Tibial Acceleration and Shock Attenuation in Trained Female and Male Distance Runners at Different Levels of Body Weight Unloading

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ABSTRACT

Running popularity has led to a rise in chronic lower limb injuries resulting from cumulative loading. Many of these injuries are tibial stress fractures. Tibial accelerometers are commonly used to measure tibial stress and may even be predictive of injury at the distal limb. Lower body positive pressure (LBPP) treadmills have become increasingly popular amongst athletes and practitioners to prevent and treat lower limb injuries by reducing effective body weight (BW) through mechanical support. The purpose of this thesis is to investigate if BW unloading affects tibial acceleration (TA) and shock attenuation. Twelve trained distance runners (Sex: 6 males and 6 females; Age: 18-30 years) were recruited for this study. TA was measured through two Blue Trident, IMeasureU step units located at the distal tibiae. A STATSports Apex unit was also used to measure acceleration at the superior trunk and calculate shock attenuation for each limb. It was found that BW unloading had no discernable effect on mean peak TA and shock attenuation, bone stimulus, or contact time, regardless of running speed. However, a significant relationship was observed between running speed and both mean peak TA and bone stimulus where an increase in speed led to an increase in TA and bone stimulus. Furthermore, running speed did not affect shock attenuation or contact time. In conclusion, BW unloading did not alter gait kinematics in trained distance runners.

Tibial Acceleration and Shock Attenuation in Female and Male Distance Runners at Different Levels of Body Weight Unloading

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GENERAL AUDIENCE ABSTRACT

Running popularity has led to a rise in chronic lower limb injuries, particularly stress fractures at the shin or tibia bone, due to greater impact forces and “stiffer” landings. Tibial accelerometers are commonly used to measure these impact forces and may even be predictive of injury at the tibia bone near the ankle. The process of reducing these impact forces is called shock attenuation. Lower body positive pressure (LBPP) treadmills have become increasingly popular amongst athletes and practitioners to prevent and treat lower limb injuries by unloading body weight (BW) through mechanical support. The purpose of this thesis is to investigate if BW unloading affects tibial acceleration (TA) and shock attenuation. Twelve trained distance runners (Sex: 6 males and 6 females; Age: 18-30 years) were recruited for this study. TA was measured through two Blue Trident, IMeasureU step units located at the shin. A STATSports Apex unit was also used to measure impact at the upper trunk and calculate shock attenuation for each limb. It was found that BW unloading did not affect mean peak TA and shock attenuation, bone stimulus, or contact time, regardless of running speed. However, running speed significantly affected both mean peak TA and bone stimulus where an increase in speed led to an increase in TA and bone stimulus. Furthermore, running speed did not affect shock attenuation or contact time. In conclusion, BW unloading did not alter impact forces in trained distance runners. Caution is advised for individuals with injuries at the shin when using LBPP treadmills.

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CHAPTER 1: Introduction

The prevalence of injuries associated with running has prompted a need for innovative approaches to injury prevention and rehabilitation. Research has consistently shown an elevated injury rate among runners (Taunton et al., 2003), primarily chronic injuries resulting from cumulative loading from ground reaction forces (GRF) (Gent et al., 2007). Among the most common injuries are tibial stress fractures (Gent et al., 2007; Worp et al., 2015; Kahanov et al., 2015; Matheson et al., 1987; Warden et al., 2006). GRFs, which can be up to 2 to 3 times body weight (BW) at the distal tibiae during running (Sasimontonkul et al., 2007), are a significant external force contributing to these injuries. With each step, impact forces from the foot travel superiorly through the tibia bone to the trunk and head. Biomechanical factors may play a role in how these forces are attenuated throughout the body including lower limb kinematics, vertical loading rates, and impact forces (Reenalda et al., 2019).

Studies have demonstrated that runners with a history of stress fractures typically exhibit higher average vertical loading rates and stiffer landings compared to runners with no history of stress fractures (Milner et al., 2006, 2013 Moran et al., 2015 Grabowski et al., 2008 Jensen et al., 2016 Macias et al., 2012). However, measuring these impact forces practically and noninvasively in the field poses a challenge. The use of tibial accelerometers has become an increasingly popular tool and have been proven to be comparable surrogates for measuring tibial stress (Kiernan et al., 2018; Milner et al., 2006; Tenforde et al., 2019; Sheerin et al., 2019; Lafortune et al., 1995; Van den Berghe et al., 2019).

Moreover, they may be predictive of stress-related injuries (Milner et al., 2006; Tenforde et al., 2019).

Multiple factors influence tibial acceleration (TA) and shock attenuation including changes in running speed, stride length, foot strike, and footwear (Derrick et al., 1998, Mercer et al., 2002, Daoud et al., 2012; Hamil and Gruber, 2017; Shih et al., 2013). Few studies have investigated how hypogravity conditions affect TA and how those impact forces are attenuated.

Lower body positive pressure (LBPP) treadmills have become increasingly popular amongst runners, triathletes, soccer players, etc. as part of their rehabilitation program after injuries. LBPP treadmills unload effective BW through an upward lift provided by air pressure. These treadmills, designed to reduce impact forces during running, have been used by elite runners to build mileage while minimizing injury risk. Previous studies have shown that running in reduced gravity environments reduces vertical GRFs (Santos et al., 2023; Raffalt et al., 2013; Moran et al., 2015; Grabowski et al., 2008; Jensen et al., 2016; Macias et al., 2012). Thus, the use of LBPP treadmills may provide a solution by reducing these impact forces through BW support, ultimately contributing both to injury rehabilitation and prevention strategies.

Figure 1. Runner on Boost treadmill.

Although several studies have investigated TA at full BW, few have studied how running with a lower effective BW affects TA. Furthermore, there have been no studies to date that have compared how hypogravity affects how shock is attenuated throughout the body.

The objective of this study was to investigate the relationship between the level of BW unloading in the BOOST Microgravity treadmill and TA. Other gait kinematics that were analyzed was shock attenuation. Contact time and bone stimulus were also investigated. The outcome of this project aims to provide insight into how the LBPP treadmill can be used as a tool for rehabilitation or injury prevention.

Specific Aims

LBPP treadmills are designed to “unload” the lower limbs while running. Thus, they are used for injury rehabilitation when reduced impact forces on an injured limb is desired.

The overall objective of this study is to examine the extent to which these devices are effective. Specifically, this investigation will determine if running on a LBPP treadmill at different speeds will alter gait kinematics. Kinematics will be defined by several variables including TA, ground contact time, and shock attenuation.

The Specific Aims of this study are:

1. To determine if gait kinematics are altered by unloading due to hypogravity.
2. To determine if gait kinematics are altered by increasing running speeds.
3. To determine if the interaction of varying running speeds and unloading alter gait kinematics.

CHAPTER 2: Literature Review

The review of literature will begin with a description on how the hypogravity condition through LBPP treadmill technologies impacts running biomechanics. An examination of TA will be made, focusing on its significance to injuries, methodologies for measurement, and its alteration by LBPP treadmills. The purpose of this review is to explore what research has been done on how LBPP treadmills affect various biomechanical factors, specifically how hypogravity affects TA and shock attenuation.

