Gait re-training: a technology-based intervention for reducing impact loading in running.

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Declaration

I hereby certify that this material, which I now submit for assessment on the programme of study leading to the award of Doctor of Philosophy is entirely my own work, and that I have exercised reasonable care to ensure that the work is original, and does not to the best of my knowledge breach any law of copyright, and has not been taken from the work of others save and to the extent that such work has been cited and acknowledged within the text of my work.

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List of abbreviations

AVLR- Average vertical loading rate

COL- Cost of locomotion

COT-Cost of transport

COM- Centre of mass

EE- Energy expenditure

F- Force

FFS- Forefoot strike

HR- heart rate

ITBS- Iliotibial band syndrome

IVLR- Instantaneous vertical loading rate

KAM- Knee adduction moment

Kcal- Kilo calories

Km/hr- Kilometres per hour

M- Mass

Mins- Minutes

MCV- Maximal Voluntary Contraction

MFS- Mid foot strike

PFP- Patella femoral pain

PFPS- Patella femoral pain syndrome

PPA-Peak positive acceleration

RE- Running Economy

RPE- Rate of perceived exertion

t- Time

v- Velocity

VIP- Vertical impact peak

vGRF- Vertical ground reaction force

List of presentations at conferences

- Gait retraining to reduce loading: what is the ideal location for providing accelerometer based biofeedback (European Conference of Sports Science, 2014, Amsterdam)
- The effect of compliant running on impact accelerations and energy expenditure (ISBS, 2013)
- The effect of compliant running on kinematics and joint kinetics (ISBS, 2013)

Abstract

Gait re-training: a technology-based intervention for reducing impact loading in running.

Ciarán Padráig Ó Catháin

Purpose: The primary aim of this thesis was to examine if a compliant running technique reduces impact accelerations, what the associated kinematics and kinetics are, and what method should be employed in the teaching of runners to adopt this technique. A secondary aim was to determine the effect of compliant running kinematics on energy expenditure.

Methods: Study 1 examined the use of a verbally directed compliant technique. Study 2 examined the success of an accelerometer-based biofeedback system using various accelerometer locations (tibia, sacrum and treadmill). Study 3 then compared the use of a tibial located accelerometer-based biofeedback system to verbal feedback. Study 4 examined a 4-week treadmill based biofeedback intervention with regard to how it altered kinematics, loading, and energy expenditure.

Results: Treadmill accelerometer-based biofeedback appears to display an ability to reduce both tibial (-26%) and sacral (-17%) accelerations acutely, with reductions increasing further when the intervention period is extended to 4-weeks (tibia: -40%; sacrum: -42%). These reductions were associated with reduced vGRFs and joint moments. In comparison, an acute and a 3-week bout of verbal feedback produced lower reductions in impact accelerations (tibia: -8%, sacrum: -22%; tibia: -10%, sacrum: -41%;); while segment based biofeedback produced large reductions but more localised to their source of feedback: sacral biofeedback (tibia: -1%; sacrum: -27%), and tibial biofeedback (tibia: -39%; sacrum: -4%). This reduced loading was achieved by increased cushioning at impact and decreased vertical oscillation of the COM. These kinematic changes demonstrated no effect on energy expenditure in study 4, but an increase in energy expenditure in study 1; possible due to the larger degree of knee and hip flexion in study 1.

Conclusion: A 4-week treadmill accelerometer-based biofeedback intervention appears to reduce loading to a greater extent than verbal feedback, or biofeedback from the tibia or sacrum. This appeared to not influence energy expenditure.

Chapter 1: Introduction

1.1 Introduction

Running is one of the most popular forms of physical activity worldwide (Voloshin, Mizrahi, Verbitsky, and Isakov, 1998). In Ireland, the popularity of running as a form of physical activity has had a greater than 2 fold increase from 2007-2011 (Irish sports council, 2011) and recent data from the USA suggests that those regularly participating in running as a physical activity has increased by 10% between 2010 and 2012 reaching in excess of 35 million (Rothschild, 2012a). It is well established that physical activity has important cardiovascular and musculoskeletal health benefits (Warburton et al, 2006). With a worldwide inactivity epidemic largely contributing to morbidity and mortality, increasing physical activity levels is essential. Globally, six percent of deaths are attributed to physical inactivity, accounting for 3.2 million deaths annually, as well as being implicated as the main cause for approximately 21–25% of breast and colon cancers, 27% of diabetes and 30% of ischaemic heart disease (World Health Organization, 2013). Apart from causing major health problems at an individual level, this also generates huge costs for health organizations and governments. In the USA and Australia it has been shown that billions of dollars are lost yearly as a result of physical inactivity and its detrimental effect on health (Chenoweth, 2005; Cadhillac et al, 2011). It is clear that limiting existing barriers to physical activity, in order to increase activity levels, is essential. In fact, Cadilhac et al (2011) calculated that a 10% increase in physical activity could lead to an annual saving of 162 million Australian dollars. Although running is one of the most popular forms of physical activity world wide, it has been shown that up to 70% of competitive and recreational runners develop overuse injuries, in any one-year period (Hreljac, 2004); potentially resulting in increased levels of inactivity. The present project aims to evaluate the effect of a compliant running style on loading and energy expenditure, and whether augmented technology-based feedback can successfully enhance its adoption.

During running, foot contact with the ground generates high impact forces (Lafortune, Lake, Hennig, 1996). As the foot strikes the ground, a force is applied to the foot by the ground. This is transmitted through the ankle to the tibia, propagating up musculoskeletal system. As a result of Newton's second law of motion (F=ma), providing mass stays constant, acceleration can be used to infer force; thus a measurement called an 'impact acceleration' is often used in the description of impact loading. When excessive, impact accelerations have been implicated in the development of numerous overuse injuries such as degenerative joint disease, tendinitis, muscle tears, and stress fractures (Whittle, 1999; Lafortune et al, 1996; McMahon, G. Valiant, Frederick, 1987). Considering the implications of excessively high impact loading, there is clear justification for the development of a running style that may reduce these forces.

Humans have evolved to utilize a running technique that is very efficient, both in relation to the cost of transport (COT: energy expended per unit distance) and the cost of locomotion (COL: energy expended per unit time). This evolutionary process facilitated, or was driven by, successful hunting and feeding strategies (Steudel-Numbers et al 2008). This running technique requires relatively high impact loading in order to make effective use of the stretch shortening cycle to increase movement efficiency. Interestingly, other species have developed more compliant or crouched running styles that have not evolved with perhaps such an emphasis on the need for efficiency of movement (as in humans), but with a greater emphasis on an inherent need to protect themselves through lower impact loading. This can be seen in elephants, where a more bent knee is observed during foot-ground contact while running (Hutchinson et al, 2006; Ren et al, 2010). This facilitates a more compliant landing, where the very large forces that are generated from the large mass of an elephant are dampened, and potentially protects the elephant from injury. Similar strategies can be seen in monkeys where a more compliant, bent knee action, is used to run across thin branches in order to reduce the force applied to the branch, thus reducing branch movement, decreasing the risk of falling and injury, as well as decreasing the likelihood of being seen by a predator. If humans can run using a more compliant running style, it may be possible to reduce the risk of running related injuries, and increase physical activity levels. [With the evolution of societal structures and the development of associated modern technology, the need for humans to adhere to traditional running form to survive may no longer be necessary].

Interestingly, the concept of altering gait mechanics is not new and has been successfully used as a treatment for a number of medical conditions. For example, limb load monitors that measure the force developed under the foot have been previously used in children with cerebral palsy to improve walking symmetry (Seeger et al, 1981), and in adults with amputations or hip replacements (Gapsis et al, 1982). Furthermore, real time feedback provided by instrumented treadmills, has been shown to help participants maintain more symmetrical gait patterns after total hip replacements (White & Lifeso, 2005), as well as trans-tibial amputations (Dingwell et al, 1996). It is therefore clear that humans do have the capability of altering running technique and mechanics.

To date, very few studies have specifically examined the effect of altering running technique to make a running style more compliant. However, evidence does suggest that humans are capable of adopting a more compliant running style (McMahon et al., 1987, Crowell et al., 2010, 2011, Cheung et al., 2011, Clansey et al., 2014). McMahon et al (1989) demonstrated that increasing stance leg knee flexion decreased vertical stiffness of the body (making the style more compliant). In addition, McMahon et al (1989) also found that decreased vertical stiffness reduced the propagation of impact accelerations travelling throughout the musculoskeletal system, but failed to report the effect on specific sites (e.g. tibia, sacrum, head). This is an important consideration as the magnitude of impact accelerations experienced at any anatomical location is an implicating factor in the development of an overuse injury at that location. For example, the magnitude of impact accelerations experienced at the tibia has been implicated in the development of tibial stress fractures (Davis et al, 2004; Milner et al, 2006).

