the development of aquatic capable ergometers. Aquatic treadmills and aquatic erg-cycles have been introduced into training and rehabilitation centres globally (Killgore, 2012). More traditional aquatic training modalities such as deep-water running and shallow water running have been used for rehabilitation and aerobic conditioning for some time. Floatation devices or floatation belts are fitted around the participants' waist to aid floatation, as the participants perform gait specific drills in water two metres or deeper, where the participant's feet are unable to contact the floor of the pool. This eliminates the stance phase of running gait and therefore any vertical ground reaction forces acting on the lower limb (Kilding et al., 2007). Shallow water running as the name suggests, occurs in shallow water or on an aquatic treadmill, where the participant makes contact with the ground. The hydrostatic forces surrounding the immersed body reduce both the magnitude and velocity in which the lower limb travels towards and contacts the ground (Killgore, 2012).

Deep water running is performed in deep water whereby the athlete does not make contact with the ground at all. Often a floatation belt is used to assist buoyancy during deep water running. Consistent with harness systems, alterations to running gait during deep water running as compared to over-ground running have been observed. Hip and knee flexion and extension which occurs twice during over-ground running, once during the stance phase and once during the swing phase; occurs only once during deep water running (Killgore, 2012). This may be advantageous to athletes returning to sport following hip or knee injury. However it may also prove to be disadvantageous due to the changes in muscle recruitment patterns which are observable between the two modes of running (Kilding et al., 2007). The hip and knee flex simultaneously during overground running, however in water the hip flexes before the knee extends. This is due to the lack of ground contact and loading forces, which don't occur in the deep aquatic environment. The neuromuscular differences may be problematic for elite level athletes with respect to muscle recruitment potentially effecting running gait. The physiological stimulus may be adequate for promoting central and peripheral adaptations to maintain fitness levels during rehabilitation. As such, the use of deep water running for highly efficient technical runners may need to be closely monitored so as not to negatively affect technical proficiencies.

Shallow water running and underwater treadmills are also used to unload the body from potentially injurious ground forces. Aquatic treadmills and shallow water running accommodate the stance phase of running, while hydrostatic forces reduce the velocity of the athletes mass towards the ground, thereby reducing ground forces. The kinematic assessment of these modalities of aquatic running highlight variations in hip, knee and ankle angles which vary depending upon the depth of water the runner is immersed in (Killgore, 2012). Kilding et.al (2007) observed a decrease in hip flexion, an increase in hip extension, and reduction in knee extension during underwater treadmill running (Kilding et al., 2007)**.** Although kinematic changes are apparent, aquatic running has been shown to be effective at reducing impact, peak and active ground forces with the preservation and possible increase of muscle activation. For elite runners, this could be of benefit for rehabilitation purposes in order to maintain metabolic and neuromuscular condition, while reducing the risk of re-injury.

While aquatic training is beneficial for reducing impact forces and maintaining cardiovascular condition, both deep and shallow water training have biomechanical and/or physiological limitations. These limitations can affect training principles used to prescribe performance and rehabilitation programs, such as training specificity, intensity and volume; and may interfere in the intricacies required for performance in highly trained athletes.

2.5.3 Lower Body Positive Pressure Treadmills

In 2006, Alberto Salazar, a prominent marathon runner and coach, commissioned the first prototype of a LBPP treadmill, the Alter-G Trainer (Woodway Inc., CA) and pioneered its use to increase training volume facilitate over speed training in non-injured athletes. Purposes for the use of these antigravity treadmills include rehabilitation for injured, post-operative and diseased populations (Chen et al., 2005; Draovitch et al., 2012; Saxena & Granot, 2011).

The treadmill design and user interface allows progressive unloading and reloading, enabling the user to maintain running training, albeit modified, whilst unloading the injured, damaged or recovering tissue (Gojanovic, Cutti, Shultz, & Matheson, 2012; Thomas et al., 2007; Thomas, Vito, & Macaluso, 2007).

Developed from modified NASA space technology, an airtight canvass ‘chamber’ is fitted to a metal framework surrounding a specialised treadmill. Wearing a pair of neoprene shorts which have a ‘kayak like’ spray skirt attached, the participant is able to ‘zip in’ to the canvass outer creating an air tight chamber, which houses the treadmill belt and participants’ lower limbs. A uniaxial vertical force sensor is fitted within the treadmill base, beneath the belt. During calibration the treadmill ‘weighs’ the participant to determine 100% body weight. Once calibrated, a differential air pressure system is used to adjust the flow of air into the chamber and therefore the amount of assistance the participant receives. Air pressure is controlled through the treadmill’s console and is measured as percentage of body weight. The air pressure at the participants’ waist ‘lifts them up’, reducing the effect of gravity on body mass during ground contact, thus reducing effective body weight. It is important to note that in commercially available products neither body mass nor weight is identified on the computer interface. Therefore it should be considered that body weight measured by LBPP treadmills may not accurately reflect actual body weight as outlined in previous studies where BWS values were approximately 5% different to previously measured body mass (Gojanovic et al., 2012; Liebenberg et al., 2010).

Currently, much of the training and rehabilitation prescription being provided by clinicians and practitioners to the athlete is based on anecdotal practice. As such, evidence of the effects of BWS treadmill running, including best practice protocols for improving performance and rehabilitation practices are limited to handful of studies (Aaslund et al., 2011; Aaslund & Moe-Nilssen, 2008; Franz et al., 2007; Hesse et al., 1999; Moore et al., 2010; Patil et al., 2013; Saxena & Granot, 2011; Tenforde et al., 2012).

In a case study examining forces in-vivo, Patil et al. (2013) report a decrease in tibio-femoral force, down from 5.1 to 0.8 times body weight, at 100% and 25% body mass respectively (Patil et al., 2013). Further, Moore et al. (2010) reported a reduction in vertical loading during an aggressive return to running protocol for an elite ultra-marathon runner. The subject’s pain, following lumbar disc herniation, was alleviated, in addition to a reduction in vertical loading, allowing for a return to modified training in just a number of days (Moore et al., 2010). The benefit of this for the elite athlete is the maintenance of physiological conditioning during the rehabilitation process and the contribution this may have to performance. Tenforde et al exemplified this in a case study in 2012. An elite female NCAA track runner, suffering a pelvic stress injury, was progressively returned to pain free overground running after 8 weeks of BWS rehabilitation. The authors reported the progressive reloading to effectively accommodate bone remodelling and maintain physiological condition. The athlete went on to win national championships shortly after (Tenforde et al., 2012). The reduction in kinetic forces through joint structures has been established in these case studies.

The use of BWS locomotion as a rehabilitation strategy to reload tendons following surgery has also been researched. In a pilot study of 16 participants who had undergone Achilles tendon surgery, Saxena et al (2011) compared returned to running rehabilitation strategies following surgery using an experimental and control group. Elite participants were excluded from the study unfortunately, due to the effect of psychological and external motivators, which may have affected the results. The control group followed a traditional walk-to-run progression overground. The experimental group ran on a BWS treadmill and progressively reloaded effective body mass over time (Table 2.2) (Saxena & Granot, 2011).

Table 2.2. Postoperative Achilles Rehabilitation Program

EXPERIMENTAL GROUP PROGRAM OVERVIEW
Week 1: Initial evaluation; ankle joint mobilization; non-weight-bearing ankle strengthening with surgical tubing; cross-friction massage to incision and posterior ankle; seated calf stretch; introduction of single-limb proprioception; bilateral concentric heel raises; home instruction on strengthening with a towel; cryotherapy for 15 minutes at session end.
Week 2: same as above plus: soft tissue massage to calf muscle and posterior ankle tendons; mobilization of subtalar joint; gluteal strengthening; stationary bike without boot; modalities such as ultrasound and electrical stimulation if needed; unilateral concentric strengthening at 60% BW on AG treadmill; walking on AG treadmill at 40% of BW for 10 minutes.
Week 3: same as above plus: standing hip Thera-band (The Hygienic Corp, Akron, OH) 4-way; Swiss ball exercises progressed from above; side steps with tubing; Pilates reformer (leg press, calf raise, hamstring arcs, and circles); ankle proprioceptive-neuromuscular facilitation (PNF); mobilization of great toe; progression to walking at 70% BW for 10 minutes; increase in home program strengthening.
Week 4: same as above with progression of single-limb strengthening from 70% to 90% BW on AG treadmill (starting with 3 × 10 reps, progress to 5 × 25); walking up to 2 miles at 70% BW. Depending on patient's individual progress: dynamic balance; step downs; calf eccentrics if pain free; lunges; BOSU (both sides up) squats; single-leg heel raises and leg press; ankle PNF.
Week 5: same as above with progression on AG to walking 75% to 85% BW for 2 miles; increase in strengthening concentrically full BW with surgical limb.
Week 6: same as above plus begin walk/run program (alternating 2 minutes of walking with 2 minutes of running) with progress on AG from 75% to 85% BW for 10 minutes. Patients were typically discharged at this time with their home program of strengthening, proprioception, stretching, and cryotherapy to be maintained until they were able to return to their full activity level.

Criteria for running overground were a pain free 10 minute run on the AlterG treadmill at 85% body weight. The experimental group returned to overground running following 18 weeks, and the control group following 20 weeks. Authors reported a non-significant difference in the duration of time it took participants to return to running (Saxena & Granot, 2011). In the elite population however, the ability to continue to stimulate the neuromuscular and physiological systems of running for an additional two week period maybe beneficial for the athlete’s return to play status and may contribute to a more effective and efficient return to sport strategy (Saxena & Granot, 2011).

2.6 Muscle activation during body weight supported treadmill running

To date, kinematic analysis of running gait using LBPP treadmills for highly trained athletes is still limited. However, kinetic variables and EMG have been addressed in untrained and trained populations. A pilot study conducted on patients following Achilles surgery identified a faster return to running after using the Alter-G trainer, compared to a control group who underwent traditional non-running based rehabilitation protocols (Saxena & Granot, 2011). Patients were cleared to run outside once they were able to complete 10 minutes of pain free running at 85% of body weight. These patients were able to return to overground running approximately two weeks earlier than their control counterparts (Saxena & Granot, 2011).

In an article by Draovitch et al. (2012), authors aimed to provide periodised weight bearing treatment solutions for return to sport following hip injury (table 2.3). Authors discussed a number of the possible rehabilitation methods for return to sport including the use of a LBPP treadmill. Ground reaction forces were reduced, muscle recruitment patterns measured using EMG, were maintained albeit at a reduced amplitude and the athlete returned to an overground running program after week 16 (Draovitch et al., 2012).

Table 2.3: Postoperative running program (2-3 x p/w)

Post Op Week	Time (min)	Speed (mph)	Body Weight %
13	Run 1/walk 1 (intervals x5)	6.0 (or 5 K race pace)	60%
14	Run 2/walk 1 (intervals x4)	6.5 (or 5 K race pace)	70%
15	Run 3/walk 1 (intervals x5)	6.5 (or 5 K race pace)	80%
16	Run 4/walk 1 (intervals x 4)	6.5 (or 5 K race pace)	85%

The benefits of maintaining normal muscle activation sequences, particularly for highly trained and efficient athletes are outlined and supported by Colby et al. (1999). The authors observed similar EMG activity in the quadriceps, hamstrings and gastrocnemius across BWS conditions ranging from 0%, to 40%, with the exception being a significant reduction in quadriceps EMG at 40% BWS as compared to full body weight loading. Liebenberg et al. (2010) reported a statistically significant linear decrease in muscle activity with increased BWS. This study also looked at the effect running velocity would have on muscle activity at varying rates of BWS. The authors reported an increase of muscle activity with the increase of velocity of the quadriceps and lower limb musculature. However, EMG activity still decreased with increased BWS across all velocities. The magnitude of muscle activity in the lower limb decreases as BWS increases as demonstrated in previous research (Liebenberg et al., 2010; Mercer et al., 2013). What is interesting to note, is that unlike aquatic running, the muscle recruitment pattern across all levels of weight support and velocities remained consistent using the LBPP treadmill. It is important when programming for elite athletes, to consider any potential interference to highly refined motor patterns, so as not to negatively impact technique and performance. The use of LBPP treadmills for running may mitigate this risk in comparison to aquatic running while unloading lower limb structures (Liebenberg et al., 2010).

Raffalt et al. (2013) investigated the effects and interaction of velocity and BWS during treadmill running. Impact forces, as well as kinematic and physiological variables, were assessed in a cohort of 12 male elite and sub-elite Danish runners. Mean and peak vertical ground reaction forces were measured, in conjunction with kinematic metrics such as ground contact times and flight time. Both mean and peak vertical ground reaction forces decreased as support increased from 75, 50, 25 and 0 % of body mass; this is consistent with Grabowski et al. (2008) study of ten recreational runners (Grabowski & Kram, 2008). Raffalt et al. (2013) also saw an increase in peak and mean ground reaction forces with increased velocities, both during steady state running and at high-speed velocities. Interestingly, at faster running velocities, the increase of BWS resulted in a larger relative decrease of peak and mean ground reaction forces than recorded at slower running speeds (Raffalt et al., 2013). These findings were similar to a study of ten elite and sub-elite Australian Football League athletes where researchers, using DorsaVi Accelerometery (measured in arbitrary units using The ViPerform from DorsaVi Ltd., Australia) also found an expected increase in lower limb loading as a result of decreasing BWS and increasing running velocity. However, the authors reported that body weight had a greater effect on lower limb loading, than did velocity with the higher running speeds showing an exponentially greater attenuation of lower limb loading at the same magnitude of BWS than slower running speeds (Raper et al., 2014).