LBPP Treadmills

The LBPP treadmill is a class of motorized treadmills that incorporate so-called "hypogravity" through means of a positively pressurized chamber surrounding the treadmill and the lower limbs of the individual using the treadmill. The athlete puts on a specially designed pair of neoprene shorts that allows them to zip into the chamber. When the participant enters the LBPP chamber, they adjust the height of the metal frame to rest at the height of their iliac crest for accurate unloading. Once the athlete is ready to run, an initial calibration process begins. The treadmill is equipped with a force plate that detects the participant's weight thus it can calculate how much differential air pressure to administer to effectively "unweight" the participant by the desired quantity. The athlete can set the LBPP treadmill from 100% BW, i.e. no hypogravity, down to 20% BW at 1% intervals and can run at a speed between .2 to 30 KPH. The grade of the treadmill can be adjusted as well (Heer et al., 2021).

When a runner initiates a training or rehabilitation program on the LBPP treadmill, it is expected that several changes to running motion and energy demand need to be considered. As described next, changes in spatiotemporal, muscle activity, kinematic, kinetic, and cardiorespiratory parameters occur with decreased BW.

Spatiotemporal

A change in the biomechanical environment from reduced BW leads to changes in running form including stride length, cadence, flight time, and CT. Researchers (Neal et al., 2016 & Santos 2023) have found that 80% BW or greater did not significantly alter the runner's form, whereas a BW of 60% or less led to significant alterations (Neal et al., 2016). One of these alterations is the increase in a runner's stride length (Sainton et al., 2015; Raffalt et al., 2013; Gojanovic et al., 2012; Jensen et al., 2016; Ueberschar et al., 2019; Moran et al., 2015). Specifically, participants ran with an additional 0.34 m longer stride length when at the 20% BW condition compared to 100% BW (Santos et al., 2023). In contrast, Mercer & Chona reported no effects of reducing BW on stride length between 20%, 30%, and 40% BW. Velocity also influenced stride length regardless of BW (Sainton et al., 2015; Raffalt et al., 2013; Gojanovic et al., 2012; Jensen et al., 2016; Ueberschar et al., 2019; Moran et al., 2015).

With increased speed and decreased BW there was an observed decrease in step frequency and contact time (Raffalt et al., 2013; Jensen et al., 2016; Santos et al., 2023; Neal et al., 2016; Thomson et al., 2021). Jensen et al. (2016) found a gradual decrease in step frequency at a running velocity of 2.22m/s. Step frequency decreased from 163 steps/min at 100% BW to 126 steps/min at 20% BW. A similar decline in step rate was found at

3.33m/s. Likewise, as BW decreased from 100% to 20%, contact time significantly decreased, except when comparing 80% to 100% BW (Santos et al., 2023). An increase in flight time occurred with a decrease in BW, regardless of velocity (Ueberschar et al., 2019; Neal et al., 2016; Raffalt et al., 2013; Sainton et al., 2015). This phenomenon could be attributed to the LBPP treadmill providing a vertical lift, reducing gravitational effect pulling the runner toward the ground.

Muscle Activity

The effect unweighting has on muscle activity is complex and the results are inconsistent (Jensen et al., 2016; Masumoto et al., 2017; Sainton et al., 2015). Researchers have found that a decrease in BW is associated with a reduction in muscle activity for the plantar flexors, rectus femoris, gastrocnemius, and biceps femoris muscles (Masumoto et al., 2017; Sainton et al., 2015), whereas the activity of the knee and ankle extensor muscles involved in braking did not change (Sainton et al., 2015). Jensen et al. (2016), had similar findings, yet they found a more pronounced decline in muscle activation for knee extensors compared to plantar flexors under BW unloading conditions. While hamstring muscle activity remained unchanged. They attributed these findings to differences in the roles that muscles play during running. The quadriceps muscle's main function is to resist gravity while the plantar flexors function to propel the body forward and provide support against gravity. The hamstrings' muscle activity patterns were unaffected largely due to their function during the leg swing phase and as stabilizers during the start of the stance phase.

Kinematics

Analyses of knee kinematics revealed that with decreased BW there was a reduction in peak knee flexion and knee range of motion (ROM) during the stance phase (Masumoto et al., 2017; Neal et al., 2016). At take-off, decreased BW significantly increased knee flexion while no significant observations were observed at initial contact. Increased velocity did significantly increase knee ROM. (Neal et al., 2016). The decrease in peak ROM during the stance phase may be attributed to a reduction in the required braking forces, likely caused by a decrease in ground reaction forces (GRF). With decreasing BW, there was a significant reduction in overall ankle range of motion, peak dorsiflexion, and time to peak dorsiflexion during the stance phase with increased plantar flexion at take-off. Velocity did not exert a significant effect on ankle kinematics throughout the stance phase (Neal et al., 2016).

Kinetics, Plantar, and Joint Loading

Experimental studies consistently reveal that decreasing BW decreases GRF (Santos et al., 2023) and knee and plantar loading (Raffalt et al., 2013; Moran et al., 2015; Grabowski et al., 2008; Macias et al., 2012; Jensen et al., 2016). Vertical impact peak and active peak GRF increased with increased running velocity and decreased almost linearly with weight support (Santos et al., 2023; Raffalt et al., 2013; Moran et al., 2015; Grabowski et al., 2008). However, it is important to note that the LBPP treadmill still provided 6% body weight support (BWS) at 0% with BWS being most accurate between 10-40% BWS (Heer et al., 2021; McNeill et al., 2015). Additionally, a BWS of 20% does not result in a 20% reduction in loading. For example, Santos et al. found that at 80% BWS, there was only a 26.7% reduction in impact. There were no significant changes in braking loads, indicating that there was little change to the horizontal component of loading (Santos et al., 2023).

Studies found that plantar load increased with increased running speed and reduced with decreased BW (Thomson et al., 2021) where the load tended to shift distally toward the forefoot (Smoliga et al., 2015; Santos et al., 2023; Neal et al., 2016). Neal et al. (2016) found that the forefoot and metatarsal heads were still under a substantial load of 500 N despite being at 20% BW. Knee force decreased linearly with unweighting (Macias et al., 2012) due to decreased knee extensor muscle forces and vertical GRF; however, the reduction in peak ankle joint forces were not as large as that seen in the knee due to greater muscular forces occurring in the plantar flexors (Jensen et al., 2016).

Increase Running Speed to Preserve Fitness

Unweighting affects metabolic and cardiorespiratory demands. Studies have found that time to exhaustion using the VO_2 max test was 34.5% longer on the LBPP treadmill with decreased BW compared to running at 100% BW (Raffalt et al., 2013). Time to exhaustion was also increased if the runner did not remain in the center of the treadmill because the treadmill provides assistive horizontal forces (Raffalt et al., 2013; Grabowski et al., 2008). Heart rate (Moran et al., 2015), VO_2, and ventilation are decreased with decreased BW (Raffalt et al., 2013; Fleckenstein et al., 2021; McNeill et al., 2015; Grabowski et al., 2008; Gojanovic et al., 2012; Ueberschar et al., 2019). The decrease in energy demand was not proportional to the percentage of hypogravity applied with 20% BWS causing a 34% reduction in net VO_2 whereas 40% BWS had a proportional reduction of 38% (McNeill et al., 2015). This was likely due to 0% BWS on the LBPP not truly being 0% because of measurable BWS (Heer et al., 2021; McNeill et al., 2015).