In order to facilitate a change in mechanics it is necessary that an appropriate intervention is developed. Research has shown that the use of technology-based real time visual biofeedback can be used to reduce tibial accelerations following an acute bout of gait retraining (Crowell et al., 2010) and a longer intervention period (Crowell et al., 2011, Cheung et al., 2011, Clansey et al., 2014). However, it is unclear from Crowell et al's work what kinematic changes have been made in order to facilitate said reductions (decrease in tibial acceleration ranging from 17-60%) or what effect this retraining may have on loading throughout the rest of the body, as these studies have only examined accelerations at the tibia. This thesis will aim to investigate the effect of such an intervention on both tibial and sacral acceleration values.

Since a change in running style may effect the energy expenditure, cost of locomotion, and cost of transport, it is important to determine how a change to the more compliant running style, suggested within this research, will influence these measures. This has a practical implication in that the likelihood of a person adopting the suggested compliant running style may be affected by these responses. McMahon et al (1989) found that compliant ("Groucho") running resulted in a significant increase in oxygen consumption of up to 50% in comparison to normal running. However, McMahon et al (1989) implemented an excessive degree of knee flexion that may not be necessary to reduce impact acceleration values. Regardless, depending on the goal of running, an increased rate of energy expenditure may have additional health and weight loss benefits and is therefore an important consideration in the adoption of compliant running.

In addition, when examining the effect of altering running style the effect of fatigue should also be considered. If an increase in energy expenditure is evident for compliant running, this may result in a shorter time to fatigue. Since, fatigue has been shown to increase the magnitude of impact accelerations in normal running, thus increasing the risk of injury development (Verbitsky et al, 1998), the effect of fatigue on a more compliant running style needs to be examined. To date, this has not been undertaken.

Primary aims of research project:

- To examine if a compliant running technique reduces peak impact accelerations, and what the associated kinematics and kinetics of this running style are.
- To examine what method should be employed in the teaching of runners to adopt this technique (verbal Vs. technology based biofeedback)

Secondary aims:

- To determine the effect of the altered running technique on:
  - Movement kinematic and kinetics, when fatigued and unfatigued
  - Cost of locomotion, cost of transport, and steady state 0_2 consumption.

The presented research is divided into 4 studies with the first study focusing on the use of verbally directed compliant running. The second study changes the focus towards a technology based biofeedback style of intervention and investigates the success of various accelerometer locations for providing real-time visual biofeedback. Subsequently, the third study then compares the success of such a technology-based biofeedback system relative to simple verbal based feedback and the last study investigates the effect of a longer technology-based intervention on kinetics, kinematics, and energy expenditure.

Aims of study 1:

<u>Aims of study 1</u>

- To investigate the effect of directed compliant running on:
  - Impact accelerations, joint kinematics, and joint kinetics in both fatigued and unfatigued conditions, in comparison to normal running.
  - Energy expenditure and other physiological responses in comparison to normal running.

Aims of study 2:

- To compare the use of three different accelerometer sites for providing visual biofeedback, in relation to their ability to reduce impact accelerations.
- To investigate the kinematic strategies used by participants to reduce the above impact accelerations.

Aims of study 3:

- To compare the use of verbal instruction, visual biofeedback and a simple information pack detailing compliant strategies, on their ability to alter running mechanics and decrease impact accelerations at the tibia and sacrum.
- To examine the kinematic changes made by participants to facilitate any reduction in impact acceleration magnitudes.

Aims of study 4:

- To examine the effect of a 4-week visual biofeedback gait-retraining programme on
    - Impact acceleration values.
    - Kinetics and kinematics
    - Cost of locomotion and Cost of Transport

- To examine the effect of speed on impact accelerations and energy expenditure, in normal and compliant running.

1.2 Summary of findings

The primary aim of this thesis was to determine if compliant running can effectively reduce the magnitude of impact acceleration values at various locations and what kinematic strategies are associated with this reduced loading. Results of each study provide a clear consensus that reduction of impact acceleration values is possible following adoption of compliant running technique, however the magnitude of this response appears to be heavily influenced by the method through which the compliant technique is taught.

Study 1, examined the effect of a verbally directed compliant technique, whereby instruction was focused around increasing knee and hip flexion. Results indicate that following this verbal instruction based intervention participants displayed reduced sacral and head accelerations (41% and 28% respectively), but not tibial accelerations. Kinematic analysis confirmed that these reductions were likely due to increased knee and hip flexion, relative to each participant's normal running style. Furthermore, this change in kinematics was associated with an increased energy expenditure of 21-24%.

Considering the success of previous technology based interventions (Crowell et al., 2010, 2011, Clansey et al., 2014) with regards to altering kinematic and reducing tibial acceleration values, the focus for introducing compliant strategies changed from the verbal instruction process in study 1, to the use of a technology based accelerometer biofeedback system in study 2. Given that previous research had only considered the effect of employing a biofeedback system via a tibial mounted accelerometer, study 2 aimed to determine whether other anatomical and off-body locations (sacrum and treadmill) might be more, or equally successful, as the tibia, for providing biofeedback, and subsequently reducing both tibial and sacral acceleration values (sacral accelerations had not previously been examined). Firstly, results indicated that this method was more successful at reducing tibial accelerations than study 1 (-26%) following an acute bout of biofeedback. Furthermore, both tibial and treadmill biofeedback appeared to reduced tibial acceleration values similarly (-23% and -26 % respectively) whereas the sacral biofeedback demonstrated a lesser effect with regard to tibial reductions (-1%). However, sacral biofeedback demonstrated the largest reduction with regard to sacral accelerations (-27%), followed by treadmill biofeedback (-17%), and tibial biofeedback (-6%). Thus it appears that treadmill biofeedback was able to facilitate adoption of a technique that brought about reductions to both the tibia and sacrum, thus potentially decreasing whole body load to a greater extent.

Study 3 then examined how an acute bout of tibial biofeedback compared, with regard to reduction of impact accelerations, against an acute bout of verbal instruction. This demonstrated that biofeedback is more effective at reducing peak acceleration values than verbal feedback (-39% Vs. -8%).

Finally, study four examined if the magnitude of reductions (in loading) displayed following biofeedback would increase if this system were employed across a 4-week intervention period, in a faded feedback design. Results indicate that this 4-week intervention significantly reduced both tibial and sacral acceleration values to a greater extent than both study 1 and 2 (-44% and -38% at the tibia and sacrum respectively). Furthermore, examination of cost of transport, cost of locomotion, and running economy indicated that the subsequent change in kinematics t did not affect the amount of energy expended.

Therefore it appears that employing a treadmill based biofeedback system across a 4-week period can successfully reduced peak tibial and sacral acceleration values, and thus potentially decrease the risk of running injury development.

Chapter 2: Review of Literature

2.1 Introduction

In order to understand the effect a movement pattern may have on the risk of injury development, it is necessary to understand the underlying mechanisms responsible for said injuries. Given that this research project aims to alter running mechanics, this review of literature will explore the association between running related injuries and specific kinetic and kinematic variables. Furthermore, the success of any change in running mechanics relies on an appropriate 'motor learning' intervention to bring about desired changes. Current information regarding appropriate feedback mechanisms to alter running mechanics will be discussed. Finally, given that running style largely influences the amount of energy expended during running this review will critically examine those kinematic and kinetic factors that affect energy expenditure. Any change in energy expended with altered running mechanics may yield additional benefits beyond injury reduction, which is the main focus of this thesis.

2.2 Factors effecting the development of running injuries

Numerous factors are associated with running related injury. From a biomechanical perspective these have been divided into kinetic factors and kinematic factors (Hall et al. 2013). All injuries are caused by relative excessive load; that is high loading relative to tissue integrity. For this reason, it is extremely difficult to determine when forces experienced by connective or soft tissues become "excessive" and subsequently result in injury, as not only does this vary from tissue to tissue, but also person to person. However, both prospective and retrospective (discussed below) research have drawn links between loading variables and specific injuries in an attempt to understand the underlying injury mechanisms and provide information that may be useful for both prevention and rehabilitation.