2.7 Conclusion

The varying BWS systems available provide athletes with a number of options for rehabilitation return to sport and training purposes. While effective at reducing lower limb impact forces, harness systems seem to be better utilised for supported walking protocols in older or clinical populations. These and pneumatic devices may be more problematic for highly trained populations due to the effects on kinematic gait characteristics, comfort and circulatory restrictions and accessibility of equipment. Aquatic training may offer a more effective alternative to harness support systems for rehabilitation purposes, and to maintain aerobic condition in the recovering athlete. Practitioners should be mindful of the changes in muscle recruitment strategies and gait kinetics in deep water, and the effect of hydrostatic forces on muscle activation in the highly trained technical runner.

Further investigation on the effect BWS running using LBPP treadmills has on gait kinematics and tibial shock in the highly trained athletic population is warranted given the scarcity of literature in this cohort. Although it is well documented that vertical ground reaction forces can be reduced with an increase in BWS, further investigation is required to determine the effects of BWS on tibial shock using accelerometery in the elite athletic population using antigravity treadmills such as the AlterG. Therefore, the aim of this thesis is to investigate the effects of unweighting using LBPP technology with highly trained populations, to determine the effects this has on tibial shock, and gait kinematics.

Chapter 3: Methodology

The following chapter outlines the study design, testing protocols, participant information and methods used to determine the effect of body weight unloading on tibial shock and gait kinematics. Additionally, the processing and statistical analysis of data collected from this study will be addressed in this chapter.

3.1 Experimental Approach to the Problem

Twenty elite, international level athletes who represented New Zealand in a variety of team and individual Olympic sports volunteered to participate in this study. A repeated-measures design, using quantitative measures, was performed to determine changes in tibial accelerometery and sagittal plane gait kinematics at five different body weight conditions using the Alter-G Trainer treadmill (Woodway, CA). Participants participated in two familiarisation sessions prior to data collection. The test-retest reliability of the protocols was determined using four participants who repeated the testing two weeks apart.

3.2 Participants

Twenty highly trained athletes (male n=9; female n=11) from a variety of field based team sports (field hockey, rugby and soccer), track and field, distance running, and aquatic sports (swimming and kayaking) were recruited to participate in this study**.** Participants had all represented their country in international competition in their chosen disciplines in the previous twelve months. Participants from team and aquatic sports, for ease of classification were termed ‘team’ athletes; while the athletes who represented track running and triathlon were termed ‘individual’ athletes:

The inclusion criteria for the observational study required the following of participants:

i) Represented one of the following sports at a minimum of international level within the previous 12 months: hockey, soccer, rugby (sevens, league or union), triathlon, athletics (running), swimming, rowing or kayaking.

ii) Aged 18 to 35 years of age on the day of testing

iii) Injury free as determined by the representative sports’ medical lead. An injury was determined as a musculoskeletal abnormality that inhibited the athlete from full training in the previous two weeks prior to testing. This also included any medical illness.

iv) Involved in a full training load for the previous two weeks prior to testing

v) Participated in running based training, either within their sport or for additional conditioning

3.3 Study Design

3.3.1 Familiarisation

All participants completed two familiarisation sessions on a specialised, BWS motorised treadmill (AlterG Incorporated, Fremont, California) prior to the data collection (Hopkins, Schabort, & Hawley, 2001). The purpose of the familiarisation sessions was to acclimate participants to running on the BWS treadmill thereby minimising any possible learning effect, which may have influenced results. Each familiarisation session consisted of running at 100%, 90%, 80%, 70% and 60% BW for 4 minutes per condition, randomly selected by the tester, at a velocity of 12kph (3.33ms^{-1}) with 2 minutes' rest between conditions.

3.3.2 Reliability

Two weeks following data collection four participants repeated the testing protocol. This test-retest reliability study consisted of running at 100%, 90%, 80%, 70% and 60% BW for four minutes per condition at a velocity of 12kph (3.33ms^{-1}) with 2 minutes' rest between conditions as per familiarisation. The final 60 seconds of data collection at each speed was used for analysis.

3.3.3 Repeated-measures study

A repeated measures design study, using quantitative methodologies, was utilised to investigate changes in gait kinematics and tibial shock during four different BWS conditions. A full body weight (100%) condition was used as the control.

All participants gave written, informed, voluntary consent (Appendix 2). The AUT University Ethics committee approved this study (approval number: 13/198).

3.4 Testing procedures

Participants reported to the High Performance Sport Performance Rehabilitation clinic to undertake the assessment protocols described below for each formal testing and familiarisation occasion.

3.4.1 Familiarisation and velocity assessment

Submaximal running velocity was assessed during familiarisation to ensure the pre-determined velocity of $3.33m.s^{-1}$, while running on the Alter G, was appropriate for this cohort. A doubled blinded assessment to determine the appropriateness of the intended submaximal running velocity of $3.33m.s^{-1}$ was performed during familiarisation. This was retested on the first day of data collection during the participants' warm up. Participants warmed up for 6 minutes at 100% body mass, at self-selected velocities. Following the warm up, a research assistant took control in adjusting the treadmill velocity. Treadmill velocity was hidden from both the researcher and participants throughout, to minimise any bias. The research assistant increased the treadmill velocity from 10kph by 0.2kph every 20sec. The participant instructed 'stop' when they felt the treadmill reached a velocity that could be sustained for 30 minutes at a submaximal intensity**.** All participants selected a velocity between 11.8 and 12.2 kph. It has been documented that non-preferred running speeds may require excessive muscle activation in the major muscle groups of the leg, thus affecting metabolic output and biomechanical efficiency (Cappellini et al., 2006; Prilutsky & Gregor, 2001). This velocity assessment was performed to ensure $3.33ms^{-1}$ (12kph) was an appropriate submaximal velocity for this cohort.

3.4.2 Reliability study

Following initial data collection, four participants completed a test-retest study to assess the reliability of the testing protocols. Protocols were as outlined in section 3.4.3, with the exception that only the final 60 seconds of video and accelerometer data was collected for reliability analysis.

3.4.3 Repeated measures study

Prior to data collection participants were measured for body mass, fitted with joint markers, and a wireless accelerometer as shown in Figure 3.1.

Figure 3.1 Marker placement

Tibial acceleration:

Participants were fitted with a three-dimensional (3D) wireless accelerometer (IMeasureU Ltd, Auckland), which was securely attached to the anterior-medial aspect of the left tibia based on protocols previously described by Crowell et al. (2010). Figure 3.2 shows the accelerometer placement.

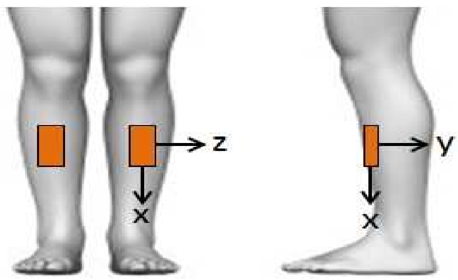

Figure 3.2 Accelerometer placement

Gait kinematics:

Small reflective light markers were attached to the lateral aspects of the left knee, ankle and foot. Bony landmarks of the left fibula head, lateral malleolus and fifth metatarsal were palpated, according to the International Society for the Advancement of Kinanthropometry (ISAK) kinanthropometry protocols (Stewart, Lower Hutt, New Zealand, 2011) for accurate and reliable marker placement. A high-speed camera (Casio EX-F1) capturing at 300 Hz was mounted 1.5m to the left side of the treadmill to capture temporal gait measures of the left limb.

Each participant ran on a specialised BWS motorised treadmill Alter G as shown in Figure 3.3. Participants began with a 6-minute warm-up with no BWS at a self-selected velocity on the AlterG treadmill. A 2-minute rest period was given to make any final adjustments to the set up prior to commencing data collection. Velocity was set at 3.33m.s^{-1} for data collection.

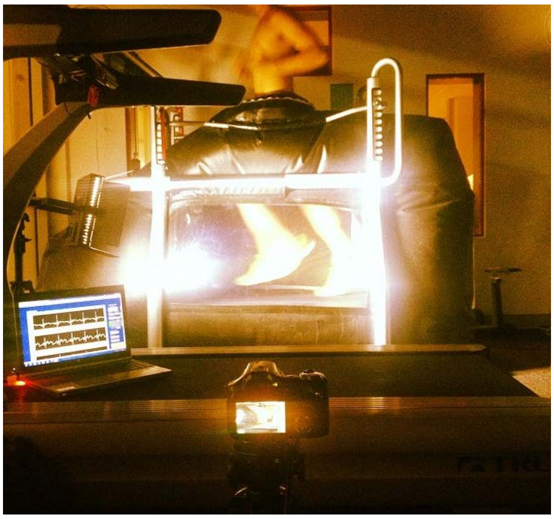

Figure 3.3. Set up for the observational study

Participants were then assessed in each of the following conditions, in a randomized order, for 6 minutes

100% of Body Weight (control)

90% of Body Weight (10% BWS)

80% of Body Weight (20% BWS)

70% of Body Weight (30% BWS)

60% of Body Weight (40% BWS)

Two minutes between each condition was provided as passive rest to minimise the effect of fatigue. During the final 3 minutes, sagittal plane video footage was recorded and 3D tibial acceleration collected for analysis. The assessment protocols lasted for approximately 60 minutes on each occasion, which included mark up and set up.

Kinematic gait variables were determined using high speed, sagittal plane video analysis. Foot strike was assessed and reported as heel strike (RFS) or fore/mid-foot strike (FFS).

Forefoot and midfoot were combined as one strike pattern (FFS) due to the small number of true forefoot strikes (n = 1). RFS was recorded when the heel of the shoe made contact with the treadmill belt prior to any other part of the foot. FFS was recorded when the foot struck the treadmill belt from the lateral or distal aspects of the foot arch and/or, the head or more distal lengths of the first metatarsal. Stride rate was determined using video analysis and reported as stride per minute (SPM).

Ankle angles were measured with the long axis of the foot and fibula aligned. Where these two segments were perpendicular the angle was defined as 0^o (Clarkson, 2000). Ankle angles were recorded at initial contact (IC), mid-stance (MS) and terminal stance (TS). Mid-stance was determined as the point where the knee reached maximal flexion and ankle reached maximal dorsiflexion during the force absorption phase. The contralateral limb was positioned lateral to the support limb, indicating full support of body mass over the support leg, just prior to the force propulsion indicated by the commencement of knee flexion and ankle plantar flexion and pronation. (Loudon, 2008; Shultz, 2005). Loaded total range of movement (tROM) was defined as the range of movement through the ankle while under body mass load. The tROM was established as the sum of range of movement from IC to MS (eccentric dorsi-flexion) and MS to TS (concentric plantar flexion).

Tibial shock was derived from resultant peak tibial axial accelerations, measured using the accelerometer (IMeasureU Limited, Auckland NZ) and calculated using customised MatLab® software (National Instruments Corporation, Austin TX). Accelerometery measurement may be described in terms of a mass-spring model of which in underpinned by Newton's 2^{nd} law of motion (F=ma) and Hooke's law (F=kx) principles (Kavanagh & Menz, 2008).

3.5 Data Processing

Acceleration data was acquired using a 16 bit, 1000Hz frequency accelerometer and was processed using customized MatLab® Software (National Instruments Corporation, Austin TX), extracting peak vertical acceleration and loading rate as a resultant peak tibial acceleration (Olin & Gutierrez, 2013). Resultant peak tibial accelerations were determined from touch down shock (time to peak shock), peak shock (peak accelerometery reading during stance) and average shock (average tibial shock over the stance phase) as previously reported (Olin & Gutierrez, 2013). The mean of the peak resultant accelerations during each condition for each subject was recorded and converted to units of gravity (g) for tibial shock. Ankle angle (measured at initial contact, mid stance and toe off), stride frequency (cadence measured in strides per minute) and foot strike pattern were measured using Kinovea video analysis software to watch video recording at 300 frames s^{-1} (www.kinovea.org). Foot strike was classified as either RFS, where the heel of the participant foot made clear contact with the treadmill belt prior to the mid or fore foot; or FFS, where the mid or fore-foot, which will be referenced as forward from the vertical of the anterior aspect of the left lateral malleolus, contacted the treadmill belt first.

3.6 Statistical analysis

All data were visually inspected and outliers >3 standard deviations from the mean were removed. Data were averaged across all strides for each BWS condition. Uncertainty was expressed as 90% confidence limits. Inference threshold values for assessing smallest worthwhile change were set at -0.2 and 0.20 (0.2 small).

Reliability

Reliability data were derived during the final 60 seconds of each running condition, including the control condition (100% body weight). Statistics for each consecutive pair of measurements assessed test-retest reliability using a customised spreadsheet to assess reliability (Hopkins, 2011). Reliability was expressed as the intra-class correlation coefficient (ICC) and standard error of measurement expressed as coefficient of variation (CV) was used to describe the variability, or error within the testing measurement. Log transformation was appropriate to reduce bias arising from non-uniformity of error (Hopkins, Marshall, Batterham, & Hanin, 2009).

The level of acceptance for reliability was set at ICC > 0.70 (Hopkins et al., 2001). An ICC close to 1.00 indicates a 'perfect' agreement with minimal variation (Atkinson & Nevill, 1998), an ICC > 0.8 represents an 'excellent' agreement, whereas ICC ranges within 0.7 – 0.8 are considered 'average' (Hopkins & Manly, 1989; Morrow & Jackson, 1993; Shrout & Fleiss, 1979). The CV was set at < 10%. A typical error >10% is considered a 'large' variation, while a typical error < 10% is considered to be a small variation (Cormack, Newton, McGuigan, & Doyle, 2008). Overall interpretation of reliability was deemed 'good' when the ICC was > 0.70 and the CV was < 10%. Reliability was described as 'moderate' when ICC was < 0.70 or CV was > 10%. Reliability was deemed 'poor' when ICC < 0.70 and CV > 10% based on methods reported previously (Bradshaw, Hume, Calton, & Aisbett, 2010).