Researchers also found that it is possible to maintain an aerobic stimulus with BWS while reducing GRF. It is well understood that increasing running velocity leads to an increase in energy demand. Studies have investigated running velocity and BWS and how both can be used to reduce impact and maintain comparable aerobic stimuli to running unloaded (Raffalt et al., 2013; Fleckenstein et al., 2021; McNeill et al., 2015; Grabowski et al., 2008; Gojanovic et al., 2012; Ueberschar et al., 2019). Grabowski et al. (2008), suggest that an individual who runs at 3 m/s could achieve the same metabolic cost running at a speed of 5 m/s at 57% BWS while decreasing their active peak GRF by 32% but increasing their vertical loading rate by 30%. Further studies have investigated how manipulating incline can increase metabolic load under hypogravity conditions. Research has shown that running at a 7% incline at 80% BW can maintain energy demand and heart rate load as compared to running without BWS (Fleckenstein et al., 2021). Overall, from a cross-training perspective, the LBPP treadmill can reduce GRF while maintaining an aerobic stimulus.

LBPP and TA

Running is a popular activity, but the high participation rate has increased the incidence of repetitive use injuries (Taunton et al., 2003). The majority of these injuries are chronic in nature and are found in the lower limb due to cumulative loading resulting from ground reaction forces (GRF) (Gent et al., 2007). Tibial stress fractures are particularly widespread (Kahanov et al., 2015; Matheson et al., 1987) and the key mechanical risk factor in developing stress-related injuries is repetitive submaximal loading (Worp et al., 2015). Factors that contribute to tibial bone loading include the bone load intensity (magnitude and direction of load), rate of bone remodeling (length between bouts of exercise), and intrinsic factors (sex, age, bone density, mineral content, muscular activity, etc.) (Warden et al., 2006; Sasimontonkul et al., 2007; Matijevich et al., 2019). It has been estimated that muscle forces contribute 7 to 8 times the BW during running on the distal tibiae, while GRF's contribute an additional 2 to 3 times BW during running (Sasimontonkul et al., 2007). GRF does not necessarily reflect tibial bone loading because it does not consider the surrounding musculature which contributes to the majority of bone loading (Sasimontonkul et al., 2007; Matijevich et al., 2019; Zandebergen et al., 2022). However, GRF is the primary external force acting on the lower limb, and it is reasonable to assume that an increase in this force would lead to greater injury risk.

The ideal measurement of tibial bone loading would be to directly measure bone strain in-vivo to predict running injury, but this is invasive and impractical. Recently, the use of tibial accelerometers to measure lower extremity loading during running has become popular (Sheerin et al., 2019). Tibial accelerometers that are mounted on the bone directly are more accurate but require high expertise to use and are invasive. Skin-mounted accelerometers have been shown to be comparable surrogates for measuring impact forces (Lafortune et al., 1995; Sheerin et al., 2019). Research has shown that peak TA along the vertical axis has a strong correlation to vertical instantaneous loading rates and peak vertical GRF (Tenforde et al., 2019; Van den Berghe et al., 2019). Furthermore, researchers have shown that loading profiles and magnitudes quantified by accelerometers may be predictive of stress-related injuries (Kiernan et al., 2018; Milner et al., 2006; Tenforde et al., 2019).

Using accelerometers to measure TA has been a useful and valid measure of external loading, but there are some key factors that affect their reliability and accuracy including attachment, placement, and axes of acceleration measured. The literature indicates the device should have a low mass and be secured to the tibia as tight as tolerable to improve accuracy (Sheerin et al., 2019). Placement of the accelerometer along the shaft of the tibia changes the magnitude of peak TA and shock attenuation measured (Gabriel et al., 2016). Gabriel et al. (2016) found that substantial proximal placement of accelerometers (below the knee) resulted in lower TA compared to when it was placed distally near the medial malleolus. The importance of placement is further demonstrated when Xiang et al. (2022) used two tibial accelerometers, one placed in the distal half and the other in the proximal half of the tibia, to assess the correlation between shock attenuation and shoes with different cushioning. They found that shock attenuation was significantly greater in the distal tibia when the participant wore minimalist shoes compared to conventional shoes, but the shock had been attenuated in both cases by the time it reached the proximal tibia (Xiang et al., 2022). The placement of the accelerometer should be distally on the tibia near the malleoli to accurately represent peak acceleration and impact experienced at this common site of injury.

The axes the tibial accelerometer measures are also an important consideration when measuring forces experienced at the distal tibia. The tibia experiences shock along three locomotor planes: axial (TA-A), anterior-posterior (TA-AP), and medial-lateral (TA-ML). Uniaxial or triaxial accelerometers are employed in measuring TA, with a vast research body reporting on TA-A alone (Moran et al., 2017; Mercer & Chona, 2015; Milner et al., 2006; Sheerin et al., 2019) A large amount of data are neglected when uniaxial accelerometers are used over triaxial, which may result in important information being overlooked (Derie et al., 2020). For example, Lafortune (1995) found that at a running speed of 4.7 m/s TA-AP had the highest peak values (7.6 G) followed by TA-A (5.0 G) and TA-ML (4.5 G). Research also supports a significant correlation of TA-ML and loading rates thus supporting the significance of analyzing all three components of acceleration (Johnson et al., 2021). To quantify total TA passing through the musculoskeleton of the lower limb, it is important to measure all axes of acceleration.

There are multiple factors that contribute to changes in TA. TA is clearly influenced by stride length and cadence, where an increase in stride length results in an increase in impact loads (Derrick et al., 1998). Running velocity also plays a role, where an increase in running speed results in greater magnitude of impact (Mercer et al., 2002). How the foot makes contact with the ground can affect how GRF is transferred throughout the lower limb. Research has shown that runners with a rearfoot strike generate a rapid and large peak vertical GRF compared to forefoot strike runners (Daoud et al., 2012; Hamil and Gruber 2017; Shih et al., 2013). Footwear and surface also have substantial influence over TA and lower extremity stiffness (Xiang et al., 2022). Previous studies have shown that running on the LBPP treadmill reduces vertical GRF (Santos et al., 2023; Raffalt et al., 2013; Moran et al., 2015; Grabowski et al., 2008; Jensen et al., 2016; Macias et al., 2012); however, few studies have investigated how running under the hypogravity condition affects TA.

Researchers have found that tibial shock magnitudes were unchanged with BWS (Moran et al., 2017), or they did not change with the level of BWS (Mercer & Chona, 2015). Mercer & Chona (2015) found leg impact was greatest at 100% BW and was reduced at 40%, 30%, 20% BW, but the impact was not different between each BW condition. Leg impact acceleration was increased with an increase in running velocity despite BW.

Most studies have reported on only selective TAs with uniaxial accelerometers (Moran et al., 2017; Mercer & Chona, 2015). These uniaxial measurements of TA include only one component of TA (i.e. TA-A) while excluding essential information about the other two locomotor planes, i.e. TA-AP and TA-ML, which can have similar or greater peak values (Johnson et al., 2021; Lafortune, 1991). Few studies have investigated how hypogravity affects TA using triaxial accelerometers (Ueberschar et al., 2019).

Using triaxial accelerometers located at the distal tibia, Ueberschar et al. had 15 runners perform 3 incremental treadmill tests until volitional exhaustion at 60%, 80%, and 100% BW. They found that not only does hypogravity not affect TA at an absolute running speed it increases TA for a given relative running speed, i.e. a running speed that maintains a comparable physiological demand under a hypogravity condition. TA's at comparable physiological demands at 80% BW and 60% BW were on average +1.8 G and +2.5 G higher than at 100% BW. Triaxial peak TA at impact and active push-off was not reduced at a given absolute running speed which is mistakenly and yet commonly presumed by many coaches and athletes.