2.2.1 Kinetic Factors

Kinetic analysis is described as the measurement and subsequent interpretation of forces and powers that cause movement (Novacheck 1998). From a kinetic perspective it has been suggested that high ground impact forces may play a large role in the development of running injuries (Hreljac 2004). Every time a runner strikes the ground there is an associated ground reaction force that applies force to the body. In rear-foot strikers, this curve has a characteristic shape that includes two distinct peaks; the impact (passive) peak and the active peak, as can be seen in figure 2.1 (Hreljac 2004) .

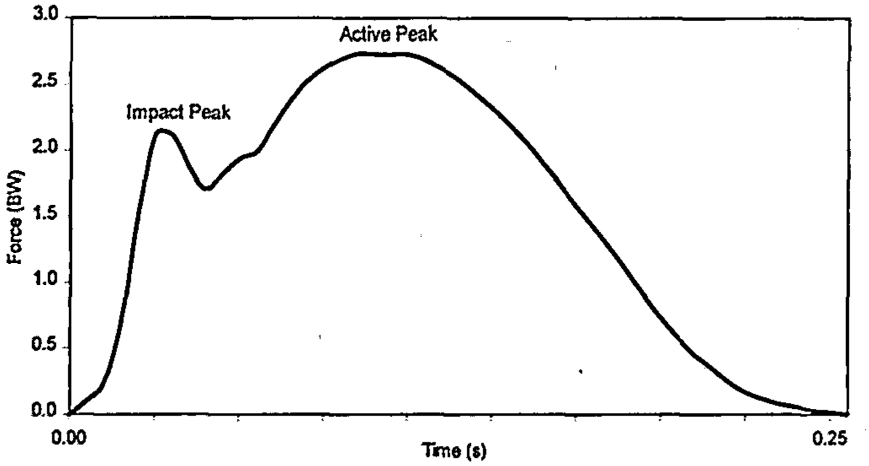

Figure 2. 1: Characteristic shape of vertical ground reaction force curve (adapted form Hreljac, 2004)

The impact peak generally occurs within the first 10% or first 30ms of foot contact with the ground and its magnitude is largely determined by the velocity of centre of mass (COM) and the foot at initial contact; in combination with any factor that may act to dampen the magnitude of force such as soft tissue, footwear, and surface (Nigg 1986). The active peak generally occurs in the latter part of the ground reaction force curve, anywhere between 65-75% of stance lasting for an average of 200ms, and occurring approximately around mid-stance. The active peak is considered the lower frequency element of a ground reaction force, as it is applied over a larger time period (Nigg 1986, Nigg 1983) in comparison to the characteristic high frequency nature of the passive impact peak that is short duration (Shorten, Mientjes 2011).

2.2.1.1 Vertical ground reaction force impact variables, impact accelerations, and injury

The magnitude of the vertical ground reaction force and the rate of loading associated with each ground reaction force have been implicated in the development of running injuries (Clement, Taunton 1980, Cavanagh, Lafortune 1980). However, consideration of the literature surrounding these variables and their relationship with injury development portrays a somewhat equivocal picture.

One of the largest prospective studies examining the occurrence of running related injuries and their relationship with kinetic variables (84 participants; injured (n=64); uninjured (n=20)) found that those who suffered running related injuries presented significantly larger vGRF impact peaks (VIP), average vertical loading rates (AVLR), and instantaneous vertical loading rates (IVLR) by 13%, 14%, and 9%(Davis, Bowser & Mullineaux 2010). Further prospective evidence indicates that participants who suffer tibial stress fractures and plantar fasciitis display average vertical loading rates and instantaneous vertical loading rates 33% and 40% larger than uninjured counterparts, respectively, and that those who suffer plantar fasciitis display vertical ground reaction force passive impact peaks 15% larger than uninjured controls (Bowser, Davis, 2010, Davis, Milner & Hamill, 2004) . These variables have also been implicated in the development of the above injuries in retrospective data, with runners who present history of tibial stress fractures displaying VIP, AVLR, and, IVLR values 5-8%, 3-18%, and 5-15% greater than uninjured control groups (Zifchock, Davis & Hamill 2006a, Creaby, Dixon 2008, Pohl et al. 2008, Milner et al. 2006a). Furthermore, participants that have suffered any running related injury, plantar fasciitis, or metatarsal stress fractures displayed VIP values 12%, 8%, and 2.26% greater than uninjured participants, respectively (Pohl, Hamill & Davis 2009, Dixon, Creaby & Allsopp 2006, Hreljac, Marshall & Hume 2000). While the aforementioned literature implicates vertical ground reaction force variables in the development of running related injuries, there is evidence to suggest that this may not be the case. In retrospective research, both Crossley et al (1999) and Bennell et al (2004) indicated no significant difference between uninjured participants and participants with a history of tibial stress fractures for vertical ground reaction force variables, while Dixon et al (2005) presented similar insignificant findings when comparing military recruits with a history of metatarsal stress fracture injury and matched uninjured controls. This has also been observed in relation to soft tissue injuries with Azevedo (2010) and Ferber et al (2010) presenting retrospective evidence highlighting no significant difference in vertical ground reaction force variables between uninjured controls, and runners with a history of Achilles tendinopathy and Iliotibial band syndrome, respectively.

It is evident from the above literature that there is no clear consensus in relation to the role that ground reaction variables play (or don't play) in the development of running related injuries. However, given the retrospective nature of those studies suggesting no relationship between ground reaction force variables and injury development, it may be appropriate to place more credence in the prospective data presented by Davis et al (2010, 2004) and Bowser et al (2010), indicating that ground reaction force variables may in fact be an implicating factor in the development of running related injuries.

Another difficulty when trying to determine potential cause-and-effect relationships between variables and injury occurrence, and that may provide insight and explanation into the varying results presented by the above authors, is both the nature of injury development and the ground reaction force as a variable. As outlined, injury is due to relative excessive load specific to an anatomical location. This makes it extremely difficult to determine if a specific variable plays a role in injury development, as the threshold for tissue damage and injury within each person and structure is unique. Furthermore, ground reaction force is a variable that represents whole body loading. However, the body does not experience load as a single body, but in-fact, deforms into multiple segments within a system. Each segment then experiences the load at different times, with different effective masses (dependent on the magnitude of joint deformation), and effectively experiencing loads of different magnitudes. Therefore a whole body measure of load may provide incomplete information in regards to what may be a relative excessive load of a specific structure or site within a system of deforming segments. We do not develop whole body injuries so how accurate can whole body measures really be in regards to determining injury risk?

It may therefore be more appropriate to measure site/segment specific loads when trying to determine an individual's risk of developing specific injuries. This concept is supported by research displaying how positional data can be used to calculate vertical ground reaction forces (Bobbert, Schamhardt & Nigg 1991) . Bobbert et al (1991) demonstrated that segmental accelerations contribute to different portions of the overall passive vertical ground reaction force curve, with the high frequency characteristics of the vertical ground reaction force passive peak primarily determined by leg segmental accelerations, and the absolute value of the passive peak depending largely on the contribution of the rest of the body. This information further highlights how limited the use of measures of vertical ground reaction force may be in determining the risk of injury development. For example, in examining the resultant intersegmental force occurring at the knee during the impact phase of running, a higher upper leg segmental peak may relate to the predisposition of an athlete to an increased risk of injury at the knee. However there may be no increase in the vertical passive (impact) ground reaction force peak. A consequence would be a masking effect whereby the risk of knee injury is hidden if only whole body loading was considered. This observation could potentially explain the confounding results presented by various authors in relation to the relationship between vertical ground reaction force variables and injury development. This leads to the question of whether or not data is present within the literature to suggest that a relationship between segmental impact accelerations and injury development exists.