Repeated measure study

Repeated measures post-only cross-over analysis was conducted with a customised published spreadsheet (Hopkins, 2006) to compare the effects of conditions across subjects. Mean and standard deviations were calculated to describe the centrality and spread of the data.

Changes in means

The changes in means between conditions of tibial acceleration and kinematic measures were determined via log transformation with confidence limits set at 90% (Hopkins, 2006) The smallest worthwhile change for outcomes measures was set at 0.2 standard deviations and the change in means were expressed as standardised Cohen's effect sizes (ES).

Effect Size

Effect size was determined for all conditions against the control condition for tibial shock, ankle kinematics and ankle tROM. Effect size for SPM was also calculated as (Mean SPM Condition - Mean SPM Control)/ Standard Deviation Control for the calculation of effect of each BWS condition as compared to the control condition. Cohen's thresholds for standardised differences in means were described as >0.91 (extremely large), 0.71-0.90 (very large), 0.51-0.70 (large), 0.31-0.50 (moderate), 0.11-0.30 (small), 0.0-0.10 (trivial) (Hopkins et al., 2009).

Magnitude-based inferences

Magnitude-based inferences were derived by calculating the probabilities of change for each BWS condition, and a qualitative term was assigned to the percentage based probability scale of change as follows: <0.5%, most unlikely; 0.5-5%, very unlikely; 5-25%, unlikely; 25-75%, possibly; 75-95%, likely; 95-99.5%, very likely; >99.5%, almost certainly.

Correlational Analysis

Pearson's product moment correlation coefficients were used to calculate the relationships between the conditions and variables on each occasion. The magnitude of correlations was described as trivial (0-0.1), small (0.1-0.3), moderate (0.3-0.5), large (0.5-0.7), very large (0.7-0.9) and nearly perfect (0.9-1.0) (Hopkins (2002).

Chapter 4: Results

The following chapter reports the results obtained from this study. Results are reported as means, standard deviations, and change in means, ES and Pearson product moment correlation correlations. These results are documented for tibial shock and gait kinematic measures. Gait kinematic measures reported are SPM, ankle angles (IC, MS, TS) and foot strike (RFS, FFS). The ICC's and CV's from the reliability study are also documented towards the end of this chapter.

Twenty participants completed this study (table 4.1). Eleven of these were female participants, age 21.9 ± 2.9 and body mass 65.5 ± 9.7, and represented team sports (n=6) and individual pursuits (n=5). The nine male participants, age 22.8 ± 4.1 and body mass 78.9 ± 7.1, also represented both team sports (n = 8), and individual pursuits (n = 1).

Table 4.1: Participant Demographics

	Age	**Body Mass**	**Team**	**Individual**
F	21.9 ± 2.9	65.5 ± 9.7	6	5
M	22.8 ± 4.1	78.9 ± 7.1	8	1

F = Female; M = Male; Age and Body Mass = Mean ± Standard Deviation;

Team = number of team sport athletes; Individual = number of individual sport athletes

4.1 Tibial Shock

The means and standard deviations for tibial shock across all BWS conditions are shown in figure 4.1. At 60% and 70% conditions, the changes in means were 0.71 and 0.59 respectively. The change in mean for 80% condition was 0.35 and for 90% condition, 0.09.

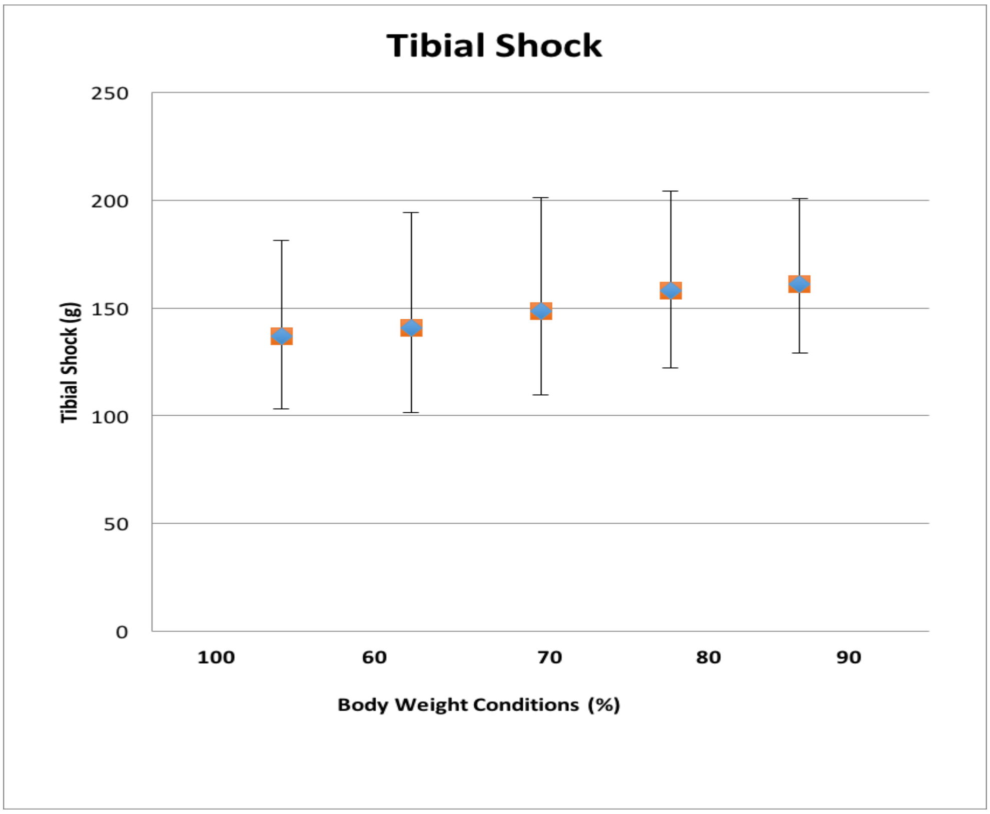

Figure 4.1: Means and Standard Deviations of group tibial shock data across all conditions

Table 4.2 shows that there was a very large ES for 60% condition. The ES for 70% condition was large. A moderate ES for 80% condition was observed while only a trivial change in mean was observed for the condition at 10% BWS.

Table 4.2: Effect Size for Tibial Shock – Group Analyses

	60%	**70%**	**80%**	**90%**
Standardised change in mean	0.71	0.59	0.35	0.09
Effect size	Very large	Large	Moderate	Trivial

Data were adjusted for athlete classification and also for gender (Table 4.3) to determine any differences between groups. Trivial effects were observed for athlete classification across all conditions, no further covariate analysis was performed. Small effects were observed for gender at 70 and 80%, conditions, while trivial effects were seen for conditions 60 and 90% conditions as outlined below. As such no further gender covariate analyses were performed.

Table 4.3: Gender Analysis for Change in Mean and Effect Size for Tibial Shock.

	60%	70%	80%	90%
Standardised change in mean	0.06	0.26	0.26	0.10
Effect size	Trivial	Small	Small	Trivial
Change in mean (%)	1.4	5.9	5.8	2.4

Appendix 6 outlines the magnitude of mechanistic inferences with standardised Cohen's Unit percentages and qualitative descriptors. At the 60% condition there was a high probability (100%, most likely) of a positive difference in tibial shock as determined by accelerometery, when compared to the control. For 70% condition, when compared to the control, the probability of tibial shock being substantially different was very likely (99%). At 80% condition, a substantial positive difference was likely (89%). Only at 90% condition was it unlikely (6%) that there was a substantial difference, and likely (94%) that the difference observed was trivial when compared to the control condition of 100% body weight. No negative effects were reported across all conditions.

Table 4.4 shows the Pearson correlations between tibial shock measures that ranged from large to nearly perfect (r=0.79 – 0.95).

Table 4.4: Pearson product moment correlations for tibial shock

	60%	70%	80%	90%
R	0.79	0.91	0.92	0.95

A summary of results for the repeated measures study across all conditions is outlined in appendix 7.

4.2 Kinematics

Stride rate

Table 4.4 shows an extremely large ES for 60% condition; a very large ES for 70% condition and a large ES for 80% condition. A small ES was observed for 90% condition.

Table 4.5 Effect of body weight support on stride rate (Strides Per Minute)

	60%	70%	80%	90%	100%
Mean	77	79	80	81	82
SD	3.5	3.4	3.5	3.5	3.7
ES v 100	1.29	0.86	0.58	0.28	

At 60% condition mean stride rate was 77 SPM. At 70% condition mean SPM was 79, this increased linearly as BWS decreased. At 100% control condition, mean SPM was 82. Figure 4.3 highlights these changes in stride rates across all conditions.

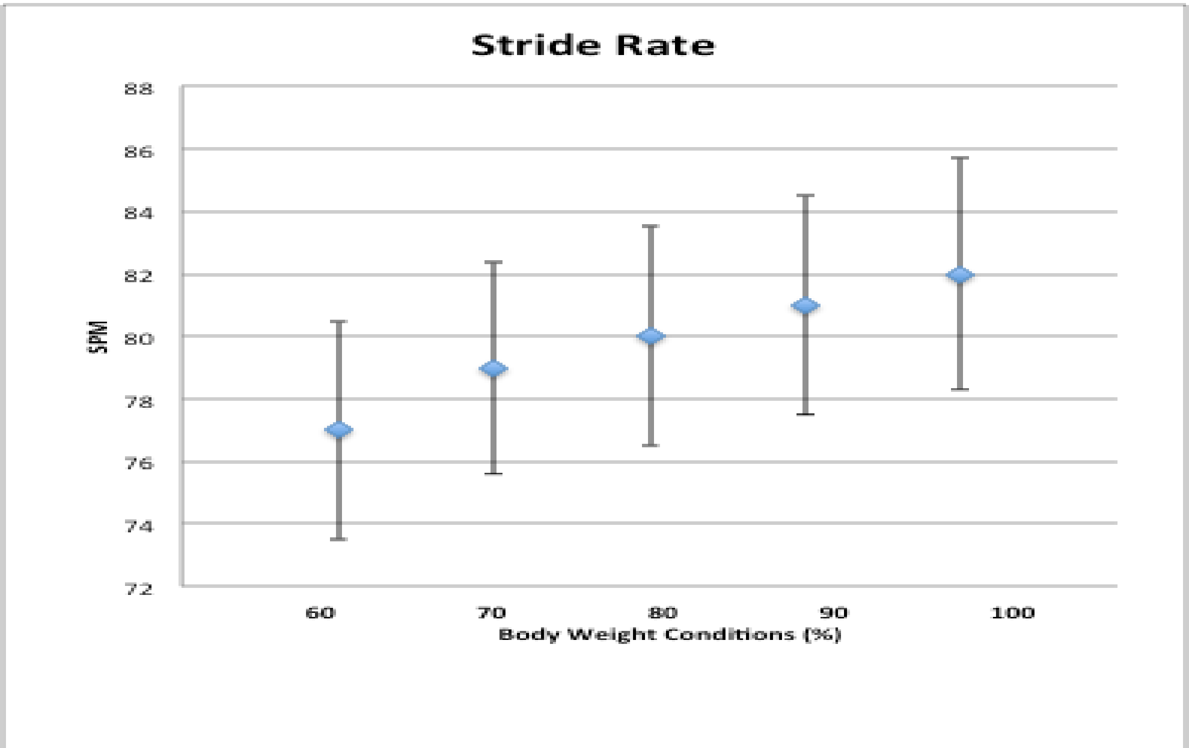

Figure 4.3: Changes in mean strides per minute across all conditions

Ankle kinematics

There was a moderate change in mean between control and 60% condition for both IC and MS ankle kinematics. Small changes were observed for all other conditions during IC, for 70 and 80% conditions during MS and for 70% condition at TS. All other changes were deemed trivial. Figure 4.4 shows the mean ankle angles at IC for the different conditions.