As stated before, TA is regarded as a valid surrogate to measure impact forces experienced at the distal tibiae (Lafortune et al., 1995 and absolute running speed, but it increases with running speeds that are physiologically comparable at standard gravity. Ueberschar et al. (2019) remind researchers and clinicians that it is the weight of the runner's body that is being changed in the LBPP treadmill and their mass (particularly that of the lower limb) remains unchanged. Ueberschar et al. further explain that when the tibia accelerates into the ground an inertial force acts in an equal and opposite direction. A greater acceleration will result in a greater mechanical force experienced at the tibia.

Summary

Running causes repetitive loading on the lower extremities, often resulting in various stress-related injuries, with the tibia being a common site of injury. Factors contributing to tibial injury include GRF, contractions of surrounding musculature, and rate of tissue remodeling. TA serves as a reliable proxy measurement for impact force experienced at the tibia. Several factors influence the tibia's capacity to absorb shock, including running velocity, stride length, and foot strike.

To address these concerns, hypogravity technologies, such as LBPP treadmills, have been employed for both rehabilitative and training purposes. LBPP treadmills, facilitated by BWS, have been shown to reduce GRF while preserving nearly normal biomechanics and physical fitness. However, there is limited data available regarding the effect hypogravity has on TA at various speeds and BW conditions. Additionally, there has been no research done on how impact is attenuated throughout the body under various BWs and speeds.

CHAPTER 3: Materials and Methods

Participants

Twelve well-trained distance runners (6 males, 6 females) were recruited for this study. All were between the ages of 18-30 years and were free from lower-extremity injury for at least six months prior to the study. The criterion for inclusion was participants were running a minimum of 30 miles per week and had trained for and competed in distance events during the past 3 years. Distance events were defined as running races of 3 miles or greater (e.g. 5 km and longer). Some participants had previous experience on the Boost treadmills (n= 6) while the other participants experienced the Boost treadmill for the first time during their warmup. All experimental procedures were approved by Virginia Tech Institutional Review Board (Appendix A). All participants gave their consent to participate in the study.

Treadmill Exercise Protocol

Participants completed a five-minute warm up at a self-selected speed on a Boost Microgravity treadmill with a motorized Woodway base (Waukesha, WI) at 0% incline and 100% BW. After the five-minute warmup, participants performed nine, two-minute running trials at different combinations of BW and running speed (refer to Table 3.1) during which TA data were collected. Each trial included a 60-second recovery at a self-selected pace, after which the treadmill was then reconfigured for the next trial. Three different treadmill speeds were used: Jogging (8.1 km/hr), Low speed running (10.9 km/hr), and Moderate speed running (13.8 km/hr). Three different hypogravity conditions were randomly assigned: 100% BW, 80% BW and 60% BW. The sequence of speed remained consistent, starting with jogging, then low speed running, and finally moderate speed running within each randomly chosen hypogravity condition. Speed remained consistent to reduce potential fatigue and resulting alterations in form. This resulted in just under 30 minutes of running (including the warm-up). Participants were asked to run as normally as possible and were given forewarning to changes in speed and BW. All participants wore their own running shoes and agreed to follow similar patterns of training before the experiment.

Table 3.1. Treadmill Exercise Protocol. Nine, 2-Minute Trials

		Load		
		60%	80%	100%
Pace (km/hr)	8.1	2 min	2 min	2 min
	10.9	2 min	2 min	2 min
	13.8	2 min	2 min	2 min

Equipment

To measure TA, the established and reputable Blue Trident, IMeasureU step (BT-IMU) unit (Auckland, New Zealand) was used. These accelerometers are small in dimensions (4 x 3x 1.5 cm) and lightweight (12grams) and operate wirelessly. The two BT-IMUs had a sufficient measurement range through two dual g compatible tri-axial accelerometers, one for smaller accelerations (+/- 16g) and a second for larger accelerations (+/- 200g). Accelerations were internally acquired at 1125hz and 1600hz respectively for each spatial axis. Participants wore two IMUs each attached noninvasively to the distal limb, just above the malleoli in accordance to the manufacturer recommendation. These tibial-mounted accelerometers were secured in place using a silicon strap.

To measure accelerations experienced at the superior trunk, participants wore a STATSports APEX unit (STATS group limited, Newry, United Kingdom). The wireless unit contained a18Hz GPS antenna, a 952 Hz triaxial accelerometer and gyroscope and a 10 Hz magnetometer. The unit's dimensions and weight were 8.4 x 4.4 x 2 cm and 48 grams, respectively. The APEX unit was attached to the back, between the scapulae, using a specially designed vest.

Following each session, the data from the BT-IMU and APEX devices were extracted using the respective manufacturer's software (IMU Step and SONRA). The sessions were divided into 9 separate speed and load combinations, and the raw data were subsequently exported to individual .csv files for additional analysis. All further analyses were conducted using custom-written MATLAB programs.

Signal Processing and Data Analyses

Gait kinematics were evaluated by several variables obtained from the manufacturers' software packages as well as bespoke analysis programs. These include:

STATSports:	Average Step Impact Left and Right	Average impact experienced by the left and right foot, respectively
IMeasureU:	Right and Left Peak Acceleration	Average peak tibial acceleration experienced by the left and right foot, respectively
	Right and Left Contact Time	The time between heel strike and toe off of an individual step
	Left and Right Bone Stimulus	A proprietary algorithm that estimates the growth stimulus provided during running for the left and right tibia bones
Both:	Shock Attenuation	The relative reduction in trunk acceleration (APEX) compared to TA (IMeasureU).

Signal Alignment. The clocks in the IMU and APEX units use different signals to track time: the IMU relies on the phone clock (Bluetooth initiated), while the APEX synchronizes with GPS satellites. Each .csv data file includes timestamps (in UNIX epoch time) and tri-axial accelerometer and gyroscope data for each sample. As described by Williams et al. (2022), temporal alignment of accelerometer signals was achieved by applying controlled movements to all three sensors simultaneously. Calibration movements were conducted before and after each session. Cross-correlation analysis was then applied to these data to determine the time delay (TD) between the IMU and APEX sensors. This TD was then used to shift and synchronize the temporal data for the remainder of the session. Accelerometer and gyroscope data from the three devices were then resampled to 1000 Hz and filtered at 40 Hz using a low-pass, fourth-order Butterworth filter. The synchronized and filtered data were then used to calculate the additional kinematic variables.

Following the above alignment procedures, the triaxial acceleration data were converted into resultant acceleration (a_r) using the relationship: $a_r^2 = a_x^2 + a_y^2 + a_z^2$, where a_x, a_y and a_z represent accelerations in the three locomotor planes: anterior-posterior, axial, and medial-lateral, respectively. The a_r data were used for the remaining variable calculations.

Contact Time: Foot strike (FS) and toe off (TO) were identified using previous studies (Aminian et al., 2002; Lee et al., 2011; McGrath et al., 2012) by analyzing IMU gyroscope data adjusted with the TD (refer to Figure 3.1). From the pitch gyroscope data (mediolateral axis), peak angular velocity during the swing phase, known as mid-swing

(MS), was determined. Starting from MS, the local minima were identified on both sides, with the one preceding MS designated as TO of the previous stance phase and the one following MS as FS of the subsequent stance phase. The interval of time between consecutive FS and TO values was defined as contact time.