The most common site for examining segmental impact accelerations is the tibia (Davis et al., 2010, Davis et al., 2004, Milner et al. 2006b, Pohl et al. 2008, Zifchock, Davis & Hamill 2006b, Milner, Hamill & Davis 2007). Prospectively, it has been shown that runners who suffer tibial stress fractures or any running related injury present peak tibial acceleration values 63% greater and 81% greater than uninjured controls, respectively (Davis, Milner & Hamill 2004, Davis, Bowser & Hamill 2010) . These results are consistent with retrospective research indicating that participants with history of tibial stress fracture injury display peak tibial acceleration values 13-28% greater than participants with no history of injury(Milner et al. 2006b, Pohl et al. 2008, Zifchock, Davis & Hamill 2006b, Milner, Hamill & Davis 2007) . This appears to offer strong evidence for the implication of peak tibial accelerations in stress fracture development and running injuries. Retrospective data is clearly limited in terms of its ability to provide insight into actual kinetic data of runners prior to injury, and is thus limited in regards to implications for injury prevention. However, this type of information (when considered in conjunction with prospective data) may give an insight into a runner's risk of re-injury post stress fracture occurrence. Interestingly, from the above research (both prospective and retrospective) it appears that the percentage difference (in comparison to healthy controls) is larger before stress fracture development (prospective-↑63%) relative to that after stress fracture development and subsequent return to health (retrospective-↑13-28%). Although this observation is limited as it is drawing from numerous populations in different studies, does this potentially suggest that post stress fracture occurrence, runners alter mechanics to reduce the magnitude of peak tibial acceleration? Examination of absolute acceleration values presented in the above studies may somewhat confound this idea (Prospective PPA= 4.73 g's Vs. Retrospective PPA= 5.9-8.1g's). However, without prospective and retrospective data on the same population this is unknown.

2.2.1.2 Joint moments, joint reaction forces, whole body moments, and injury

Human movement is a consequence of contracting muscles that generate moments of force about segmental joints; it is thus thought that the magnitude of joint moments and joint forces may play a role in running injury development (Scott, Winter 1990) and may therefore be important to discuss.

Currently there is very little prospective or even retrospective research available that shows clear associations between joint moments or joint reaction forces and running injuries, and thus makes it very difficult to draw conclusions from these measurements when trying to assess injury risk. However, knee joint moments have been implicated in the development of patella femoral syndrome both prospectively and retrospectively (Stefanyshyn et al. 2006, Stefanyshyn et al. 1999). From a group of 80 runners, Stefanyshyn et al (2006) presented data that indicates prior to the onset of PFP, runners display significantly larger knee abduction moment impulses than matched uninjured participants, by 65%. Similarly, participants with a history of PFP have been shown to present knee abduction impulses 31% greater than asymptomatic participants(Stefanyshyn et al. 2006). Furthermore, it has been suggested that increased hip adduction moments may result in an altered foot position at ground contact causing increased knee abduction moments, and thus risk of patella femoral pain development (Stefanyshyn et al. 2006). This is supported by prospective (Noehren, Davis 2007) and retrospective (Willson, Davis 2008) evidence indicating a relationship between excessive hip adduction and PFPS.

Similar evidence is present portraying a relationship between knee osteoarthritis and peak knee adduction moments (KAM) during walking; with patients who suffer from medial knee osteoarthritis and medial knee compartment cartilage damage presenting significantly higher KAM values in comparison to healthy controls (Baliunas et al. 2002). This is further supported by evidence suggesting a strong association between knee adduction moments and medial joint space (for every 1 unit increase in adduction moment there is an associated 0.63mm decrease in joint space width in patients with knee OA) (Sharma et al. 1998); in fact the magnitude of KAM is now accepted as a marker to determine the clinical severity of this degenerative joint disease (Hurwitz et al. 2002). Furthermore, intervention based research indicates that manipulation of joint kinematics; resulting in a decrease in KAM impulse (by up to 13%) subsequently reduces pain and WOMAC scores (survey that assesses pain, stiffness, and physical function in patients with osteoarthritis)(Haim et al. 2012).

The above-presented data appears to be the only prospective or retrospective evidence supporting a relationship between increased joint loading in running or walking and increased risk of injury development. However, it seems logical that any increase in joint loading may result in an increase in the risk of that load becoming relatively excessive.

Free moment, which is the measure of torque about the vertical axis due to friction between the foot and ground during stance phase of running (Holden & Cavanagh, 1991), has been suggested to be an implicating factor in the etiology of tibial stress fracture development (Milner et al, 2006). Retrospective data indicates that participants with a history of tibial stress fractures display significantly greater values for adduction free moment (ADDFM), free moment at impact peak, and absolute free moment by 48%, 83%, and 44%, respectively. Given the larger effect size and higher values for the absolute free moment, in comparison to free moment at impact and ADDFM, the magnitude of torque may be more important than its direction in relation to stress fracture development at the tibia (Milner, Davis & Hamill 2006) . This was confirmed by a binary logistic regression (R^2=0.27) demonstrating that absolute free moment is a good predictor of past stress fracture injury (Milner, Davis & Hamill 2006) . Similar evidence exists in relation to Iliotibial band syndrome (ITBS); participants with history of ITBS present peak rearfoot inverter moments 43% greater than uninjured controls(Ferber et al. 2010).

2.2.1.3 Internal loading, muscle forces and injury

It has already been demonstrated that measurements of whole body loads (such as vertical ground reaction forces) have numerous shortcomings as they may mask increased localised loading, and thus injury, at specific sites. Segmental accelerations clearly provide an advantage in that regard. However both vertical ground reaction forces and segmental accelerations are determined by external loading sources. Given the aforementioned prospective information relating such measurements to injury occurrence it seems that these loads may also be reflective of internal loads experienced at specific sites of injury. To support this, evidence presented by Edwards et al (2009) indicates that external force measures may be able to accurately account for internal conditions, and thus risk of injury development. Edwards et al (2009) demonstrated that vertical ground reaction force impact peak and peak impact acceleration accounted for 94% and 84% of tibial bone strain (measured via an internal strain gauge) when combined with cross sectional area and effective mass measurements respectively. This suggests that in a repeated measure design, whereby bone geometry remains unchanged, external force transducers represent an accurate estimation of internal bone strain. However, this study only examined axial compressive forces, neglecting the effect of muscle forces and thus potentially limiting implications.

During the stance phase of running, the forward inclination of the tibia relative to the ground causes posteriorly convex bending as a result of the vertical ground reaction force, while the line of force of the plantar flexors act to oppose this bending moment (Phuah et al. 2010). Given that the net internal moment will be resolved via summation of the moments generated via the vertical ground reaction force and muscle moments mentioned above, it seems intuitive that both aspects must be considered when determining injury risk due to bending forces experienced during running. Examination of 9 equidistant points along the tibia during running indicates that the resultant bending moments, reaction moments and muscle moments increase linearly from proximal to distal; represented by an average increase of 318% in resultant moment at the most distal location relative to the most proximal. This can be explained by an increase in moment arm for both reaction moments and muscle moments and the fact that the rate of increase is larger for the reaction moments, as the position becomes more distal. The larger magnitude of the negative reaction moment than the positive muscle moment creates a resultant net negative moment, with the largest values occurring during mid-stance (Phuah et al. 2010)(as can be seen in figure 2.2). This is supported by Scott and Winter (1990) who also demonstrated that reaction moments play a dominant role in relation to net tibial bending moments causing the tibia to bend convex posteriorly and peaking at an average of -89 N.m at mid-stance.

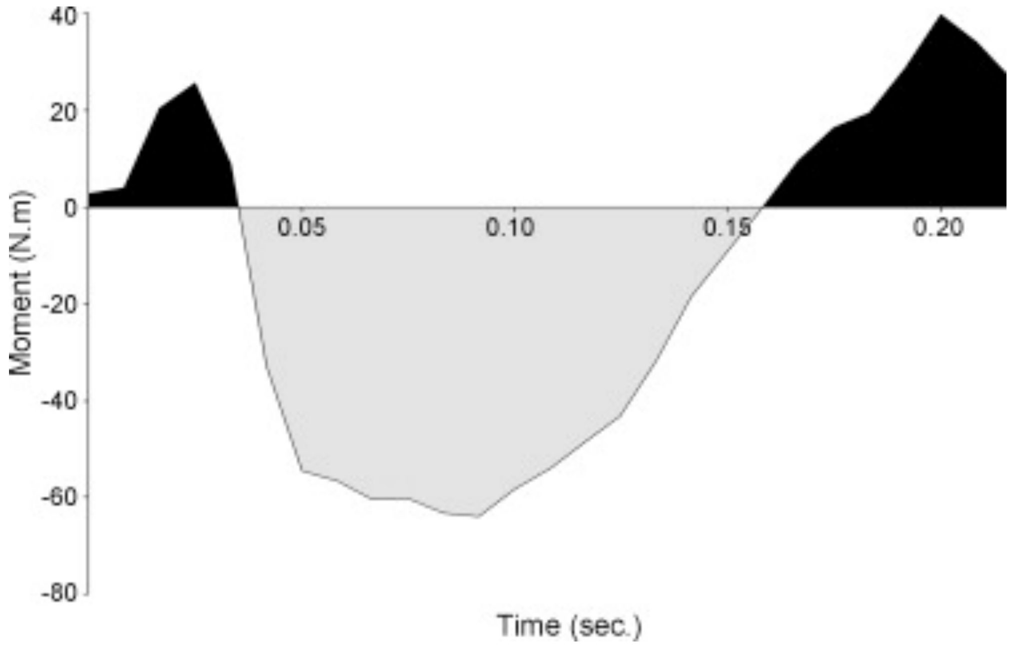

Figure 2. 2: Net moment experienced by the tibia during running (adapted from Phuah et al., 2010)