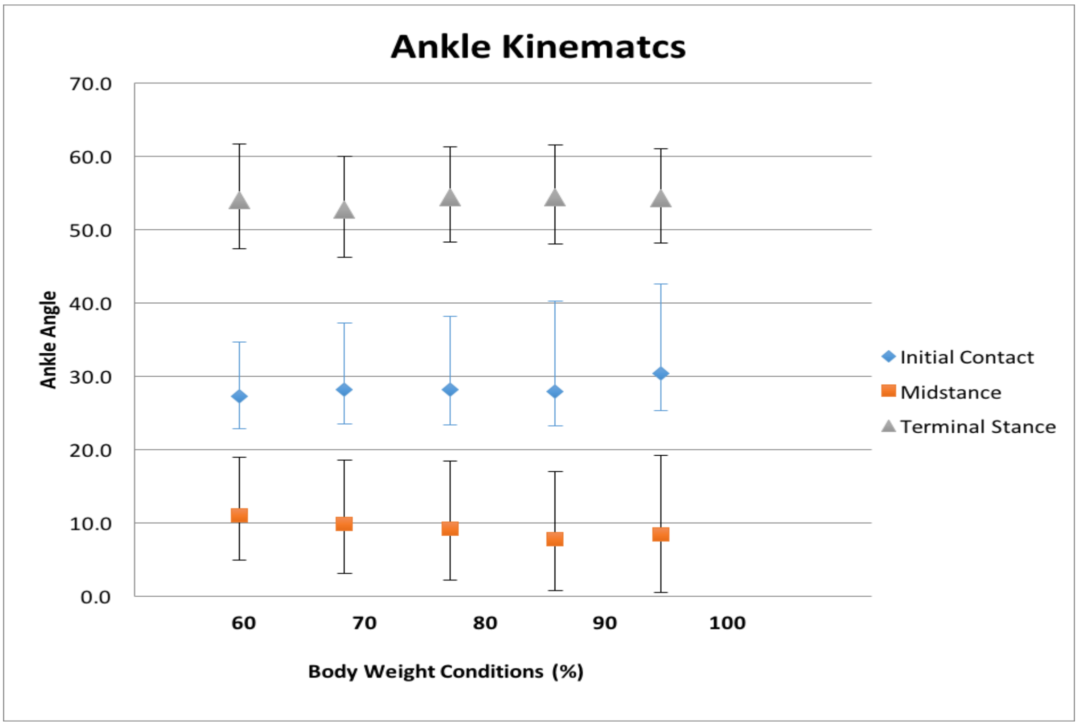

Figure 4.4: Mean and Standard Deviation ankle angles across all conditions

At initial IC a moderate ES was observed for 60% condition, with a standardised change in mean of 0.35. Small ES were reported for 70, 80 and 90% conditions. Change in means for these conditions were 0.23, 0.24 and 0.27 respectively. Small and trivial ES were observed for MS across 70, 80 and 90% conditions for MS measures. A moderate ES (0.31) was observed for 60% condition for MS. The ES for TS at 70% condition was small (0.24). Trivial ES were seen across the remaining BWS conditions. Both 60 and 70% conditions for tROM, had large changes in means. The standardised change in mean tROM for 60% condition was 0.51, while the standardised change in mean was 0.79 for 70% condition. A small (0.24) and trivial (0.00) ES was observed for 80 and 90% conditions respectively.

The percentage change in means and standardised change in means for sagittal plane ankle angle at IC, MS and TS, as well as tROM, for all BWS conditions are outlined in table 4.6.

Table 4.6: Change in Means and Effect Size of ankle kinematics

	60%	70%	80%	90%
Initial Contact				
Standardised change in mean	0.35	0.23	0.24	0.27
Effect size	Moderate	Small	Small	Small
Change in mean (%)	11.5	7.6	7.9	8.8
Midstance				
Standardised change in mean	0.31	0.20	0.14	0.04
Effect size	Moderate	Small	Small	Trivial
Change in mean (%)	25.2	17.1	11.8	4.3
Terminal Stance				
Standardised change in mean	0.03	0.24	0.01	0.02
Effect size	Trivial	Small	Trivial	Trivial
Change in mean (%)	0.4	3.0	0.2	0.2
tROM				
Standardised change in mean	0.51	0.79	0.24	0.00
Effect size	Large	Very Large	Small	Trivial
Change in mean (%)	6.0	9.3	2.7	0.1

Table 4.7 shows the Pearson product moment correlations for kinematic ankle measures as compared to the same measure's related control condition. For tROM, a moderate correlation was observed for 60% condition ($r = 0.31$), a large correlation was observed for 70% ($r = 0.68$) and very large correlations ($r = 0.80$) were seen for both 80 and 90% conditions with respect to tROM control. The correlations for 90% condition were nearly perfect for IC, MS and TS with respect to IC, MS and TS controls ($r = 0.99$, 0.96 and 0.96 respectively). For tROM, IC, MS and TS at 60% condition, very large correlation was observed against their respective controls as seen in table 4.7 ($r = 0.81$ to 0.88). At 70% condition correlations were large ($r = 0.62$) for IC, very large for MS ($r = 0.89$) and nearly perfect for TS ($r = 0.92$). Both

IC and TS for 80% condition were nearly perfect, while a large correlation was reported for MS (r = 0.87).

Table 4.7: Pearson product moment correlations for ankle kinematics

r values	60%	70%	80%	90%
tROM	0.31	0.68	0.80	0.80
Initial Contact	0.81	0.62	0.97	0.99
Midstance	0.88	0.89	0.87	0.96
Terminal Stance	0.87	0.92	0.98	0.96

The correlation matrix (Appendix 9) outlines a number of further significant correlations between measured variables. Moderate and large correlations were observed for stride rate and ankle kinematic measures with moderate correlations observed between SPM at 80% condition and IC at 70% (r = 0.49), the control SPM and MS at 60% condition (r =0.47) and MS at 70% condition (r = 0.45). A large correlation was observed between the control SPM and MS at 80% body weight (r = 0.52). Moderate correlations were observed between MS at 80% condition and tibial shock at 80, 90 and 100% conditions (r = 0.45).

The measurement tROM also was correlated with IC. Moderate correlations were observed between tROM at 90% condition and IC at 100% condition (r = 0.45), tROM at 100% condition and IC at 70% condition (r = 0.45), IC at 80% condition (r = 0.48) and IC at 90% condition (r = 0.47). Large correlations were seen between tROM at 60% condition and IC at 60% condition (r = 0.61) and 70% condition (r = 0.56); tROM at 70% condition and IC at 70% (r = 0.56); tROM at 80% condition and IC at 60, 70, 80, 90 and 100% conditions (r = 0.5, 0.5, 0.55, 0.5, 0.53 respectively); tROM at 90% condition and IC at 70 and 80% conditions (r = 0.57, 0.50 respectively).

Foot strike

Figure 4.5 shows the percentage of FFS compared to RFS incidences. The incidence of FFS compared with RFS patterns observed was 45% of the total number of observations in the 60% condition. For 70%, 80% and 100% conditions, 15% of foot strike incidences observed were FFS. During the 90% condition, 10% of foot strike patterns were FFS.

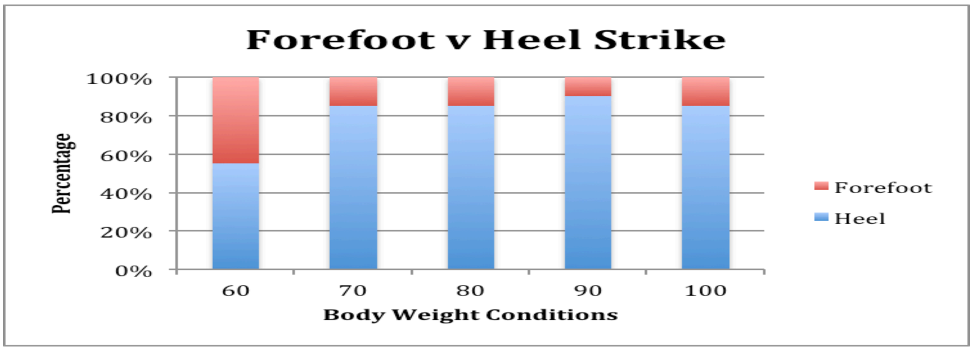

Figure 4.5: Percentage of forefoot and heel strike incidence for all conditions

4.3 Reliability Study

A velocity assessment was used to determine the reliability of the submaximal running used for this study. The ICC (0.78) and CV (0.6%) were both within acceptable ranges of reliability, therefore velocity set was considered good.

Test re-test reliability was high across all conditions for tibial shock (ICC > 0.82). The control condition of 100% body weight had the lowest ICC = 0.82. Similarly, the CV's were also within acceptable limits of variability (CV% = 0.9 – 3.0 %) across all conditions. The overall reliability of the measure for tibial shock was therefore considered 'good'. The ICC and CV's for tibial shock and gait kinematics are outlined in the reliability study results summary (Table 4.8).

Table 4.8: Reliability study results summary

		60%	70%	80%	90%	100%
Accelerometry	**ICC**	0.98	0.99	0.97	0.92	0.82
	CV%	1.70	0.90	1.30	1.80	3.00
SPM	**ICC**	0.99	0.99	0.98	0.95	0.98
	CV%	0.20	0.10	0.30	0.20	0.20
Initial Contact	**ICC**	0.99	0.97	0.98	0.99	0.97
	CV%	2.90	4.80	3.40	7.20	1.70
Midstance	**ICC**	1.00	0.99	0.98	0.98	0.98
	CV%	13.50	11.10	10.40	14.50	9.70
Terminal Stance	**ICC**	0.73	0.89	0.89	0.95	0.97
	CV%	4.10	2.80	3.40	2.30	1.80
tROM	**ICC**	0.73	0.67	0.74	0.54	0.98
	CV%	5.30	6.40	6.40	7.10	0.90

ICC = intra-class correlation; CV = Coefficient of variation

SPM - Strides per minute; tROM – total range of motion

Overall levels of reliability were good for SPM. Excellent ICC's and small CV's across all body weight conditions for SPM was observed. ICC's ranged from 0.95 (90% body weight) to 0.98 (80 and 100% body weight) and 0.99 for 60% and 70% conditions. The CV's recorded for SPM across all conditions ranged from 0.1% - 0.3%.

The ICC's for kinematic measures were excellent (>0.95) for most ankle angle variables, across all conditions. ICC's ranged from 0.97 – 0.99 and reliability was considered good for IC. CV's were also within the acceptable range for of error (CV% = 1.7 – 7.2) for IC. Overall reliability of IC measure was considered good. The ICC for TS at 60% (ICC = 0.73) was considered average. The CV's for TS were all considered to be small to moderate (CV% = 1.8 – 4.1). These are within overall reliability scores, therefore the test measure for TS was considered to be reliable. The CV's for MS were more variable (CV% = 9.7 – 14.5). At 60, 70, 80 and 90% conditions, MS CV's were outside the acceptable 10% error margin (CV% = 10.4 – 14.5) and variability was considered 'large'. The CV for MS for the control condition was 9.7% therefore the variability, or error was considered 'moderate'. All ICC's for MS were considered excellent (ICC =0.98 -1.00). As such, overall reliability for MS at 100%

condition was considered good. Reliability for 60, 70, 80 and 90% conditions were deemed moderate.

Ankle tROM ICC was within the acceptable range for reliability for conditions 60, 80 and 100% body weight. BWS conditions of 10 and 30% were below the acceptable reliability. All CV's were within and acceptable error range for tROM. Reliability of the measure for foot strike was considered good. Two (5%) differences in foot strike pattern were observed across all 40 observations (i.e. 4 participants, 2 trials, 5 conditions).

Chapter 5: Discussion

The purpose of this study was to investigate the effects of BWS treadmill running on tibial shock, foot strike pattern, and stride rate and ankle kinematics in elite National level athletes from a range of different sports. The main finding of the study was that there were significant decreases in tibial shock at 60 and 70% body weight conditions (ES = 0.71 and 0.59 respectively). At the 80% condition a moderate ES was observed, however a large amount of variability across kinematic measures was evident. At 90% condition only a trivial ES was observed (0.09). These findings support the effectiveness of BWS treadmill running for reducing tibial shock when BWS is increased to 20% or greater in athletes. There were also significant effects on a range of biomechanical gait variables in this group of athletes.

With respect to ankle angles small or trivial changes (ES = 0.01 – 0.24) were seen across all body weight conditions, with the exception of the moderate changes observed at IC (ES = 0.34) and MS (ES = 0.31) during the 60% condition. This suggests that ankle angles are not significantly affected as a result of unloaded treadmill running. This may be beneficial for the elite athlete if ankle kinematics are not a contributing factor to injury mechanisms, and are required to be preserved during rehabilitation.

It was also shown that footstrike patterns were consistent across 70, 80 and 90% conditions. Only at 60% condition, did footstrike shift from RFS to FFS by an additional 30%. There was also a large change for tROM at 60% condition (ES = 0.51). These results suggest that the change from RFS to FFS strike, which occurred at the higher level of BWS, was concomitant with a greater tROM at the ankle. This may have occurred due to the increase in air pressure within the treadmill chamber, lifting the participant up higher, or given the altered perception of body mass, and hence causing a change in coordination of segments at footstrike.

Significant effects on gait kinematics, and tibial shock were observed at the 60% condition and similarly at the 70% condition. These findings suggest that there may be an effective threshold of BWS for the elite athletic population, which may effectively reduce tibial shock, without negatively impacting gait kinematics. Stride rate increased linearly and inversely with the decrease in BWS across all conditions. This could be possibly due to the lift experienced during higher levels of BWS, and thus an increase in flight time and possibly stride length. It is well documented that at a set velocity, stride rate decreases as stride length increases (Mercer & Chona, 2015; Storen, Helgerud, & Hoff, 2011; Thomson, 2014).

5.1 Tibial shock

The first study in this thesis examined test-retest reliability of both kinetic and kinematic measures selected for the repeated-measures study. Tibial shock was measured using accelerometery. The ICC's for accelerometery within this study were higher (ICC 0.82 – 0.99) across all body weight conditions than has been previously reported (Henriksen et al., 2004; Turcot et al., 2008). In a study assessing the test-retest reliability of accelerometery for gait analysis in 20 healthy subjects, Henriksen et al (2004) reported an average level of reliability (ICC = 0.77) with the accelerometer placed on the trunk. Further, Turcot et al (2008), reported a good level of reliability (ICC = 0.81) for both tibial and femoral accelerometery (Turcot et al., 2008). The typical error observed in this study, for accelerometery across all body weight conditions, was small with CV's ranging from 0.9 – 3.0%. In the Henriksen et al. (2004) study, CV% for resultant peak tibial acceleration was 2.9%, which is similar to the highest level of error reported in the current study.