Shock Attenuation: During each stance phase (or contact time interval), trunk accelerations are attenuated compared to ankle accelerations. Peak attenuation for each limb was calculated using peak trunk a_r (a_{APEX}) and ankle a_r ($a_{BT\text{-}IMU}$) during the stance phase, using the equation below from Dufek et al. (2009) and Reenalda et al. (2019). An increase in peak attenuation indicates increased dampening of the wave of acceleration traveling from the ankle to the trunk.

$$Peak\ Attenuation\ (\%) = 100 \times\ [1 - (a_{APEX}/a_{BT-IMU})]$$

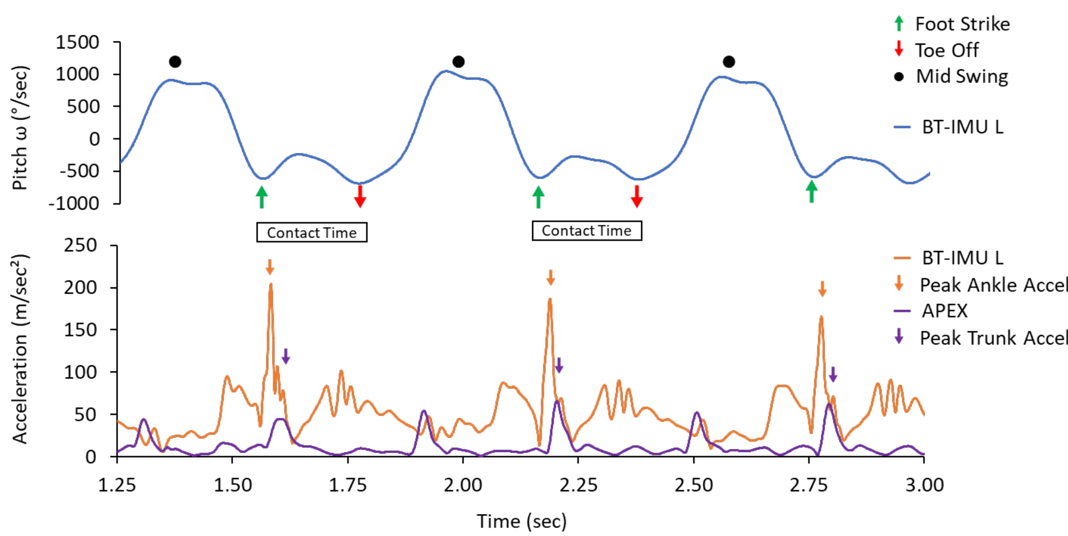

Figure 3.1. Determination of step contact time and peak accelerations used for the attenuation calculations. Shown is data from two left limb steps. Figure obtained from Williams et al. (2022)

Bone Stimulus: The IMU-step software automatically calculates the bone stimulus metric for each session using an estimate of the mechanical stimulus required for bone response and remodeling. The IMU-step manufacturers define these mechanical factors as the magnitude of impact from each step (peak TA) and the frequency of loading cycles, emphasizing that bone responds more to the magnitude of impact loads than to the number of cycles. The manufacturers have calculated the bone stimulus metric, known as daily load stimulus (DLS, Figure 3.2), by integrating these two variables for each training session. However, it is important to note that this metric is not representative of total bone load stimulus because accelerometers do not measure muscular contributions to bone load.

$$DLS = \left[\sum_{j=1}^{k} \overbrace{n_j}^{\text{number of cycles}} \underbrace{(\sigma_j)}_{\substack{\text{mechanical stress} \\ \text{(peak tibial acceleration)}}}^{\overbrace{m}^{\text{empirical constant}}} \right]_{\text{per day}}^{\overbrace{\frac{1}{2m}}^{\text{empirical constant}}}$$

Figure 3.2. ImeasureU step's calculation for bone stimulus or Daily Load Stimulus (DLS) where n_j represents the number of loading cycles and σ represents peak tibial acceleration from each step. Figure obtained from https://support.imeasureu.com/article/21-what-is-bone-stimulus

Statistical Analyses

Data were analyzed using standard statistical analyses (i.e. descriptive statistics). A two-way analysis of variance was employed and adjusted for repeated measures made on each participant to compare the effects of running speed and unloading as well as their interactions. A Tukey's post-hoc exam was used, when appropriate, to detect individual differences. This was done on each dependent variable. Effect sizes were based on partial eta (ηp^2), with benchmarks of 0.01 indicating a small effect size, 0.06 indicating a medium effect size, and 0.14 indicating a large effect size (Hopkins et al., 2009; Cohen, 1988).

CHAPTER 4: Results

There was no significant relationship found between BW and TA regardless of running speed ($p = 0.389$, $\eta_p^2=0.082$; medium effect). There was a significant relationship found between running speed and mean peak TA ($p = 0.00$, $\eta_p^2=0.889$; large effect) where an increase in running speed led to an increase in mean peak TA (Figure 4.1). TA was not influenced by the interaction in running speed and BW ($p = 0.532$, $\eta_p^2=0.068$; medium effect).

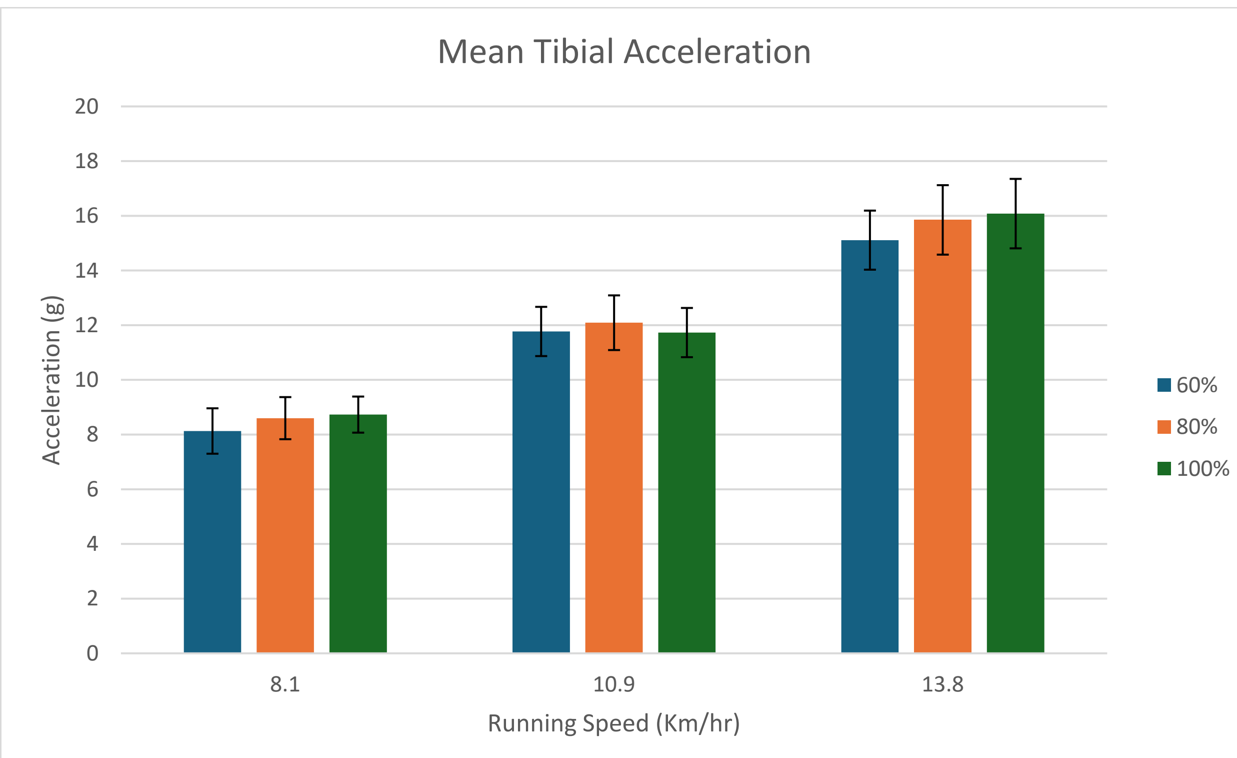

Figure 4.1. Mean peak tibial acceleration (g) at 3 different running speeds (km/hr) and body weight conditions (60%, 80%, and 100% body weight).