This indicates that during running the tibia experiences posterior tensile forces which become larger with a more distal position, and are largest during mid-stance. Considering that the distal tibia is one of the most common sites of stress fracture development (Ruohola et al. 2006, Brukner et al. 1998) and that cortical bone is relatively weak under tensile loading (Caler & Carter. 1989), it seems logical that the increased load experienced at this location relative to the rest of the tibia may be an implicating factor. This is somewhat supported by Sasimontonkul et al (2007) who demonstrated that the distal tibia experiences the highest compressive forces (combination of both external compressive forces and muscle forces) and shear forces (primarily a result of the reaction force), relative to the rest of the tibia, during mid-stance. Determining the dominant factor in relation to injury development at the distal tibia must be a priority if the goal is to reduce risk. Perhaps there is no dominant factor and interventions may be best served to address all variables that appear to predispose the distal tibia to higher loading. However, ease of measurement and ease of manipulation of kinematics to bring about a change in these variables must be considered. Examining each factor separately seems to indicate that reduction of the vGRF peak may have the most positive effect and is also a relatively simple measure that can be manipulated via simple alteration of running mechanics based on the impulse-momentum relationship (discussed in section 2.2.2). Reducing the vGRF peak would cause a subsequent decrease in joint reaction forces, causing a decrease in compression, shear forces, and tensile forces created by sagittal bending moments, thus decreasing each factor that appears to place the distal tibia under high loads during running. Further support for the relationship between ground reaction forces and joint loading comes from landing and jumping studies. Vertical ground reaction force has been shown to explain 99% of variance in knee joint reaction forces in double and single leg counter movement jumps and depth jumps, with joint reaction forces increasing with increased vertical ground reaction forces (Leissring et al. 2010). Thus, altering mechanics to reduce the magnitude of the vertical ground reaction force may have a positive effect on reducing the load experienced at the distal tibia via reductions in bending moments and compressive forces. However, this is unknown.

The above-mentioned research is somewhat in disagreement with the previously discussed literature that associates tibial stress fracture development and other running injuries to the loading experienced during impact (Bowser, Davis. 2010, Davis, Milner & Hamill. 2004). Examination of literature that has directly examined differences in mid-stance vertical ground reaction force values (active peak) in stress fracture groups versus healthy groups indicate no significant difference between groups, and in fact stress fracture groups consistently demonstrate smaller (however insignificant) values for the active peak (occurring during mid-stance) than the uninjured healthy controls (Creaby & Dixon. 2008, Bennell et al. 2004, Zifchock et al. 2006, Davis et al. 2004, Crossley et al. 1999, Grimston et al. 1994, Dixon et al. 2006). This suggests that ground reaction forces occurring during mid-stance may not be an implicating factor in stress fracture development. This raises the question of what determines injury? If the dominant factor is relative excessive load, then it would make sense that higher loading at the distal tibia during mid-stance would increase injury risk. Is there therefore another factor to consider? Given that the participants in the above studies (Phuah et al. 2010, Sasimontonkul et al. 2007) were healthy and uninjured, it may be that the distal tibia experiences a unique loading pattern that predisposes that specific location to stress fracture (relative to the rest of the tibia), but alone does not cause a fracture to occur. However, without prospective or retrospective data for bone bending moments, joint reaction forces and tibial compressive forces, it is not possible to fully understand the effect of these loads in relation to stress fracture development.

Similar work indicates that regions of the femur that are particularly prone to stress fracture development display unique loading patterns that may be an implicating factor in the development of injury during running (Edwards et al. 2009). Up to 50% of all femoral stress fractures occur at the neck of the femur (McBryde 1985, Niva et al. 2005a). Examination of load occurring at the femoral neck indicates large anteriorly and medial oriented shear forces (relative to the rest of the femur), occurring simultaneously with peak axial forces and moments during the impact phase of running. The impact phase of ground contact is associated with extremely high loading rates, which are in turn associated with cortical bone damage (Schaffler, Radin & Burr 1989) . Therefore it may be possible that the relatively high shear forces, axial reaction forces and sagittal hip moments occurring during the impact phase, in combination with the extremely small diameter of the femoral neck may pose a threat to skeletal integrity(Edwards et al. 2008a). Given previous data demonstrating associations between loading variables during impact and running injuries (Davis et al. 2010) it seems logical that the unique loading experienced at the femoral neck relative to the rest of the femur may be an implicating factor in stress fracture development in this region. The largest medial-lateral bending moments, axial forces, torsional moments, and anterior posterior shear forces were observed at the distal femur proximal to the femoral condyle (Edwards et al. 2008b). This location has also been identified as a relatively common site for femoral stress fracture development (Niva et al. 2005b, Schmidt-Brudvig 1985, Milgrom et al. 1985, Giladi et al. 1986). Peak loads for these variables occurred during mid-stance and were associated with peak patella-femoral contact forces, and peak muscular force of the gastrocnemius and quadriceps. Research indicates that simultaneous torsional loading and axial loading can cause a seven-fold reduction in cortical bone fatigue (George, Vashishth 2005). Therefore it seems the aforementioned variables may play a role in the development of femoral stress fractures of the distal femur close to the femoral condyle. However, without prospective data these relationships remain unknown.

To conclude, there appears to be a number of kinetic factors that play a role in the development of running related injuries. Both vertical ground reaction force impact variables and segmental impact accelerations present strong prospective and retrospective associations with general running injuries, stress fracture development, and plantar fasciitis (Davis et al., 2010, Davis et al., 2004, Bowser et al., 2010, Zifchock et al., 2006, Pohl et al., 2008, Dixon et al., 2006). Similarly, joint loading appears to play a role in the development of patella femoral pain and medial knee osteoarthritis and free moments appear to be strongly associated with tibial stress fracture injury (Stefanyshyn et al., 1999, 2006, Sharma et al., 1998, Milner et al., 2996). Examination of internal and external forces together suggests that a combination of factors predispose the distal tibia to a unique loading pattern that may indicate increased susceptibility to injury (Phuah et al., 2010). This data indicates that it is the external forces (ground reaction forces, joint reaction forces, and moments created as a result) that play a dominant role in internal shear forces, bone-bending moments, and compressive force; indicating that reduction of the vertical ground reaction force may have a positive effect on internal loading conditions and thus injury risk. Given that injury is a result of relative excessive load its seems intuitive that reduction in any of the above variables may have a positive effect on reducing injury risk. Finally, given that impact accelerations provide information specific to a segment, there is a clear benefit to using accelerometers to determine the risk of injury to specific sites (e.g. tibia and sacrum). Also the relative low cost of accelerometers and their ecological advantage (not confined to a lab) provides numerous benefits in comparison to a lab-based motion analysis system.

2.2.2 Kinematic measures and running injury

Given that the aim of this project is to examine how changes in kinematics may influence loading applied to the body when running, and hence the potential predisposition to injury, it is necessary to understand (i) what kinematic variables can be manipulated, and (ii) what effect these manipulations may have on the loading/kinetic measures associated with injury development.

Manipulation of running technique to reduce loads experienced by the body can be explained via the impulse-momentum relationship. When altered so that the equation is essentially defining force (as seen below) it becomes clear what variables can be targeted to reduce the force responsible for injury development as a result of the impact experienced during running.

$$\bar{F} = (m \cdot \Delta v) / \Delta t$$

Assuming mass remains constant the two mechanisms through which whole body force can be decreased are (i) decreasing the change in velocity, and (ii) increasing the time over which the impact force is applied. However, given that we are interested in segment-specific loading, effective mass (as defined in section 2.2.1) may be manipulated by altering the amount of joint deformation at impact, thus introducing a third variable that has the potential to influence impact force. Mechanisms through which running mechanics can be altered to subsequently bring about change in the aforementioned variables will therefore be discussed.