The overall reliability of tibial shock measures for this study was deemed to be good and is supported by previous research in untrained populations (Moore et al., 2010; Raffalt et al., 2013; Saxena & Granot, 2011; Tenforde et al., 2012)**.** The ICC's in this study may have been higher than previously reported due to a number of factors. Accelerometer placement on the tibia, rather than the trunk, has been shown to be a more reliable measure of tibial acceleration (Henriksen et al., 2004). Also, as the cohort of this study were elite athletes and well accustomed to running, the learning effect between test-retest trials may not have been as significant as might be observed in untrained populations. Further, the athletes that participated in the reliability study performed the re-test at the same time of day, and after the same training conditions they had experienced prior to the initial testing day. These controls have contributed to the stability of the reliability data. The high level of reliability validates the use of accelerometery as an effective measure of tibial shock in elite level athletes. This is beneficial for future studies, and for training and rehabilitation purposes where tibial shock may be assessed during real game and competition situations.

The observational study in this thesis examined the effects of BWS treadmill running on tibial shock and gait kinematics. The change in means and associated ES, suggested the likelihood of a mechanistic effect between control and condition trials with the reduction in tibial shock across body weight conditions. The results expressed as ‘g’ for tibial shock when adjusted for positive accelerations (Laughton et al., 2003) were similar to previous studies that evaluated the effect of downhill running and stiffness (Chu & Caldwell, 2004), footstrike (Laughton et al., 2003) and fatigue (Mizrahi et al., 2000; Mizrahi et al., 2000) on tibial shock. However, the tibial shock measures in this study were slightly higher when expressed as g, than previously reported (Derrick, Hamill, & Caldwell, 1998) (Appendix 8). The difference in g may be explained by differences in data processing methods used by the researchers.

In the current study, large correlations for tibial shock were expected within the same outcome measure. Although a very large correlation ($r = 0.79$) was observed between the control and 60% conditions, there were nearly perfect correlations between 70, 80 and 90% conditions and the control condition ($r = 0.91 – 0.95$). It is therefore evident that the 60% condition is the least similar to the control condition when compared against other BWS conditions. This is supported by the significant ES (ES = 0.71) seen for tibial shock at 60% condition compared to the control. In addition, the magnitude-based inference, for 60% condition, suggested a 100% substantially positive effect of reducing tibial shock at 60%. Further, qualitative measures considered the change from 100% to 60% body weight ‘most likely positive’ at reducing tibial shock. These results highlight the effectiveness of reducing tibial shock while running at submaximal velocities at 60% body weight, for elite level athletes. The magnitude of this effect may be considered beneficial in the early stages of rehabilitation, particularly if the attenuation of impact forces acting upon the lower limb is the most important variable to consider (Hreljac, 2004). Due to the magnitude of reduction in tibial shock, it is also important to consider the kinematic effects of this level of BWS.

At the 70% condition, the results suggested a reduction in tibial shock to be ‘very likely positive’. A large ES was also observed (ES = 0.59) for the 70% condition against the control condition, indicating the effectiveness of reducing tibial shock during submaximal running at 70% condition for this cohort. The findings for both 60 and 70% conditions suggest tibial shock is effectively reduced at these BWS levels. These findings are beneficial for the practitioner in that there are significant reductions of tibial shock during both conditions, and also the progressive nature of this reduction. With the knowledge that tibial shock is attenuated progressively with increased BWS, it can also be assumed that progressive re-loading is possible with the reduction of BWS. This alone may be positive for the progressive overload of injured tissue; however kinematic variables must also be considered.

At 80% condition, the change in tibial shock was moderate (ES = 0.35) however there was still an 89% likely reduction in tibial shock at this BWS level. These findings support the assumption that progressive re-loading during BWS submaximal running is possible. Conversely at 90% condition it was only a 6% likelihood that tibial shock was reduced (ES = 0.09). These results highlight that only a small reduction in tibial shock occurs at this BWS level as highlighted by the group mean for tibial shock (Appendix 8). The difference in effect between 80% and 90% conditions indicate the need for a carefully monitor transition from 20% down to 10% BWS. At the 80% body weight condition, it is evident that changes in kinetic and kinematic variables occur. Ankle kinematic measure, MS at 80% condition, was moderately but significantly correlated to tibial shock at 80, 90 and 100% conditions (r = 0.50, 0.45, 0.45). Active loading peaks occur during MS as the tibia moves over the foot, absorbing the vertical ground forces acting in the opposite direction to the body's COM vector (Novacheck, 1998). These ground forces are experienced through the lower leg as tibial shock. These findings suggest that at MS during the 80% condition, there is a relationship between increasing rates of tibial shock and decreasing BWS of up to 20%. Indeed, previous studies have reported a return to overground running to be possible once the athlete has reached 85% body weight without pain (Draovitch et al., 2012; Saxena & Granot, 2011). Further research into the effect of transitioning between 80% and 90% conditions may provide insightful information as to the precise threshold at which BWS treadmill running shifts from being significant to non-significant for reducing tibial shock.

The findings in this study in relation to tibial shock are supported by previous research, reporting a progressive decrease of tibial shock (Mercer & Chona, 2015) and active vertical ground forces (Grabowski & Kram, 2008; Raffalt et al., 2013) as BWS is increased. The research by Raffalt et al (2013) was the only study that used highly trained athletes. Raffalt et al (2013) observed decreases in mean and peak vertical ground reaction forces as well as a decrease in stride frequency with increased BWS from 25% to 75%. Similar to the findings of this study, stride frequency was suggested to be due to an increase in flight time (Raffalt et al., 2013). Mercer et al (2015) recently assessed ten healthy active participants and observed a reduction in tibial impact accelerations across 60, 70 and 80% body weight conditions with respect to full body weight running. This was similar to findings by Grabowski et al (2008) who also assessed ten healthy participants. Vertical ground forces were assessed at 25%, 50% and 75% of BWS. Significant decreases in active peaks were also observed across all conditions in relation to full body weight running at a similar velocity to that used within this research, i.e. $3 m.s^{-1}$ and also at $4 m.s^{-1}$ (Grabowski & Kram, 2008).

5.2 Stride Rate

In a recent study, Riva et al (2014) recently reported an average level of reliability for measuring stride rate via video analysis methods and suggested the need for a high number of strides (125 - 127) to improve reliability and limit variability between testing days (Riva, Bisi, & Stagni, 2014). The current thesis assessed three minutes of video footage, per participant, per condition. This provided a range of between 210 to 290 strides to analyse. As such a high level of reliability and low level of variability was achieved. The test-retest reliability for SPM was nearly perfect across all conditions. ICC's ranged from 0.95 (90% condition) to 0.99 (60 and 70% conditions) with CV's ranging from 0.10 to 0.30, indicating excellent reliability. The spread of data for SPM was also consistent between conditions (SD ~ 3.5). Increased SPM has been reported as favourable to improve gait kinematics that may have contributed to overuse injury (Hafer, Freedman Silvernail, Hillstrom, & Boyer, 2015). In addition to this, SPM has been investigated in relation to ground forces during running (Hobara, Sato, Sakaguchi, Sato, & Nakazawa, 2012). At a set running velocity, limb loading measured as vertical ground force, has been shown to decrease with an increase in stride frequency (Hobara et al., 2012) with no significant change to muscle activation measured by EMG (Chumanov, Wille, Michalski, & Heiderscheit, 2012). Hobara et al (2012), in a study of 10 male runners, demonstrated that a 15% increase in step frequency decreased loading variables such as vertical ground forces. This was possibly due to a change of foot strike pattern or due to changes in ankle and knee angles during impact. Reductions in tibial accelerations have been reported with a 5% increase in SPM, resulting in reduced joint loading at the knee (Heiderscheit, Chumanov, Michalski, Wille, & Ryan, 2011). Chumanov et al (2012) assessed the muscle activation patterns of 45 healthy recreational runners. Participants ran at a preferred stride frequency, and with 10% greater, and 10% less than their preferred SPM. While muscle activation remained the same during loading, pre-activation of tibialis anterior and rectus femoris during the swing phase was observed. This pre-loading may alter joint moments and energy absorption rates, in particular at the knee as SPM increases (Chumanov et al., 2012). The reduction in step length as a result of an increase in SPM at a set velocity, has also been linked to reductions in patello-femoral pain (Willson, Sharpee, Meardon, & Kernozek, 2014). Further, Lieberman et al (2015), in a study of 14 runners, found that an increase of five SPM resulted in an increase of 5.8% in sagittal plane hip flexor moment during the swing phase. This increase in hip flexor moment may result in a decrease in the horizontal distance between landing positions of the foot in relation to the hip, thereby reducing impact peaks.

In this study, SPM decreased linearly with an increase in BWS. IC and foot strike were significantly affected during the 60% condition. There was a large negative correlation (r = -0.59) between SPM and IC ankle angle suggesting a reduction in SPM with greater plantar flexion. BWS is more likely to have influenced stride length due to increased flight time at higher BWS (Crowell et al., 2010; Laughton et al., 2003), consequently affecting stride frequency. Mercer et al (2015), in a study of 10 runners also observed a decrease of stride length and an increase in tibial acceleration as BWS decreased in 10% increments, however this was at higher levels of support than the current study. In comparison, participants of the Lieberman et al (2015), and Hobara et al (2012) studies, deliberately controlled their stride frequency in response to an audible tone. The reduction in SPM at higher BWS conditions may contribute to the total unloading effect of BWS treadmill running. Further the gradual increase in SPM as BWS decreased, may also accommodate the gradual re-loading of musculoskeletal tissue with the progression back to overground running following injury. This is supported using an applied spring-mass model example (Morin, Samozino, Zameziati, & Belli, 2007) which is an effective theoretical model of determining vertical ground forces during hopping and running (Blum, Lipfert, & Seyfarth, 2009).

Stride frequency has been strongly correlated to ground contact time. Specifically, the deliberate increase in stride frequency during treadmill running at $3.33 m.s^{-1}$ has been shown to decrease contact time, which may indirectly influence leg-spring stiffness. This gradual increase in leg-spring stiffness may be an important factor to consider during the rehabilitation process, particularly in relation to the type of lower limb injury experienced. While greater levels of leg-spring stiffness have been correlated to performance, high levels of leg stiffness may be implicated in the increase risk of bone related overuse injuries, while lower levels of leg-spring stiffness may be important in soft tissue lower limb injuries (Butler, Crowell Iii, & Davis, 2003).

5.3 Ankle kinematics

The reliability of video analysis measures between days for determining ankle kinematics was varied. The reliability scores for ankle kinematics measure, IC was deemed to have good reliability at 60, 70, 80 and 90% conditions with excellent levels of reproducibility and small to moderate levels of error for this measure. The highest level of variability observed, was at 90% condition for IC, was considered to be moderate and was equivalent to a 2.4^{o} difference in ankle angle. This level of error is supported to be reliable by a previous study in which authors assessed the reliability of using model-based image-matching motion analysis to measure ankle joint kinematic measurements (Mok et al., 2011). Bone pin markers were used on cadaver models to determine accurate ankle angles for model-based motional analysis. The methods used were deemed reliable and typical error for sagittal plane ankle angle was found to be 3^{o} (Mok et al., 2011). In a similar study assessing the reliability of 2D video analysis for determining sagittal plan knee kinematics (Damsted, Nielsen, & Larsen, 2015), authors deemed a difference of three to nine degrees reliable for knee flexion. This suggests sagittal plane video analysis is reliable, however it is noted that degrees of freedom at the ankle and knee may not be comparable due to the differences in joint motion, structure and function.

The MS kinematic measurements at all conditions were shown to have excellent ICC scores (ICC = 0.98 – 1.00) for test-retest between days. The variability of MS as measured by CV were considered to be large with all conditions with the exception of the control recording CV > 10%. The CV's ranged from 10.4 to 14.5% for MS. This equated to 4.0 to 5.2^{o} difference in MS angle between days, suggesting that there was large variability. Overall reliability of this measure, when considering both ICC's and CV's, was therefore considered poor for all BWS conditions. It has been reported previously that runners who mid-foot strike display the greatest variability in sagittal plane kinematics during MS (Stearne, Alderson, Green, Donnelly, & Rubenson, 2014). This may have accounted for the variability seen at 60% condition where footstrike shifted from RFS for FFS in an additional 30% of incidences as compared to the control. The reliability for the control conditions was considered to be good. While MS had the largest range of error, TS was the least consistent across conditions with respect to ICC (0.73 – 0.97). The control condition had the highest level of reliability with the lowest error rate (ICC = 0.97, CV% = 1.8), while the 60% condition had an average level of reliability (ICC = 0.73), and the highest error (CV% = 4.1) albeit within acceptable ranges to be deemed a reliable test measure. A possible explanation for this may have been due to some of the video footage being obscured at TS. The clear window at the side of the treadmill partially occluded visibility of some of the taller participants' rear foot. As such, the ankle marker on the lateral malleolus was partially obstructed in some cases, making visual identification of this marker difficult during some of the analysis. All error measures were within reliable ranges, and were similar to previously mentioned research with respect to angles.