Cumulative bone load stimulus remained unaffected by BW unloading ($p = 0.331$, $\eta p^2=0.096$; medium effect), while speed had a significant effect on mean bone stimulus ($p = 0.000$, $\eta p^2=0.851$; large effect), where an increase in speed led to an increase in bone stimulus, irrespective of BW unloading.

Figure 4.2. Cumulative bone stimulus (au) as calculated by IMU Step dependent on total number of steps taken and magnitude of tibial acceleration at 3 different speeds (km/hr) and body weight conditions (60%, 80%, 100% body weight).

Concerning shock attenuation, neither speed ($p = 0.084$, $\eta p^2=0.202$; large effect) nor BW ($p = 0.091$, $\eta p^2=0.195$; large effect) significantly influenced mean shock attenuation (Figure 4.3).

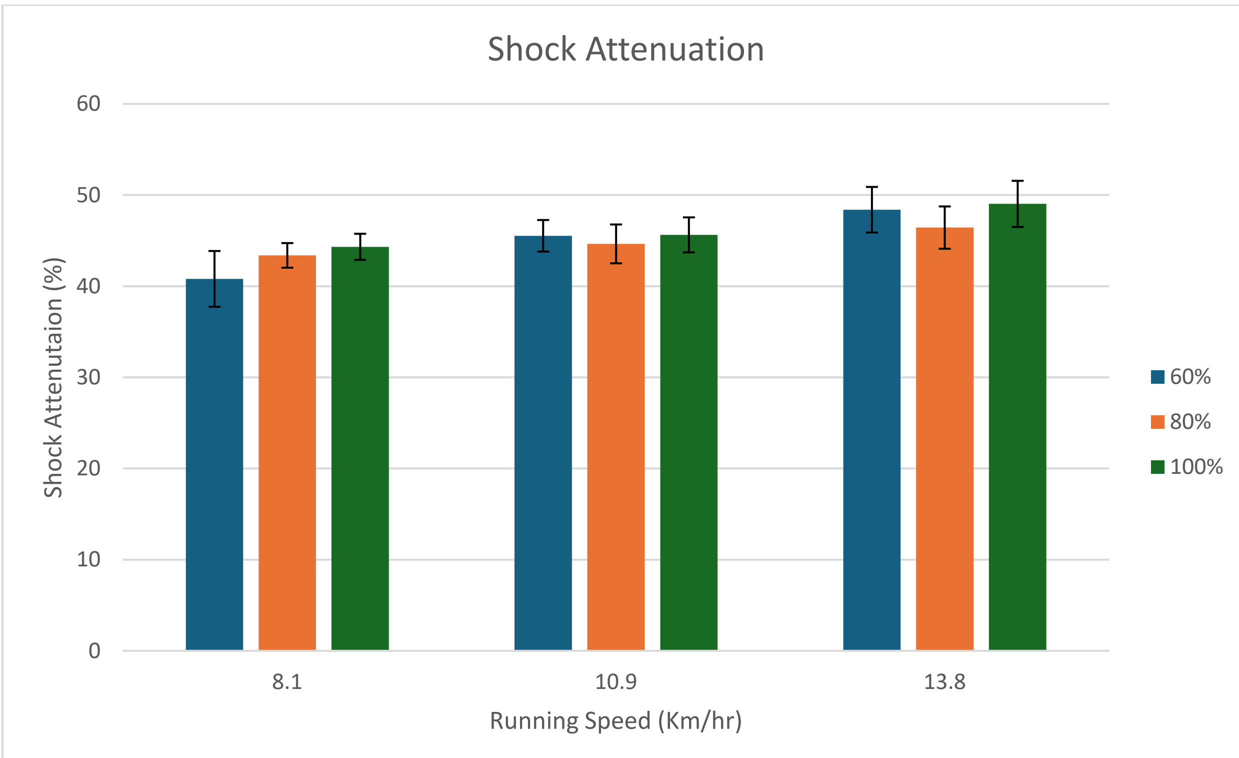

Figure 4.3. Shock attenuation (%) between the tibia and trunk at 3 different running speeds (km/hr) and body weight conditions (60%, 80%, and 100% body weight).

Contact time was unaltered by running speed ($p = 0.264$, $\eta p^2=0.114$; medium effect) and BW unloading ($p = 0.073$, $\eta p^2=0.360$; large effect) (Figure 4.4). Additionally, there was no significant interaction between running speed and BW on contact time ($p = 0.069$, $\eta p^2=0.176$; large effect).

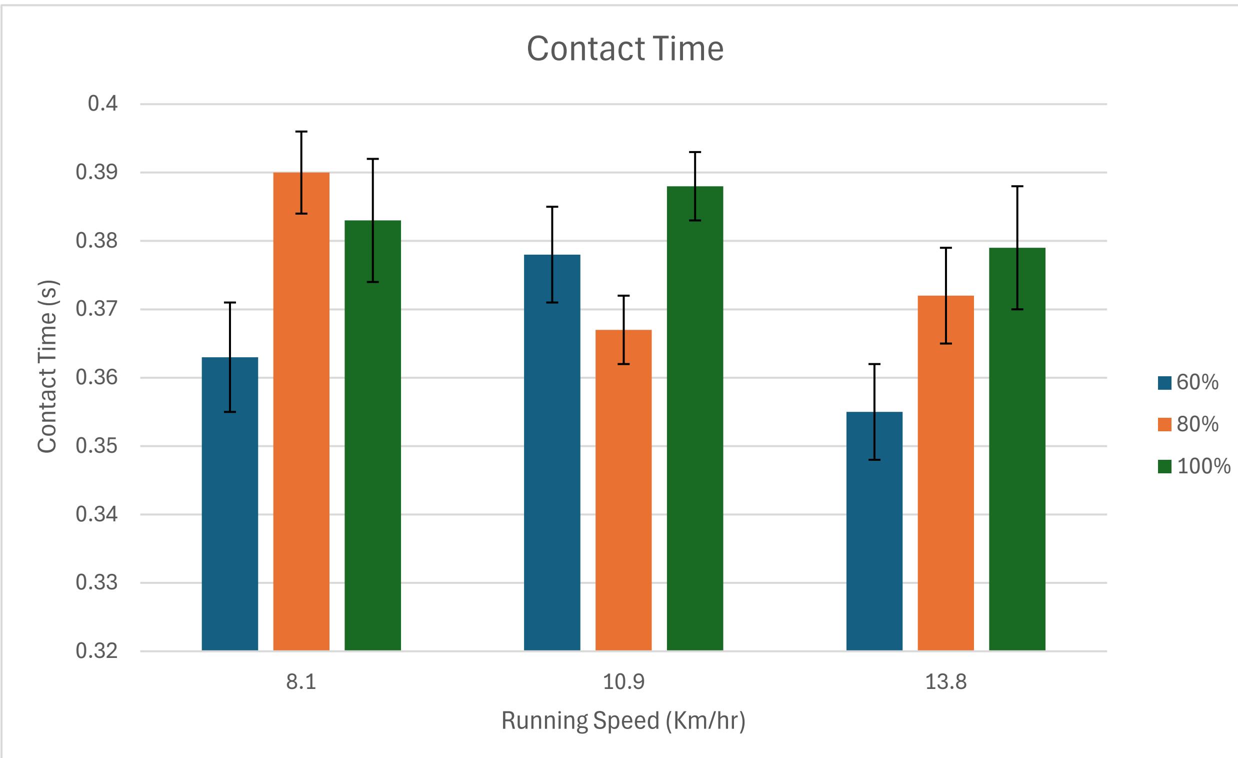

Figure 4.4. Contact time (sec) 3 different running speeds (km/hr) and body weight conditions (60%, 80%, and 100% body weight).