2.2.2.1 Kinematics, effective mass, and impact loading

Effective mass is the portion of the total system mass that would be needed to accurately model an impact if a single mass particle were used instead of a system of rotating and deforming segments such as in the human skeletal system during running (Derrick, 2004).

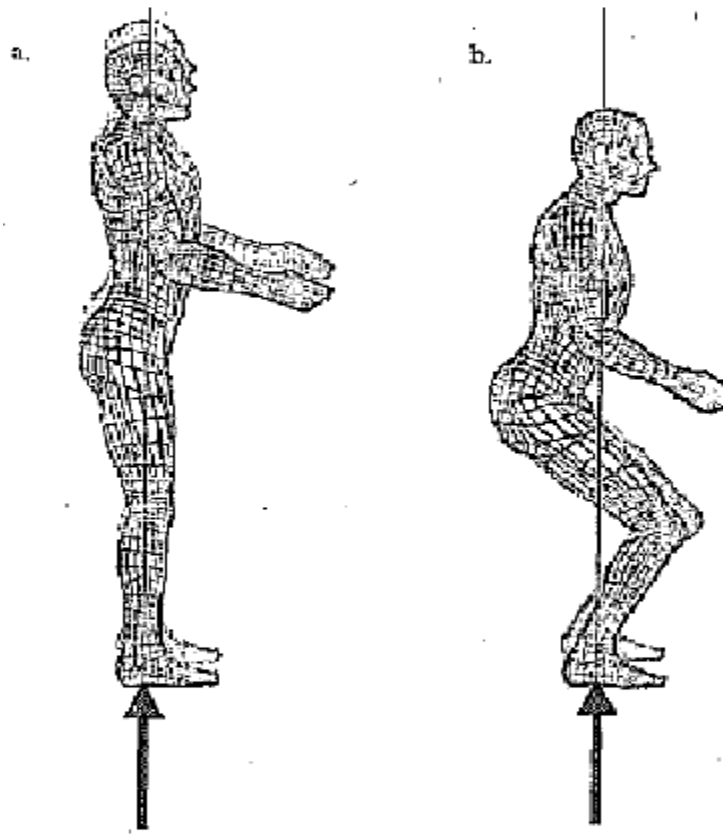

Figure 2. 3: Compliant landing versus stiff landing (adapted from Derrick, 2004)

Upon impact, if joint deformations remain negligible, the ground reaction force vector passes through the joint centre (for example, when the body is in a fully extended position) (figure2.3). When this occurs the effective mass is essentially the mass of the body and the whole body is accelerated as a single rigid unit (Derrick, 2004). However, if joints such as the knee are flexed, the body deforms from a single rigid unit into multiple segments that accelerate separately, causing the greatest accelerations to occur at the segments closest to the impact surface (Derrick, 2004). This indicates that both joint stiffness and joint orientation may play a role in determining effective mass. If a system were infinitely stiff, effective mass would remain the mass of the entire body regardless of joint orientation, as no joint deformation would occur. On the other hand, if the lower extremity were infinitely compliant relative to the foot then the effective mass would be equal to the foot only (Daoud, 2009). Given that running is neither infinitely stiff nor infinitely compliant joint orientation at contact may influence joint stiffness and effective mass. Joint stiffness is calculated by dividing the change in moment of force at that joint by angular displacement. Therefore, under conditions whereby flexion occurs at impact (for example at the knee), the ground reaction force vectors distance from the point of rotation of that joint is increased, thus generating larger external moments, subsequently encouraging joint rotation (and influencing stiffness), resulting in deformation and reduced effective mass (Devita & Skelly, 1992). In fact, evidence suggests that increasing knee flexion at impact from 5° to 20° can decrease effective mass of a body from 11kg to 5kg in a 65kg participant (Denoth, 1986). Interpretation of Newton's second law and the impulse-momentum relationship would therefore indicate that a decrease in effective mass might also represent a decrease in impact force. This is supported by Gerritson et al (1995) who demonstrated for every 1° increase in knee flexion there is a subsequent 68N decrease in vertical impact peak. Considering the association between ground reaction force impact variables and running injuries (Davis et al., 2010), this may potentially result in a subsequent decrease in injury risk. Therefore, both joint stiffness and joint geometry may play an important role in determining injury risk and will be discussed further in the next section (section 2.2.2.2).

However, further consideration of Newton's second law indicates that a decreased effective mass would also be associated with an increase in tibial acceleration and presents a potential caveat for the use of accelerometers to infer impact loading, as accelerations may be artificially inflated (an increase in acceleration due to decreased effective mass that does not represent increased loading). This should therefore be considered when interpreting accelerometer data. Despite this, accelerometers provide a unique ecological advantage and offer information regarding segment specific loading that is extremely advantageous as previously discussed.

2.2.2.2 Joint orientation, stiffness, and impact force

The association between joint orientation, stiffness, and impact force has been further examined using a swinging pendulum device to mimic foot-strike. Lafortune et al (1996b) demonstrated that increasing knee flexion angle at impact from 0-40° decreased the vertical impact peak by 30%. This was further associated with a decrease in effective axial stiffness 114% (Lafortune et al, 1996). However, in this work Lafortune et al (1996 & 1996b) failed to account for muscle activity. Given that stiffness of the leg is determined interdependently by joint angle and the force generated by the muscles crossing the joint, this is an important consideration (Denoth, 1986; Bobbert et al., 1991). Subsequent work completed by Potthast et al (2010), using a similar pendulum device, found that knee joint angle and muscle pre-activation explained 25% and 48% of impact force, respectively (figure 2.4). At constant muscular contraction, this was represented by a 9N decrease in vertical impact peak with an increase in knee flexion from 0-20° (at 60% maximal voluntary contraction (MCV)) and a decrease of 158N as knee angle increased from 0-40° (at 30% MCV). Furthermore, impact force decreased at all knee angles as muscle activity decreased (figure 2.5). This indicates that with a stiffer joint (due to increased muscle activity) the effect that increased knee flexion has on impact force becomes smaller and that although altering initial knee angle at contact has the ability to alter impact forces it does not necessarily dictate stiffness. Thus it seems increasing joint flexion and decreasing joint stiffness may have a protective effect due to the association between the vertical impact peak and running injury (Davis et al., 2010).

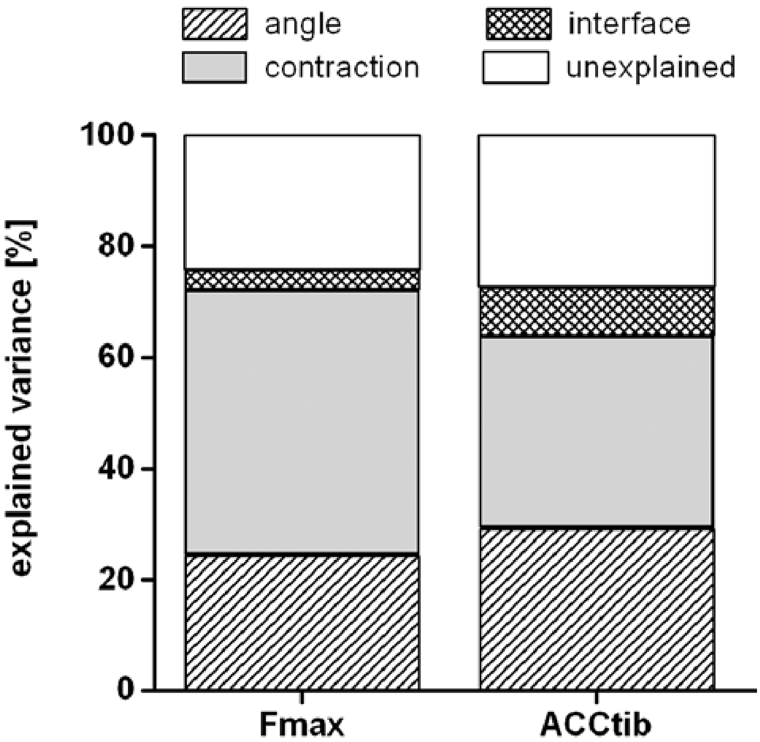

Figure 2. 4: contribution of knee joint angle and muscle pre-activation to impact force (adapted form Potthast et al., 2010)