During the gait cycle, the impact peak occurs at IC. Footstrike is also determined at IC. BWS significantly affected IC at 60% condition where the magnitude-based inference suggested an 85% likely chance of change in ankle angle at IC. Additionally, it was at 60% condition that foot strike patterns were significantly different with 45% of foot strike incidences being FFS, as compared to only 15% at 70, 80 and 100% conditions and 10% at 90% condition. During FFS running, braking forces and impact peaks are reduced. This is due to the foot contacting the running surface closer to the body's COM (Lieberman et al., 2010), in a more plantar flexed landing position (Lorenz & Pontillo, 2012). A FFS strike pattern may reduce the knee joint moments and associated patella femoral pain (Diebal, Gregory, Alitz, & Gerber, 2011; Willson et al., 2014) as well as reduce intra-compartmental pressure associated with compartment syndrome (Diebal, Gregory, Alitz, & Gerber, 2011) commonly experienced by athletes. There is more plantar flexion at IC during FFS strike, resulting in greater ankle instability and range of motion. In non-habitual FFS runners, this may increase the risk of ankle and lower limb kinetic and kinematic variables, and possibly injury. Stearne et al (2014), assessed the kinetic and kinematic variability of switching between habitual and imposed foot strike patterns during running. Sixteen highly trained male runners were assessed under both habitual and imposed FFS and RFS conditions. Sagittal plane ankle kinematics were similar in the habitual RFS group when they ran with a FFS pattern, to the kinematics observed of habitual FFS runners, however there was a significant increase in ankle internal rotation (33%), produced ankle power (21%) and absorbed ankle power (85%) in the RFS group when running with a FFS pattern (Stearne et al., 2014). Statistical analyses in the current study suggested that a significant change in tROM at both 60 and 70%

conditions was very likely. In addition to this, tROM was also significantly correlated with IC (Appendix 9). These relationships suggest that IC influences the tROM, more so at the higher levels of BWS where larger correlations were observed. The juxtaposition to the relationship between plantar-flexion and dorsi-flexion loading through tROM, is mentioned by Stearne et al (2014) in relation to power absorbed and produced during imposed FFS running. These findings demonstrate the risk that changes to gait kinematics may have on joint moments and other kinetic variables in habituated foot strike patterns. The benefit a FFS strike pattern may have on more proximal joints such as the patella-femoral joint may outweigh these risks however. This highlights the need to carefully evaluate the effects of altering gait kinematics during running with BWS for the specific rehabilitation purpose, particularly if knee or ankle pathology are evident.

During the stance phase of gait, active loading peaks occur. This is the phase of the gait cycle where tibial shock is experienced at the greatest magnitudes, in particular from MS to late stance. In athletes presenting with stress related symptoms, the rate of loading experienced during the stance phase may contribute to further damage to affected tissue, as well as pain mechanisms. The highly adapted neuromuscular system is able to respond to a high level of kinematic disturbance during MS, the caveat being healthy tissue and neuromuscular function (Hafer et al., 2015). As the ability to respond to perturbation diminishes with injury, the ability to minimise intrinsic and extrinsic variation during rehabilitation is as important as reducing tibial loading. During 60% condition, there was a very likely chance that there would be a reduction in MS ankle angle at 60% condition compared to the control. These findings indicate the reduction in loading rate during MS with 40% BWS. This was supported by the reduction in tibial shock at the 60% condition (Appendix 8). These findings suggest a likely change in gait kinematics at this higher level of BWS. Changes in MS kinematics are possible at 70 and 80% conditions however are very likely to be trivial at 90% condition. Changes in joint kinematics and segment coordination have been implicated with overuse injuries due to the abrupt alteration of stress to tissues not accustomed to repetitive loading (Hafer et al., 2015; Osu et al., 2002). Further, injured runners demonstrate altered segment coordination and coordination variability (Hamill, Palmer, & Van Emmerik, 2012). Optimal coordination variability is associated with highly refined motor patterns, and healthy states (Stergiou, Harbourne, & Cavanaugh, 2006). Coordination variability tends to be highest during MS (Hamill, Van Emmerik, Heiderscheit, & Li, 1999). In the non-injured athlete this allows for adaptation to perturbations and normal motor development and segment coordination. Large changes in MS kinematics as observed at 60% condition, may be a contraindication for the injured athlete and their ability to adapt to perturbations, likely to be experienced given the greater rate of ankle compliance and rotation previously discussed at the higher level of BWS.

All changes in ankle kinematics at TS likely were trivial apart from 70% condition. There was still a 38% likelihood of a trivial change in kinematics at this level of BWS. Although ES for IC, MS and TS were small or trivial across all conditions, considering the likely or unlikely possibility of changes to gait kinematics may be important for highly technical runners where by minor alterations to technique could negatively impact performance over time. When designing an effective return to play rehabilitation program, considering the combined effects of the kinematic and kinetic variables is important. For example, while tibial shock is most greatly reduced at 60% condition, IC, MS and foot strike are also affected greatly. With the effective reduction of tibial shock at 70% condition for example, as well as the reduction in the likeliness that MS kinematics will be altered, this may be a better level of BWS to begin early stage rehabilitation programs at.

5.4 Foot Strike

Foot strike may be a contributing factor to the risk of lower limb injury (Rooney & Derrick, 2013). Heel strike pattern may result in a greater risk in knee pain and injury in runners. This is suggested be due to an increase in knee extensor moments during running. Heel strikers' display limited knee flexion at IC, often with the heel in front of the COM, and therefore with increased braking propulsions (Lieberman et al., 2010; Rooney & Derrick, 2013). Further, greater knee flexion during midstance may be observed as the COM continues to downward until it is over the support leg. As this is the active peak, high levels of forces are experienced on the lower limb at this phase of the gait cycle. In comparison, athletes who FFS strikes may be at greater risk of sustaining an injury in the foot, ankle or lower leg compartment (Hatala et al., 2013). The increase in plantar flexion during IC increases mechanical load and tensile stress on the ankle (Damsted, Larsen, & Nielsen, 2015; Damsted, Nielsen, et al., 2015). The reliability for footstrike was deemed high, with two of 40 (5%) observed foot strike occurrences differing between testing occasions. Variability in this study may have been due to intra-rater error between testing days, or as a result of subject variability between testing days. A recent study assessed the intra- and inter-rater reliability of visual identification of foot strike pattern during treadmill running using two-dimensional video analysis (Damsted, Larsen, et al., 2015). These methods are in agreement with this study with high intra-rater reliability between and within days.

5.5 Overuse injury rehabilitation

The effectiveness of using LBPP treadmills for reducing tibial shock during submaximal running was demonstrated in this study. The unloading of the lower limb during impact and active peaks during running is a critical component to rehabilitating the lower limb from overuse injuries. Bone and connective tissues are dynamic tissues that accumulate micro-trauma from load and impacts. Excessive loads contribute to overuse injury in at sites such as the tendon and bone in the lower limb in athletes. However, tendon and bone also require mechanical stress for remodelling when injured.

It is well known that mechanical stress is a potent stimulus for bone cell activity. Osteocytes are strain-sensitive cells that can transduce mechanical signals derived from mechanical loading into cues that decrease bone loss and increase bone formation (Robling, Castillo, & Turner, 2006). Overuse or excessive mechanical loads cause osteocyte death which causes micro-damage to the bone (Robling et al., 2006). The increase in micro-damage is associated with the progressive loss of bone strength and stiffness, resulting in structural failure as evident in stress fractures. Mechanical stress also improves bone strength and collagen alignment with new bone tissue formation. Strength increases in long bones such as the tibia, through new bone formation at the periosteal layer of the bone. Formation steadily increases with increased loading (Robling, Hinant, Burr, & Turner, 2002). High levels of BWS can reduce tibial shock in the early rehabilitation stages and therefore limit any further structural failure of the bone. Progressive re-loading shown by the increase in tibial shock with a decrease in BWS during treadmill running, can accommodate the required re-loading of bone for adaptation. Loading magnitude and frequency are important determinants of bone morphology. A higher frequency of mechanical loading enhances formation of bone (Hsieh & Turner, 2001). Findings from this study showed an increase in SPM with the decrease of BWS, this increase in stride frequency is positive for effective bone formation. As bone loses more than 95% of its mechanosensitivity after approximately 20 loading cycles, the duration of re-loading as well as appropriate rest bouts, are important factors to consider when planning rehabilitation protocols for athletes with lower limb stress fractures (Robling et al., 2002). This disputes the rationale of traditional rehabilitation protocols where over-ground walking duration is the primary stimulus for continued bone reformation. As such, the use of LBPP treadmill running may lend favorably towards a gradual but more rapidly progressed re-loading protocol to support this. Using a running protocol, which resembles the athlete's healthy state kinematics, may also contribute to the effectiveness of the bone remodeling process. Due to the poor neural innervation to bone cells and ensuing lack of ability to integrate and distribute mechanical signals from the central nervous system, bone requires a similar mechanical environment to that previously exposed to in order to adapt to a new mechanical loads (Robling et al., 2006). Choosing an appropriate level of BWS that limits kinematic variability may therefore also be important to rehabilitation programs for athletes suffering stress fractures.

Both tendon and connective tissues interact closely with contractile elements of the skeletal muscle in order to transmit force. Similar to bone tissue, tendon and connective tissue respond to mechanical loading. The development of overuse injuries in tendon involves morphological and biochemical changes such as altered collagen expression and tendon thickness (Kjaer, 2004) To counteract the responses to unaccustomed or excessive stresses, the adjustment of load has been shown to be more advantageous over the absence of load in the affected tissue (Kjaer et al., 2005). Reducing pain-inducing loads in the acute stages of tendon or connective tissue overuse injuries is important. Following which increasing mechanical load to the affected tissue should be initiated. The gradual re-loading of tendon and connective tissue through training stimulates biomechanical changes to occur. An increase in growth factors responsible for stimulating collagen and extracellular matrix protein synthesis occurs following mechanical loading (Hobara et al., 2012). This enhances viscoelastic characteristics and increases the cross-sectional area of the collagen, both of which contributes to the resilience of the tendon (Kjaer et al., 2005). Eccentric load (Crowell et al., 2010) and heavy resistance training (Henriksen et al., 2004) have been shown to be effective in the modulation of pain symptoms and performance of the injured tendon.

Tibial shock results from the transmission of forces from the ground, and through the structures of the foot and ankle into the tibia (Mizrahi et al., 2000). As tendons detect tension during loading and moves body segments during muscle contraction, it may be assumed that the reduction in body weight during BWS running also results in a reduction of stress on the lower limb tendons during both loading and during movement. The ability to reduce forces being transmitted through the tendons of the lower limb while running on the AlterG treadmill may be beneficial for athletes to limit muscle atrophy. The re-loading capabilities of BWS treadmill running may also be advantageous for performing running specific tasks during the rehabilitation process, to enhance the neuromuscular re-education. While the use of the AlterG treadmill is effective at unloading and re-loading tendons following Achilles tendon surgery (Saxena & Granot, 2011), there is substantial evidence suggesting heavy load and extended time under tension is important for the biochemical and biomechanical morphology required for the overuse injured tendon to occur (Alfredson et al., 1998; Beyer et al., 2015; Kulig et al., 2009; Murtaugh & Ihm, 2013; Silbernagel, Thomee, Thomee, & Karlsson, 2001). Additional load may be applied to the rehabilitating tendon using the AlterG treadmill by adjusting running velocity as previously reported (Brughelli et al., 2011; Mercer & Chona, 2015). It is however important to recognise that a combination of clinical, functional and sport specific treatments, rather than a single rehabilitation or exercise variable, is required for the successful return to play rehabilitation strategy.

Chapter 6: Summary and Practical Applications

6.1 Summary

The purpose of this thesis was to examine the effect BWS treadmill running has on tibial shock and sagittal plane gait kinematic variables in elite athletes. The magnitude of tibial shock has previously been linked to lower limb injuries such as stress related overuse injuries (Beck, 1998; Newman et al., 2013) and tendinopathy (Maffulli et al., 2003; Magnan et al., 2014)**.** There are a number of influencing factors that may lead to athletes experiencing over-use injuries of the lower limb (Ferber et al., 2009b; Stanish, 1984; Wen, 2007; Wilder & Sethi, 2004; Yang et al., 2012) and also a number of methods of unloading (Aaslund & Moe-Nilssen, 2008; Kilding et al., 2007; Killgore, 2012) and rehabilitation options (Crowell et al., 2010; Galbraith & Lavallee, 2009; Moen et al., 2012; Raasch & Hergan, 2006; Tolbert & Binkley, 2009) for the return to running. A biomechanical approach was taken in this study to provide insight to the loading rate and mechanical factors of the lower limb during BWS running in the elite athlete population.

The kinetic and kinematic assessments used in this study were assessed for reliability. The procedures described for the measurement of tibial shock had very good reliability across all conditions. All kinematic measures, with the exception of MS ankle angle, generally had very good reliability across all conditions. MS ankle angle was reliable for the control condition; however poor reliability was observed at BWS conditions. The methodology used for tibial shock and all kinematic assessment measures compared favourably with previous studies and could easily be replicated in subsequent research. The protocols within this research could also be used in the elite training environment or laboratory with similar equipment and facilities. Further, with the use of wireless accelerometery, there is the potential to use this technology in the field. This may provide insightful information to the practitioner and sport coaches regarding loading rates in the lower limb during various sport specific running tasks. This information may further contribute to monitoring load in at risk athletes, athletes returning to sport following injury or creating loading baselines for specific positions or individuals.

The observational study demonstrated a decrease in tibial shock with an increase in BWS, highlighting the effectiveness of the use of LBPP BWS treadmills for reducing tibial load at submaximal running velocities for elite athletic populations. Interestingly the decrease in tibial shock was not linear. When considering the effect of tibial shock independent of other kinematic factors, the 80% to 90% conditions revealed the most interesting changes. In this study, it may be that 20% of BWS, is considered the threshold of significant unloading of tibial shock for the elite population.

Kinematic variables were significantly affected by BWS at 60% and 70% conditions, with some affected favourably. Stride rate for example, decreased linearly with increased BWS. The inverse to this is that as BWS decreases, SPM increased. An increase in SPM at a given velocity has been correlated with shorter ground contact times and an increase in leg spring stiffness (Morin et al., 2007) both of which are beneficial to performance. The gradual increase in SPM with the reloading of body weight may contribute to the progressive overload of lower limb tissues during rehabilitation also. Conversely, foot strike, was affected as BWS increased to 40%. For athletes working on gait retraining or un-injured athletes adding volume to their training regime, this may not be problematic. For the injured and rehabilitating athlete however, a shift from a habituated strike pattern and towards a FFS pattern may have implications with segment coordination and neuromuscular control (Stearne et al., 2014). It is important to consider the interaction of these kinematic variables with the rate of tibial unloading.