CHAPTER 5: Discussion and Conclusion

Discussion

The primary aim of this study was to examine the relationship between the level of BW unloading, TA, and shock attenuation. What was found aligned with previous research (Ueberschar et al., 2019) which was that unloading BW through hypogravity does not reduce 3D mean peak TA at impact, regardless of running speed. There was no discernable difference between different levels of BW unloading. Furthermore, running speed was found to have a significant influence on TA, with higher speeds resulting in increased TA, irrespective of the level of hypogravity (Figure 4.1). As anticipated due to the lack of change in TA, the bone stimulus remained consistent across different levels of BW support and increased with higher running speeds (Figure 4.2).

There was also no significant relationship found between shock attenuation, running speed, and BW. Shock attenuation remained consistent despite changes in running speed (Figure 4.3). This implies that while TA increased with increasing running speed, the acceleration experienced at the trunk also increased proportionally. Nonetheless, the magnitude of shock being attenuated between the tibia and trunk remained constant. This demonstrates the body's capacity to mitigate impact forces despite variations in running speed.

Contact time was unaltered by running speed and BW unloading (Figure 4.4). This observation was not in agreement with previous findings (Raffalt et al., 2013 Santos et al., 2023 Thomson et al., 2021, Sainton et al., 2015, Ueberschar et al., 2019). Although there was not a statistical difference, the effect size was large. The differences between the present study and previous works could be due to several factors including methods used to detect contact time (accelerometry vs ground contact), sensor placement (shoe mounted vs tibial mounted), and the treadmill used (Boost® vs AlterG®).

TA is commonly regarded as a valid proxy for impact forces (Kiernan et al., 2018; Milner et al., 2006; Tenforde et al., 2019; Sheerin et al., 2019; Lafortune et al., 1995; Van den Berghe et al., 2019). The findings in this study suggest that BW unloading does not reduce tibial stress from impact forces at an absolute running speed, but it actually increases with faster running speeds. This observation holds crucial implications for coaches, athletes, and practitioners who utilize LBPP treadmills for injury prevention and rehabilitation. It is often mistakenly assumed that using LBPP treadmills can reduce tibial load from impact forces during intense training sessions. However, this study demonstrates that these assumptions may not be entirely true. To maintain an equivalent physiological effort to running without BW unloading while running on an LBPP treadmill with BW unloading, there must be an increase in running speed (Raffalt et al., 2013 Fleckenstein et al., 2021 McNeill et al., 2015 Grabowski et al., 2008 Gojanovic et al., 2012 Ueberschar et al., 2019). This leads to an increase in stress experienced at the distal tibia despite BW unloading.

Nevertheless, there may be evidence to suggest that BW unloading could decrease osseous stress. GRFs are reduced with BW unloading, likewise there is evidence that the muscular activity of the plantar flexors surrounding the distal tibia do have a partial reduction in activity (Jensen et al., 2016), as these muscles also play a role in supporting

BW against gravity. As mentioned previously, muscles contribute a large proportion to the stress experienced at the distal tibia (Sasimontonkul et al., 2007; Matijevich et al., 2019; Zandebergen et al., 2022), and a partial reduction in this force from BW unloading could lead to lessened osseous stress. Although TA is not reduced, there might still be less osseous load from surrounding musculature. Further research would be needed to explore the muscular contribution to tibial stress and injury, as well as how muscle activity is affected by BW unloading.

In summary, individuals with an injury at the distal tibia should exercise caution when using LBPP treadmills for injury rehabilitation and training. If they were to use the treadmill, running speeds faster than what they normally run on land should be avoided to prevent greater stress on the tibia bone; however, this is not to say that the treadmill may not be beneficial for other injuries, especially ones occurring closer to the center of gravity where BW unloading will have a greater effect on muscular and impact forces. It is important to interpret these findings tentatively since the study did not include injured runners. Therefore, readers should consider additional research or individualized assessments before applying them to specific rehabilitation or training scenarios.

To determine why TA was unaffected by hypogravity it is important to keep in mind what forces are acting on the body during running. When an individual runs on a LBPP treadmill, their effective BW is being partially counteracted by an upward positive pressure force and the body's mass remains constant. The forces in the horizontal direction generated by muscles to accelerate the lower limb's mass and maintain running speed remains constant at varying levels of BW. This is because the muscles still need to overcome inertia and move the legs forward at a given speed, regardless of BW. This is reinforced by Jensen et al.'s (2016) findings where the activity of muscles that play an active role in supporting the body's weight against gravity were reduced with unloading while the activity of muscles involved in propulsion and leg swing during running were unaffected.

Furthermore, it is necessary to look at the changes in biomechanics that occur with increased BW unloading. Experimental studies have consistently shown that stride length and flight time are increased, and step frequency and contact time are decreased with an increase in BW unloading (Raffalt et al., 2013 Santos et al., 2023 Thomson et al., 2021, Sainton et al., 2015, Ueberschar et al., 2019). Flatter leaps and higher horizontal and lower vertical impact/push off peaks have been observed because of these adaptations (Polet et al. 2017). This demonstrates the body's ability to adapt to run in the most energy efficient manner where vertical work is lower due to reduced gravity pull and horizontal work is greater to maintain forward momentum.

Limitations and Strengths

The current study's findings regarding BW unloading and its impact on gait kinematics should be interpreted tentatively, given the absence of injured runners in the sample. Previous studies have indicated that different foot strike patterns (Daoud et al., 2012; Hamil and Gruber 2017; Shih et al., 2013) and running with stiffer landings (Milner et al., 2006) can elevate TA, yet these factors were neither controlled for nor was it measured. However, the study's strengths lie in its utilization of each participant as its own control, enabling comparisons of speed and BW across trials. Additionally, the study benefited from including well-trained distance runners who had the ability to adapt to different speeds without excessive fatigue preventing greater accelerations throughout the duration of the study.

Conclusion

LBPP treadmills are a new device increasingly utilized by athletes and practitioners for training, injury prevention, and rehabilitation. However, this study indicates that BW unloading has no effect on TA, while speed significantly alters TA. Additionally, shock attenuation was not affected by BW or a change in running speed. These findings suggest that LBPP treadmills may not reduce tibial bone stress and should be used with caution for injury rehabilitation and injury prevention at the distal tibiae. Future studies should explore the effect of BW unloading on muscular forces and how those forces, in conjunction with TA, contribute to tibial bone loading.

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APPENDIX A: Institutional Review Board Approval

Division of Scholarly Integrity and
Research Compliance
Institutional Review Board
North End Center, Suite 4120 (MC 0497)
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MEMORANDUM

DATE: February 15, 2024

TO: Jay H Williams

FROM: Virginia Tech Institutional Review Board (FWA00000572)

PROTOCOL TITLE: Gait Kinematics During Running on a Hypogravity Treadmill

IRB NUMBER: 24-293

Effective March 15, 2024, the Virginia Tech Institution Review Board (IRB) approved the New Application request for the above-mentioned research protocol.