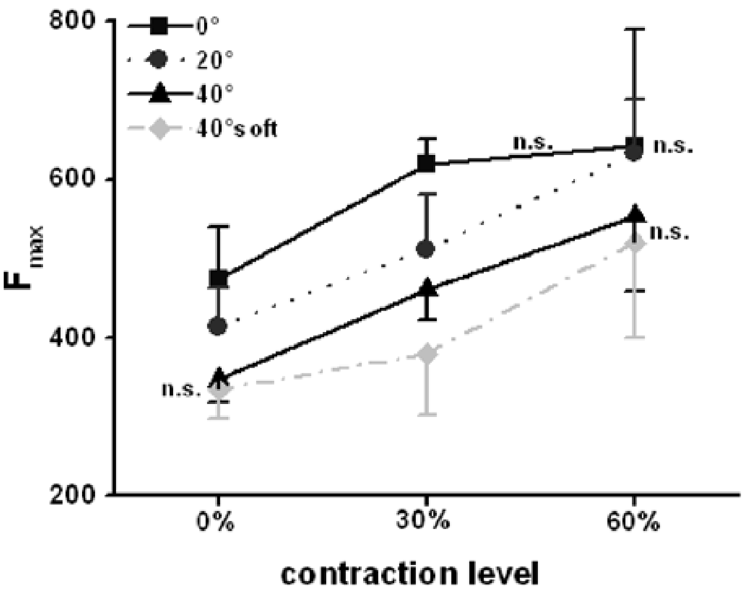

Figure 2. 5: Knee joint angle, pre-activation and impact force (adapted form Potthast et al., 2010)

The above studies also investigated the effect of altered knee angle and stiffness on impact acceleration values. Lafortune et al (1996) demonstrated that as knee flexion increased from 0-40° tibial impact acceleration increased by 57%. This is supported by Potthast et al (2010) who showed that tibial acceleration increased by 39%, 48%, and 46% as knee flexion increased from 0-40° at MVC values of 0%, 30%, and 60% respectively. Furthermore, when knee angle was controlled, tibial acceleration increased by 30%, 48%, and 22% at knee flexion angles of 0°, 20°, and 40° as MVC decreased from 60%-0%(Potthast et al., 2010). Thus, increased knee flexion and decreased joint stiffness appear to increase tibial acceleration values. This is further ratified by Derrick et al (2004) who suggested that every 1° increase in knee flexion during heel toe running is associated with a 0.27g increase in tibial impact acceleration. However, this increase in tibial acceleration is likely due to a decrease in effective mass as opposed to an increase in load experienced at the tibia. Although absolute impact acceleration values at the tibia appear to increase with increased knee flexion, this appears to be associated with an increased whole body attenuation capacity. Lafortune et al (1996) found that increased tibial impact acceleration of 57% (that resulted with increased knee flexion from 0-40° and decreased stiffness) was associated with a decrease in head acceleration values of 47%. Thus it seems that the body's ability to attenuate impact as it travels up the musculoskeletal system becomes greater with increased knee flexion. Similar findings have been presented by McMahon et al (1987) who found that Groucho running, characterised by increased knee flexion and decreased effective axial stiffness (50° knee flexion and an 82% decrease in stiffness at mid-stance), was associated with a 20% improvement in attenuation of impact acceleration from the tibia to the head; represented by a decrease in head to tibia acceleration ratio from 0.4-0.6 in normal running to 0.1 for Groucho running. However, the important finding is not the increased attenuation, as this could theoretically occur as a result of increased tibial loading without any change in loading experienced at the head, but that this increased attenuation was due to a significant reduction in head acceleration values (actual decrease not reported). Thus it seems the body may adapt to protect the head from adverse accelerations that may damage the visual vestibular system that controls human location, and is thus is a priority (Lafortune et al., 1996). This phenomenon is not supported by Milner et al (2007) who demonstrated similar positive associations between knee flexion and tibial acceleration (r=0.583 for controls and r=0.063 for tibial stress fracture group), but also demonstrated a positive relationship between knee stiffness and tibial acceleration (r=0.406 for the tibial stress fracture group and r=0.161 in the control group); indicating that tibial acceleration would decrease with decreased knee stiffness. However, Milner et al (2007) examined these relationships in normal running (where individual variations of knee flexion and stiffness are relatively small), thus the proposed effect of increased tibial acceleration as a result of decreased effective mass (due to decreased stiffness and increased knee flexion) may only be a factor when stiffness and joint angle are drastically altered (as in Lafortune et al 1996, Derrick et al 2004, and Potthast et al., 2010).

To conclude it seems both lower limb joint angle and stiffness play an important role in determining load experienced by the body, and must be considered when trying to alter kinematics to reduce injury risk.

2.2.2.3 Lower limb orientation, stiffness, and duration of impact phase

As already discussed, based on the impulse-momentum relationship, increasing the time over which force is applied can reduce the magnitude of said force, thus potentially decreasing the likelihood of injury development. Thus, any kinematic change that can extend the duration of the impact phase during running may be beneficial. Manipulation of running mechanics to increase knee joint flexion to 50° at mid-stance and decrease axial stiffness by 82% (a running style called Groucho running) was shown to increase contact time (McMahon et al., 1987). This is supported by Blickhan (1989) who predicted that leg spring stiffness would have a negative relationship with contact time if vertical landing velocity, speed, and leg length remained constant; indicating that as leg spring stiffness decreases contact time would increase. Thus, it appears that decreasing lower limb stiffness and increasing knee flexion may reduce impact force by both decreasing effective mass and by increasing the impact time interval. Landing studies further supports this relationship. Devita and Skelly (1992) demonstrated that compliant drop landings were associated with significantly larger contact times and impact phase duration (by 125% and 48% respectively) than stiff drop landings. This suggests that increasing the time period over which force has to dissipate may potentially reduce the risk of injury associated with impact force.

2.2.2.4 Lower limb stiffness, joint orientation and injury

Regardless of how lower limb stiffness and joint angle influence effective mass, impact phase duration, impact force, and impact accelerations, both have been retrospectively associated with injury development. Evidence suggests that participants with a history of tibial stress fracture present knee stiffness values 47% greater during the impact phase of running than uninjured controls (Milner et al., 2007). However, no difference was observed for either knee angle at contact or knee flexion excursion. Considering that knee joint stiffness is determined by dividing the change in joint moment by the change in joint angle during initial contact (Farley and Gonzalez. 1996), it is logical to assume that this increased stiffness in the stress fracture group was due to the 33% larger change in sagittal knee joint moment, which was suggested to be a subsequent result of the increased rate of loading in the stress fracture group relative to the healthy controls (15%, and 19% increased AVLR and IVLR, respectively) (Milner et al 2006, 2007). This suggests that increased knee joint stiffness may be an implicating factor (along with numerous other factors) in tibial stress fracture development.

Similarly, evidence suggests that participants suffering from pre-osteoarthritic knee pain present initial knee flexion angles 300% and 85% smaller (more extended) during the impact phase and stance phase of running, respectively, when compared to healthy uninjured control (Radin et al., 1991). Furthermore, this indicates decreased knee joint deformation. This was associated with a 21% and 51% increase in tibial and thigh acceleration respectively, an increase in peak loading rate of 41%, and increased vertical heel velocity of 21%. Although stiffness was not measured, the decreased joint deformation in conjunction with increased rate of loading experienced by the knee pain group would increase sagittal knee moments (as in Milner et al., 2007) and thus knee stiffness.

To conclude, it appears that injured runners display kinematic abnormalities in comparison to healthy controls that may predispose them to injury development. Increased leg stiffness and decreased knee flexion appear to be characteristic of injured runners, however due to the retrospective nature of this data it is unknown if these differences predisposed these participants to injury or are a subsequent effect of injury.

2.2.2.5 Vertical landing velocity and impact loading

When the foot strikes the ground during running, velocity must be brought to zero, therefore, any change in vertical landing velocity will have the subsequent effect of altering the rate of change of velocity and thus manipulate the impulse-momentum relationship to either increase or decrease force. Employing a 4 segmental skeletal model Gerritson et al (1995) estimated that as vertical landing velocity of the COM increased from -0.2m/s there would be an associated increase in vertical impact peak of 212N (for every 0.1m/s increase in velocity) and a subsequent decrease in time to peak (~0.2 seconds at -1.0m/s and 0.35 seconds at -0.2 m/s). This was confirmed at a later date by Liu and Nigg (2000) who, implementing a simplified spring-dampener model, showed that as landing velocities (of COM) increased across the same range as previously detailed by Gerritson et al (1995) (-0.2 to 1.0 m/s) there was a similar increase of 222N in vertical impact peak associated with every 0.1 m/s incremental increase in landing velocity. Zadpoor et al (2007) utilised a similar model, but with velocities more representative of those observed during running, and found a similar relationship whereby the magnitude of vertical impact peak increased by 36% as landing velocity increased from 0.56m/s to 1.36m/s.