The observations made within this research have provided useful insight into the effects of BWS treadmill running on some factors influencing lower limb biomechanics in elite athletes. The effect BWS treadmill running has on each of these biomechanical variables, independent of one another, is important to consider when designing return to sport rehabilitation programmes. It is also important to be aware of subsequent interactions between other biomechanical variables that may occur due increased BWS. For example, does an altered foot strike pattern with increased BWS, also affect knee kinematics? Further research is required to determine more accurately the threshold of where BWS no longer has a significant effect on tibial shock measures, specifically in small increments of increase BWS (1%). Moreover, research that assesses three-dimensional gait analysis may provide even more insight to the specific alterations to gait kinematics in conjunction with BWS.

6.2 Practical Applications

Load and impact are two of the common variables implicated in many overuse injuries in athletes. Bone and connective tissues are dynamic tissues that accumulate micro-trauma from load and impacts. Paradoxically, tissues such as tendon and bone require mechanical stress for cellular renewal when healthy, as well as remodelling when injured. Mechanical stress is a potent stimulus for bone tissue reformation, and mechanical load is a crucial component to the remodelling of connective tissue such as tendon. Combined with an adequate rehabilitation treatment plan, running with BWS on a LBPP treadmill provides an environment for effective and specific rehabilitation. The progressive and graduated re-loading of tibial shock using the AlterG, in a controllable environment, appears to be effective for an elite athlete recovering from lower limb overuse injuries.

The findings in this study, suggest the body weight condition of 80% (i.e. 20% BWS) appears to have the most effective combination of kinetic and kinematic results for unloading the injured tissue without compromising technique. For load and impact related injuries, such as MTSS, where reducing tibial shock is the key element to the initial stages of rehabilitation, a higher BWS of 30% may be more appropriate level to start at. The caveat to this is the mechanism of injury if known. This greater level of BWS may not be advantageous if the injury was caused as a result of change in ankle kinematics. Specific considerations should be made for each individual case. At 80% condition, IC, MS and foot strike patterns were not compromised, but tibial shock was significantly reduced. Therefore, it may be beneficial for athletes with injuries of the foot or ankle to commence submaximal running at ≤ 20% BW. Whereas athletes rehabilitating patella tendinopathy, may find the benefit of higher levels of BWS of ≥ 30% due to the shift towards FFS strike and an increase in SPM. Both biomechanical variables are reported to unload the patella tendon which is particularly beneficial in the early stages of rehabilitation in order to reduce pain. Decreasing BWS loads gradually and in conjunction with pain score monitoring is crucial.

Increases in BWS at 10% increments were used for the purpose of this research, however in practice, smaller increments may be more applicable for reloading recovering tissue. This is particularly important as athletes' approach a body weight condition of 80% and begin to progress through this and up to 90% of body weight. Comparable research suggests return to overground running may commence at 85% body weight without pain (Saxena & Granot, 2011). Based on the findings in this study, it is recommended that during the transition from 80% up to 90% of body weight increments of 2 to 3% are used for each progression. As only trivial changes were observed in the 90% condition (10% BWS) with respect to tibial shock unloading and changes in gait kinematics, this may be an effective level of support for the transition back to performance training and competition. This could be particularly critical in the cases where the athlete needs to build confidence in their ability to run, or where the volume of running needs to be increased and/or monitored either for fitness or during the initial return to play stage.

For the non-injured athlete, BWS treadmill running may also be an appropriate method of increasing volume or intensity in cases where additional aerobic capacity, body composition management, or high speed running to stimulate anaerobic responses is required (Figueroa, Manning & Escamilla, 2011; Grabowski et al., 2008). Based upon findings in this study, BWS treadmill running may be advantageous in stimulating physiological and neuromuscular adaptations in the elite population to enhance performance when BWS is set at 15% of support or less.

6.3 Study Limitations

There were a number of limitations to this study that warrant discussion. The ranges of potential outcome measures of this study were limited due to the design of the AlterG treadmill. The model of treadmill used within this study did not have built in force plate technology in the treadmill belt; therefore, direct force measures could not be obtained. Determining direct force measures during this study would have been advantageous so as to correlate the direct relationship between tibial acceleration measures and ground forces. A possible solution to mitigate this limitation for any future research may be to run additional control and reliability groups on a force measuring treadmill.

Two-dimensional video analyses were performed in these studies. This was due to the location of the treadmill, whereby walls were blocking the space required to set up the three-dimensional video equipment and required lighting. Three-dimensional analysis would have provided a more comprehensive insight into the kinematic gait changes during BWS running.

Another limitation as a result of the design of the AlterG was the canvass chamber surrounding the treadmill. The chamber is made from solid black canvass materials, with the exception to small viewing windows at each side and ends. The size of these clear windows obscured the lower limb from above the knee up. Collecting sagittal plane hip and knee kinematics would have provided a more complete picture of the participant's sagittal plane gait patterns and any changes that may have occurred more proximal to the torso. This information would be valuable to make further inferences on gait kinematics in relation to knee and hip related injuries.

The number of participants in this study (n=20) was in alignment to the estimated sample size calculated using an equation from www.sportsci.org website, and greater than used in similar studies assessing elite subjects (Raffalt et al., 2013). However, a larger sample size may provide greater power for estimation of any inferences derived from the participant group assessed. In this study, a larger sample of male, female, team and individual athletes may have proven beneficial for deriving inferences based on these sub-groups. There are often difficulties in recruiting large numbers of participants in this cohort. Some of these include the training demands of the elite athlete and reassuring the athlete, their coaches and their national sporting organisation that the additional work data collection does not impact training or recovery, in any way. Athlete fatigue, injury and illness may also hinder the recruitment processes, all of which are common in elite athletes at some level. As our exclusion criteria were for the athlete to be injury free, this eliminated some prospective participants'. The challenges experienced with recruitment numbers within this study were due to injury and training load.

Further research would be beneficial to determine the effects of BWS treadmill running during the transition form 20% down to 10% BWS. This would be advantageous for more accurately determining the effects of decreasing BWS in smaller increments when approaching the return to overground running stage of rehabilitation. Also assessing these kinetic and kinematic measures at faster running velocities may be useful for end stage rehabilitation to bridge the gap between rehabilitation and training readiness. With a modified treadmill design as seen in more recent models of the AlterG, a broader range of kinematic measures may also be addressed. This could be beneficial for developing more accurate rehabilitation protocols for specific lower limb injuries. From this further training studies comparing these protocols with more traditional rehabilitation protocols could be assessed for validity and robustness.

While running with BWS on a LBPP treadmill is effective for unloading the lower limb from tibial shock, and may also be an effective piece of rehabilitation technology, the interaction of kinetic and kinematic changes should be considered specific to the lower limb injury. Therefore, it is important to include a combination of clinical, functional and sport specific treatments, such as strength training and physiotherapy treatment, in conjunction with the use of the AlterG for the successful return to play rehabilitation strategy of elite athletes.

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Appendices

Appendix 1: Ethical Approval

AUTEC

SECRETARIAT

22 August 2013

Mike McGuigan
Faculty of Health and Environmental Sciences

Dear Mike

Re Ethics Application: **13/198 The effect of increasing body weight support on gait kinematics and tibial shock during treadmill running.**

Thank you for providing evidence as requested, which satisfies the points raised by the AUT University Ethics Committee (AUTEC).

Your ethics application has been approved for three years until 21 August 2016.

As part of the ethics approval process, you are required to submit the following to AUTEC:

A brief annual progress report using form EA2, which is available online through http://www.aut.ac.nz/researchethics. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 21 August 2016;

A brief report on the status of the project using form EA3, which is available online through http://www.aut.ac.nz/researchethics. This report is to be submitted either when the approval expires on 21 August 2016 or on completion of the project.

It is a condition of approval that AUTEC is notified of any adverse events or if the research does not commence. AUTEC approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to obtain this. If your research is undertaken within a jurisdiction outside New Zealand, you will need to make the arrangements necessary to meet the legal and ethical requirements that apply there.

To enable us to provide you with efficient service, please use the application number and study title in all correspondence with us. If you have any enquiries about this application, or anything else, please do contact us at ethics@aut.ac.nz.

All the very best with your research,

K O'Connor

Kate O'Connor
Executive Secretary
Auckland University of Technology Ethics Committee

Cc: Marissa Downes marissa.downes@hpsnz.org.nz

Appendix 2: Participant Information Sheet

Participant Information Sheet

AUT UNIVERSITY

Date Information Sheet Produced:

17/07/2013

Project Title

The effect of increasing body weight support on gait kinematics and tibial shock during treadmill running.

An Invitation

I, Marissa Downes, am a Masters student based at the Sports Performance Research in New Zealand at AUT-Millennium Institute, School of Sport and Recreation, Faculty of Health and Environmental Sciences.

I would like to invite you to participate in a research study using biomechanical analysis to assess alterations in gait kinematics and kinetics with body weight support. Participation is entirely voluntary and you may withdraw at any time without any adverse consequences.

What is the purpose of this research?

The purpose of this study is to investigate the changes in running technique and ground reaction forces in the lower limb using the Alter-G treadmill (shown in Figure 1), designed to unweigh the body during rehabilitation and for performance purposes. This study makes up part of my proposed Master of Sport and Exercise Science degree.

How was I identified and why am I being invited to participate in this research?

You were identified as a participant for this research based on your level of involvement in your sport and by meeting all inclusion criteria as outlined in advertising for this research project. All participants are national level and highly trained athletes aged between 18 – 35 years.

Exclusion criteria includes any injuries current, or in the past 2 weeks which have hindered or stopped normal training.

What will happen in this research?

You will be assessed on three occasions. The first two sessions will be familiarization with the equipment and assessment protocol and the final session will be the actual data collection. Each session will involve having an electro-goniometer (small electronic device placed across a joint which measure's joint angles) attached wireless inertial sensors attached to the lower portion of the leg (anterio-medial tibia). You will also be filmed with high-speed 2-D cameras to capture gait kinematics. i.e, lower limb running technique. Prior to data collection you will undertake a standardised warm-up protocol consisting of light jogging on the AlterG Treadmill. Each testing session will take approximately 60-mins including the warm-up.

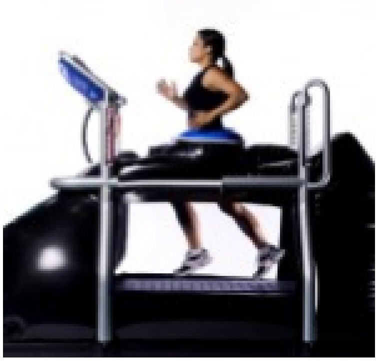

Figure 1. AlterG Treadmill

What are the discomforts and risks?

There are minimal anticipated discomforts and risks from participating in this testing. The training induced discomfort and fatigue will be similar to or less than that of your regular sport training and testing sessions. You may experience some mild fatigue in your legs; this response is normal and triggered by the onset of any exercise. The other possible discomfort is delayed onset of muscle soreness (DOMS) on the day following or subsequent two days after testing, however due to the low level of activity this is unlikely.

How will these discomforts and risks be alleviated?

You will have the opportunity to familiarise yourself with the testing procedures.

If you do not feel you are able to complete the testing requested, you should notify the researcher immediately and the testing will be terminated.

Finally, you should notify the researcher, if you have a current injury or have had an injury within the last two weeks that might affect your performance, or that might be worsened or aggravated by the required activity. For example, any strains and sprains must be reported, specifically to the hip, knee and ankle.

What are the benefits?

By participating in this study, you will receive specific information about your running gait/mechanics and the forces that you exert into the ground during running at various body weights. You will also improve our understanding of how unloading or reducing body weight relates to the biomechanics of running, which will help improve the practice of our physiotherapists, strength and conditioning specialists and coaches, particularly for rehabilitation.

What compensation is available for injury or negligence?

In the unlikely event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

How will my privacy be protected?

The identity and results of each participant will be kept confidential. However due to the small number of elite participants that are being recruited for this research, it is possible but unlikely that people will be able to identify who participants are when the results are published. Only my primary supervisor (Prof. Mike McGuigan) and I will have access to, and analyze your results

The results of your testing performance will only be given to your coach and/or physiotherapist (if applicable) with your permission (please check the appropriate box on the *Consent Form*).

What are the costs of participating in this research?

Costs to participation is minimal and only requires scheduling your time (~1 hour for each session) to be available for testing. Petrol vouchers will be provided to cover transport costs.

What opportunity do I have to consider this invitation?

A response to this invitation would be appreciated by no later than COB 28th February 2014

How do I agree to participate in this research?

If you would like to participate in this research, you need to sign the attached *Consent Form*, and return it to myself prior to participating in any of the tests.

If at any stage after volunteering, you do not wish to participate in this research, please notify me as soon as possible. You may withdraw at any time without any prejudice.

Will I receive feedback on the results of this research?

Yes, you can receive a summary of individual results once the information is ready for distribution (around one month after completing the study). Please check the appropriate box on the *Consent Form* if you would like this information.

What do I do if I have concerns about this research?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor, Prof. Mike McGuigan, michael.mcguigan@aut.ac.nz, or mobile 021 605 179

Concerns regarding the conduct of the research should be notified to the Executive Secretary of AUTEC, Kate O'Connor, *ethics@aut.ac.nz* , 921 9999 ext 6038.