This approval provides permission to begin the human subject activities outlined in the IRB-approved protocol and supporting documents.

Plans to deviate from the approved protocol and/or supporting documents must be submitted to the IRB as an amendment request and approved by the IRB prior to the implementation of any changes, regardless of how minor, except where necessary to eliminate apparent immediate hazards to the subjects. Report within 5 business days to the IRB any injuries or other unanticipated or adverse events involving risks or harms to human research subjects or others.

All investigators (listed above) are required to comply with the researcher requirements outlined at:

https://secure.research.vt.edu/external/irb/responsibilities.htm

(Please review responsibilities before beginning your research.)

PROTOCOL INFORMATION:

Approved As:	**Expedited, under 45 CFR 46.110 category(ies)**
Protocol Approval Date:	**February 15, 2024**
Progress Review Date:	**February 15, 2025**

ASSOCIATED FUNDING:

The table on the following page indicates whether grant proposals are related to this protocol, and which of the listed proposals, if any, have been compared to this protocol, if required.

Invent the Future

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APPENDIX B: Analysis of Variance Tables

Tibial Acceleration (m/s²)

				Alpha	**0.05**	
	SS	***df***	***MS***	***F***	***P value***	***P Eta-sq***
Speed	933.315	2	466.658	88.486	**0.000**	0.889
Error	116.024	22	5.274			
Load	6.273	2	3.137	0.987	0.388	0.082
Error	69.886	22	3.177			
Speed x Load	3.210	4	0.802	0.800	0.532	0.068
Error	44.120	44	1.003			
Participant	918.585	11	83.508			
Total	2091.413	107	19.546			

Power (based on effect size) = 0.988

Means Comparisons

Speed	*P value*
5.0 vs 6.8	**4.972E-07**
5.0 vs 8.6	**5.385E-07**
6.8 vs 8.6	**2.486E-06**

Shock Attenuation (%)

				Alpha	0.05	
	SS	*df*	*MS*	*F*	*P value*	*P Eta-sq*
Speed	471.557	2	235.778	2.778	0.084	0.202
Error	1867.163	22	84.871			
Load	51.512	2	25.756	2.672	0.091	0.195
Error	212.058	22	9.639			
Speed x Load	78.920	4	19.730	1.076	0.380	0.089
Error	806.897	44	18.339			
Participant	2728.068	11	248.006			
Total	6216.176	107	58.095			

Power (based on effect size) = 0.925

Contact Time (sec)

				Alpha	0.05	
	SS	*df*	*MS*	*F*	*P value*	*P Eta-sq*
Speed	0.002	2	0.001	1.415	0.264	0.114
Error	0.017	22	0.001			
Load	0.006	2	0.003	6.195	0.073	0.360
Error	0.010	22	0.000			
Speed x Load	0.005	4	0.001	2.346	0.069	0.176
Error	0.023	44	0.001			
Participant	0.012	11	0.001			
Total	0.074	107	0.001			

Power (based on effect size) = 0.982

Bone Stimulus (AU)

				Alpha	0.05	
	SS	***df***	***MS***	***F***	***P value***	***P Eta-sq***
Speed	43795.518	2	21897.759	62.963	0.000	0.851
Error	7651.368	22	347.789			
Load	4724.383	2	2362.191	1.164	0.331	0.096
Error	44661.439	22	2030.065			
Speed x Load	47.758	4	11.940	0.148	0.963	0.013
Error	3561.124	44	80.935			
Participant	24337.971	11	2212.543			
Total	128779.561	107	1203.547			

Power (based on effect size) = 0.965

Means Comparisons

Speed	*P value*
5.0 vs 6.8	**0.002**
5.0 vs 8.6	**2.383E-09**
6.8 vs 8.6	**0.027**

Mean ± Standard Error of Mean Values

Load	60%		
Speed	**5 mph**	**6.8 mph**	**8.6 mph**
Tibial Acceleration (m/s^2)	8.13 ± 0.83	11.77 ± 0.90	15.11 ± 1.08
Bone Stimulus (AU)	204.95 ± 12.85	215.23 ± 10.65	226.89 ± 9.34
Shock Attenuation (%)	40.80 ± 3.06	45.53 ± 1.73	48.39 ± 2.50
Contact Time (sec)	0.370 ± 0.009	0.378 ± 0.007	0.371 ± 0.007

Load	80%		
Speed	**5 mph**	**6.8 mph**	**8.6 mph**
Tibial Acceleration (m/s^2)	8.60 ± 0.77	12.09 ± 1.00	15.85 ± 1.27
Bone Stimulus (AU)	204.49 ± 10.95	214.61 ± 8.98	227.60 ± 7.97
Shock Attenuation (%)	43.38 ± 1.35	44.64 ± 2.13	46.43 ± 2.32
Contact Time (sec)	0.390 ± 0.006	0.367 ± 0.005	0.367 ± 0.006

Load	100%		
Speed	**5 mph**	**6.8 mph**	**8.6 mph**
Tibial Acceleration (m/s^2)	8.73 ± 0.66	11.73 ± 0.90	16.08 ± 1.27
Bone Stimulus (AU)	200.86 ± 10.77	211.16 ± 9.33	226.51 ± 8.04
Shock Attenuation (%)	44.32 ± 1.43	45.63 ± 1.92	49.03 ± 2.53
Contact Time (sec)	0.383 ± 0.009	0.388 ± 0.005	0.379 ± 0.009

APPENDIX C: Informed Consent Form

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Consent to Take Part in a Research Study

Title of research study: Gait Kinematics During Running on a Hypogravity Treadmill
IRB #24-293

Principal Investigator: Jay H. Williams, Department of Human Nutrition, Foods and Exercise

Key Information:

The following is a short summary of this study to help you decide whether or not to be a part of this study. More detailed information is listed later on in this form. Hypogravity treadmills are designed to "unload" the lower limbs while running. Thus, they are used for injury rehabilitation when reduced impact forces on an injured limb is desired. The overall objective if this study is to examine the extent to which these devices are effective. Specifically, this investigation will determine if running on a hypogravity treadmill at different speeds will alter gait kinematics. The results will help understand how hypogravity treadmills impact injury rehabilitation.

Why am I being invited to take part in a research study?

We invite you to participate in this study because you are a trained, recreational runner, someone who runs more than 20 miles per week and participates in road race competitions. Also, you are between the ages of 18 and 30 years. Because we are interested in studying recreationally trained runners, you are an ideal candidate for this study. Pregnant individuals and those suffering from illness or injury that would limit running are excluded from the study.

What should I know about being in a research study?

- Someone will explain this research study to you
- Whether or not you take part is up to you
- You can choose not to take part
- You can agree to take part and later change your mind
- Your decision will not be held against you
- You can ask all the questions you want before you decide

Why is this research being done?

It is important to understand how the running gait may be altered when running on a hypogravity treadmill. Because these treadmills are designed to "unload" the limbs, changes in loading and speed could affect how you run. It is important to understand the relationships between these two variables as artificially altering the gait could hinder rehabilitation.

How long will the research last and what will I need to do?

We expect that your participation in this research will last 2, non-consecutive days, an orientation session and a testing session.

During the orientation session, the testing protocol will be fully explained as will the use of the treadmill. You will also have the opportunity run comfortably on the anti-gravity treadmill at varying