Whom do I contact for further information about this research?

Researcher Contact Details:

Marissa Downes; email: marissa.downes@hpsnz.org.nz or cell 021 807 956

Project Supervisor Contact Details:

Supervisor, Prof. Mike McGuigan, michael.mcguigan@aut.ac.nz, mobile 021 605 179

Approved by the Auckland University of Technology Ethics Committee on ***type the date final ethics approval was granted*****,**
AUTEC Reference number ***type the reference number*****.**

Appendix 3: Participant Consent Form

Consent Form

Project title: **The effect of increasing body weight support on gait kinematics and tibial shock during treadmill running.**

Project Supervisor: **Prof. Mike McGuigan**

Researcher: **Marissa Downes**

- O I have read and understood the information provided about this research project in the Information Sheet dated dd mmmm yyyy.
- O I have had an opportunity to ask questions and to have them answered.
- O I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way.
- O I am not suffering from any injury, heart disease, high blood pressure, any respiratory condition (mild asthma excluded), any illness or any infection that will impair my physical performance (or that might be aggravated by the tasks requested).
- O I agree to take part in this research.
- O I wish to receive a copy of the report from the research (please tick one): YesO NoO
- O I agree for the research results to be released to my coach (please tick one): YesO NoO
- I agree for the research results to be released to my physiotherapist (please tick one):
- YesO NoO

Participant's signature: ....

Participant's name: ....

Participant's Contact Details (if appropriate):

……………………………………………………………………………………..

……………………………………………………………………………………..

……………………………………………………………………………………..

……………………………………………………………………………………..

Date:

***Approved by the Auckland University of Technology Ethics Committee on** type the date on which the final approval was granted **AUTEC Reference number** type the AUTEC reference number*

Note: The Participant should retain a copy of this form.

Consent and Release Form

Project title: **The effect of increasing body weight support on gait kinematics and tibial shock during treadmill running.**

Project Supervisor: **Prof. Mike McGuigan**

Researcher: **Marissa Downes**

- O I have read and understood the information provided about this research project in the Information Sheet dated dd mmmm yyyy.
- O I have had an opportunity to ask questions and to have them answered.
- O I understand that I may withdraw myself, my image, or any other information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way.
- O If I withdraw, I understand that all relevant information will be destroyed.
- O I permit the researcher to use the photographs that are part of this project and/or any drawings from them and any other reproductions or adaptations from them, either complete or in part, alone or in conjunction with any wording and/or drawings solely and exclusively for (a) the researchers literature, Masters thesis and examination purposes.
- O I understand that any copyright material created by the photographic sessions is deemed to be owned by the researcher and that I do not own copyright of any of the photographs.
- O I agree to take part in this research.

Participant's signature: ....

Participant's name: ....

Participant's Contact Details (if appropriate):

....

....

....

....

Date:

***Approved by the Auckland University of Technology Ethics Committee on** type the date on which the final approval was granted **AUTEC Reference number** type the AUTEC reference number*

Note: The Participant should retain a copy of this form.

Appendix 4: Recruitment Poster Advertisement

SPORTS PERFORMANCE
RESEARCH INSTITUTE, NEW ZEALAND
AN INSTITUTE OF AUT UNIVERSITY

ATTENTION ATHLETES

At the Sports Kinesiology, Injury Prevention and Performance Laboratory we are currently undertaking research into running technique and lower limb impact forces while running on a specialized treadmill which alters effective body weight, called the AlterG.
We will be looking at the changes in kinetic and kinematic variables at various body weight assisted loads during running. This will help us to identify biomechanical factors that are important during rehabilitation and return to sport running.

We are looking for highly trained male and female athletes between the ages of 18 and 35, who have represented their club or sport at a National level or higher in the past 12 months, who are injury free and have been training regularly without disruption from injury for the past 2 weeks.

As a part of this study, you will be assessed on three occasions. Each session will involve having wireless inertial sensors attached to the lower portion of the leg. You will also be filmed with high-speed 2-D cameras to capture gait kinematics, i.e, lower limb running technique.

Participants involved in this study will receive a fuel voucher and a brief report detailing their movement patterns during running.

If you are interested in taking part, or would like further information, please contact Marissa on 021 807 956 or email marissa.downes@hpsnz.org.nz

Appendix 5: Recruitment Email Advertisement

To: undisclosed recipients

Cc:

Subject: Research Participants Invited _ National Sporting Organisations

To the national directors and coaches,

I, Marissa Downes, am a Masters student based at the Sports Performance Research in New Zealand at AUT-Millennium Institute, Division of Sport and Recreation. Health and Environmental Sciences.

I would like to inform you of an upcoming research project which may interest you and your sporting organisation due to its relevance to running biomechanics and injury rehabilitation in elite athletes. We are extending invitations to elite level athletes who are healthy and uninjured. Participation is entirely voluntary and you or your athletes may withdraw at any time without any adverse consequences.

The project title is "The effect of increasing body weight support on gait kinematics and tibial shock during treadmill running"

The purpose of this study is to investigate the changes in running technique and ground reaction forces in the lower limb using the Alter-G treadmill, designed to unweigh the body during rehabilitation and for performance purposes.

Participants will be assessed on three occasions, in October and November. The first two sessions will be familiarization of the equipment and assessment protocol the final will be actual data collection. Each session will involve having a wireless inertial sensor attached to the lower portion of the leg (anterio-medial tibia) to collect tibial accelerometry data. Participants will be filmed with high-speed 2-D cameras to capture gait kinematics, ie, lower limb running technique. Prior to data collection a standardised warm-up protocol will be performed to minimise the risk of soreness or injury, and to ensure the data collected is reliable. Each testing session will take approximately 60-mins including the warm-up.

By participating in this study, participants will receive specific information about their running gait/mechanics and the forces that the exert into the ground during running at various body weight support levels. This will improve our understanding of how unloading or reducing body weight relates to the biomechanics of running, which will help improve the practice of our physiotherapists, strength and conditioming specialists and coaches, particulalry for rehabilitation. Participants will also receive a petrol voucher as compensation for their time. An individualised report can also be requested for your records and information.

There are no costs to participation, apart from scheduling your time (~1 hour for each session) to be available for testing.

Attached are full participant information and consent forms, including ethical approval details for your information.

If you believe this research to be beneficial to your athletes, team and sport, please feel free to pass this information on to the athletes for their pursual.

Any queries or concerns may be directed to myself or my primary supervisor, as outlined on the attached form.

Kind Regards

Marissa Downes
AUT University/SPRINZ

email: marissa.downes@hpsnz.org.nz
+64 21 807 956
skype: marissadownes

Appendix 6: Magnitude of Mechanistic Inferences Results for Tibial Shock

	100 v 60 %	**100 v-70 %**	**100 v 80 %**	**100 v 90 %**
Chance of statistically positive effect (%)	100	99	89	6
Chance of statistically positive effect (qualitative text)	Most Likely	Very Likely	Likely	Unlikely
Chance of statistically trivial effect (%)	0	1	11	94
Chance of statistically trivial effect (qualitative text)	Most unlikely	Very unlikely	Unlikely	Likely

Appendix 7: Repeated Measures Results Summary for Tibial Shock

Condition vs Control	**Change in mean (%)**	**Standardised Cohen's change in mean (%)**	**Cohen's effect size**	**Qualitative probable mechanistic inferences**	**Pearson correlation (r)**
100 v 60 %	17.7	0.71	Very Large	Most likely +	0.79
100 v 70 %	14.5	0.59	Large	Very likely +	0.91
100 v 80 %	8.4	0.35	Moderate	Likely +	0.92
100 v 90 %	2.0	0.09	Trivial	Likely trivial	0.95

Appendix 8: Tibial shock as units of gravity

	100 %	60 %	70 %	80 %	90 %
Mean (g)	16.4	13.9	14.3	15.1	16.1
SD+:	4.1	4.5	5.5	5.4	4.7
SD-:	3.3	3.4	4.0	4.0	3.6
Adjusted for +ve acceleration (g)	8.9	9.2	10.1	11	11.3

g = units of gravity; SD = standard deviation; +ve = positive

Appendix 9: Pearson Correlation Matrix

	SR60	SR70	SR80	SR90	SR100	IC60	IC70	IC80	IC90	IC100	MS60	MS70	MS80	MS90	MS100	TO60	TO70	TO80	TO90	TO100	tROM60	tROM70	tROM80	tROM90	tROM100
A60	-0.181	-0.264	-0.352	-0.173	-0.047	-0.095	-0.011	0.142	0.161	0.168	0.094	0.306	0.418	0.270	0.159	0.135	0.114	0.200	0.169	0.125	-0.016	-0.289	-0.255	-0.095	0.150
A70	0.031	-0.144	-0.289	-0.161	-0.021	-0.097	0.062	0.274	0.294	0.296	0.100	0.310	0.434	0.324	0.242	0.163	0.182	0.237	0.217	0.186	-0.131	-0.214	-0.169	-0.054	0.157
A80	0.002	-0.162	-0.272	-0.172	-0.060	-0.076	0.089	0.340	0.337	0.349	0.151	0.347	.449*	0.356	0.255	0.214	0.144	0.237	0.200	0.196	-0.161	-0.159	-0.083	0.003	0.263
A90	-0.003	-0.175	-0.291	-0.184	-0.068	-0.081	0.047	0.305	0.325	0.327	0.108	0.313	.451*	0.352	0.301	0.119	0.145	0.193	0.148	0.152	-0.121	-0.269	-0.178	-0.101	0.095
A100	-0.088	-0.218	-0.258	-0.082	0.000	-0.069	0.016	0.309	0.324	0.323	0.132	0.305	.447*	0.357	0.309	0.037	0.032	0.070	0.043	0.037	-0.188	-0.319	-0.206	-0.158	0.037
SR60		.880**	.550*	0.352	_0.401_	0.172	0.376	0.371	0.370	0.349	0.175	0.153	0.189	0.236	0.318	0.226	0.183	0.192	0.181	0.232	-0.100	0.234	0.146	0.184	-0.055
SR70			.692**	.568**	_.628**_	0.274	0.419	0.336	0.318	0.303	0.344	0.278	0.277	0.288	0.379	0.173	0.137	0.123	0.141	0.163	-0.050	0.209	0.097	0.194	-0.078
SR80				.786**	_.704**_	0.399	.488*	0.391	0.253	0.311	0.282	0.184	0.136	0.163	0.217	0.142	-0.001	-0.051	0.000	-0.029	0.160	0.377	0.271	0.231	0.132
SR90					_.936**_	0.301	0.397	0.286	0.160	0.205	0.438	0.362	0.392	0.295	0.308	0.204	0.025	-0.007	0.111	0.027	-0.073	0.117	-0.086	-0.015	-0.049
SR100						0.232	0.331	0.209	0.109	0.133	.474*	.449*	.516*	0.375	0.374	0.191	0.061	0.047	0.186	0.076	-0.118	-0.036	-0.276	-0.148	-0.196
IC60							.623**	.803**	.807**	_.808**_	.756**	.671**	.526*	.663**	.730**	0.028	0.041	0.005	-0.061	0.054	.605**	0.139	.500*	0.417	0.345
IC70								.763**	.603**	_.623**_	0.406	.531*	0.415	0.355	0.435	0.224	0.055	0.110	0.131	0.119	.520*	.562**	.502*	.569**	.445*
IC80									.955**	_.966**_	.685**	.719**	.622**	.720**	.762**	0.200	0.085	0.106	0.036	0.159	0.377	0.222	.550*	.496*	.478*
IC90										_.987**_	.708**	.701**	.611**	.765**	.814**	0.101	0.097	0.074	-0.024	0.143	0.324	0.056	.501*	0.441	0.393
IC100											.707**	.690**	.595**	.748**	.777**	0.175	0.133	0.111	0.017	0.172	0.307	0.106	.528*	.446*	.471*
MS60												.930**	.863**	.928**	_.884**_	0.294	0.188	0.245	0.206	0.325	0.081	-0.249	0.065	0.052	0.122
MS70													.959**	.939**	_.891**_	0.283	0.131	0.244	0.225	0.304	0.160	-0.238	0.003	0.052	0.119
MS80														.929**	_.874**_	0.311	0.160	0.283	0.278	0.330	0.015	-0.389	-0.215	-0.145	-0.059
MS90															_.956**_	0.238	0.142	0.226	0.170	0.302	0.032	-0.398	-0.032	-0.082	0.005
MS100																0.086	0.102	0.129	0.071	0.215	0.160	-0.305	0.061	0.015	-0.048
TO60																	.743**	.891**	.882**	_.872**_	-0.414	0.017	-0.119	-0.102	0.173
TO70																		.921**	.911**	_.916**_	-0.314	-0.041	-0.064	-0.041	0.045
TO80																			.971**	_.981**_	-0.369	-0.075	-0.158	-0.134	0.034
TO90																				_.955**_	-0.396	-0.044	-0.233	-0.168	-0.010
TO100																					-0.397	-0.095	-0.135	-0.121	0.015
tROM60																						0.433	.568**	.590**	_0.386_
tROM70																							.757**	.788**	_.677**_
tROM80																								.899**	_.800**_
tROM90																									_.800**_

. Correlation is significant at the 0.01 level (2-tailed).*. Correlation is significant at the 0.05 level (2-tailed). A = tibial shock. Ankle angle at IC = Initial Contact; MS = Midstance; TO = Toe Off. tROM = total loaded range of movement. SR = stride rate. Numerical values represent body weight support condition. **BOLD = significant correlation; *Italics = variable control for variable condition;* yellow highlight = significant correlations across variables