The question arises: how can vertical landing velocity be decreased during running? Considering the laws that govern projectile motion it seems rational that decreasing vertical displacement of the COM overall, as well as of the foot relative to the pelvis, before initial contact would decrease the vertical velocity of the foot at landing, and thus impact force. This has been observed in Groucho running where a decrease in vertical oscillation of COM was associated with decreased vertical landing velocity (0.5-08m/s in normal running versus almost 0m/s in Groucho running) (McMahon et al., 1987). This may also be influenced by manipulation of stride length and frequency as will be discussed in the next section.

2.2.2.6 Stride length, frequency, and impact loading

Alteration of both stride length and frequency can result in subsequent manipulation of the impulse-momentum relationship, resulting in either increased or decreased average loading. This is likely due to changes in vertical displacement of COM and subsequently vertical landing velocity. Habaro et al (2012) demonstrated via regression analysis that vertical impact peak, and vertical loading rates were minimised with increased stride frequency (relative to preferred) when running at constant speed (figure 2.6).

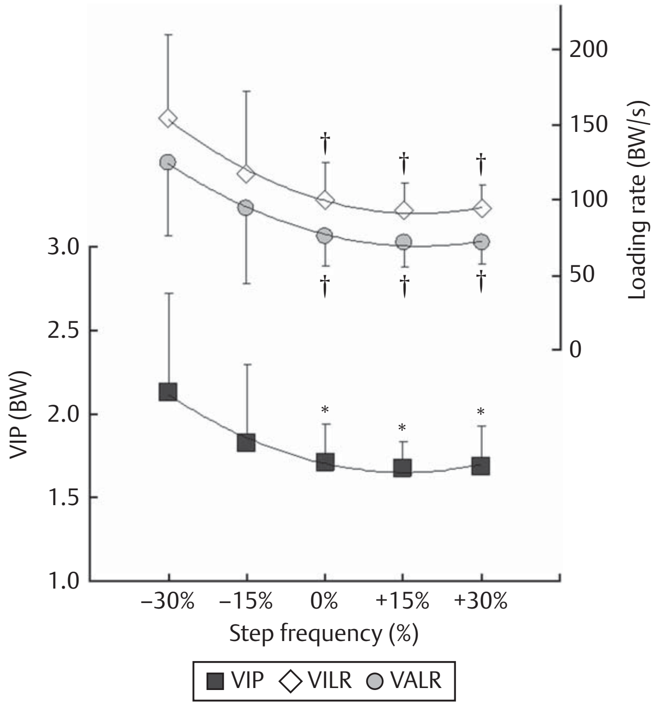

Figure 2. 6: Effect of stride frequency on vertical impact peak (adapted from Habaro et al., 2012) (VIP= vertical impact peak; VILR= Vertical instantaneous loading rate; VALR= Vertical average loading rate)

This is supported by Derrick et al (1998) who indicated that the duration of impact phase increased from 0.58m/s to 0.68m/s as stride length changed from +20% to -20% preferred stride length, indicating that decreased stride length is associated with decreased loading rates. Given that decreased stride length is associated with increased stride frequency it is likely that the above changes in vertical loading and loading rate occur as a result of the same mechanism. This is supported by Heiderscheit et al (2011) who demonstrated that stride length decreased by 38% as stride frequency increased from -10% to +10% preferred frequency. Heiderscheit et al (2011) also found a 28% decrease in COM vertical excursion associated with the above alterations to stride frequency. Thus increased stride frequency and decreased stride length, may act to reduced vertical displacement of

COM, subsequently reducing vertical impact velocity, and thus potentially explaining the decreased loading observed by both Derrick et al (1998) and Habaro et al (2012). This is somewhat supported by Derrick et al (1998) who demonstrated increased vertical velocity of the heel at initial contact of 68% at +20% stride length relative to preferred stride length. Although vertical velocity of COM and the foot are not completely interdependent, given that vertical impulse magnitude was also shown to increase by 16.5 % (from preferred step length to +20%), it seems possible that this was due to an increased vertical velocity of COM. Further support for the relationship between increased stride frequency (and consequently decreased stride length) and subsequent decreased loading comes from numerous authors that indicate a positive linear relationship with peak tibial acceleration (Hamill et al., 1995; Derrick et al., 1998; Mercer et al., 2002; 2003; Heiderscheit et al., 2011).

Although increased stride frequency would result in an increased number of loads for any distance or time, evidence suggests that the magnitude of load plays a dominant role over the number of loading cycles, in relation to risk of stress fracture development (Edwards et al., 2009). Edwards et al (2009) demonstrated using a probalistic stress fracture model that the risk of developing a tibial stress fracture decreases when running at -10% stride length by 3% at 3 miles per day, 4% at 5 miles per day, and 6% at 7miles per day (as can be seen in figure2.7). Thus the increase in the number of loading cycles that would be associated with reduced stride length appears not to have a negative effect on risk of stress fracture development. It is important to note that the probalistic stress fracture model employed by Edwards et al (2009) is a mathematical method developed by Taylor (1998) that accounts for bone failure, bone repair, and bone adaptation. Probability of failure is determined by the amount of stress applied to the bone and the volume of bone but also considers the scatter present in data. Subsequently, both increased stress and increased volume will increase the probability of fracture. For further information refer to Taylor (1998).

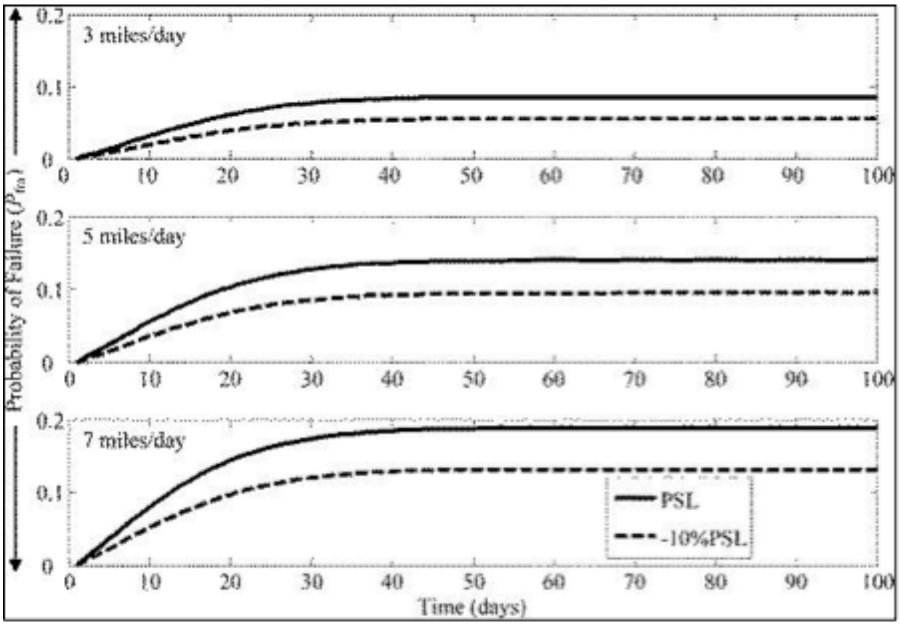

Figure 2. 7: The effect of stride frequency and mileage on tibial stress fracture development (adapted from Edwards et al., 2009); PSL=preferred stride length

To conclude, it seems evident that when utilising a traditional running style, altering kinematics to reduce stride length and increase stride frequency reduces the magnitude and rate of loading experienced by the runner (via manipulations of the impulse-momentum relationship), and ultimately may reduce the risk of injury development.

2.2.2.7 Running speed and impact loading

Numerous authors have demonstrated that as running speed increases both vertical impact forces and vertical loading rates also increase, and thus may influence risk of injury development. Munro et al (1987) found a 48% increase in vertical impact peak, 46% increase in vertical impact loading rate, and 48% increase in vertical landing velocity when running speed was increased from 3-5m/s. This is supported by both Bates et al (1983), who found that time taken to reach vertical impact peak decreased by 9% (thus increasing rate of loading) and vertical impact peak increased by 58% when running speed increased from 4-7m/s, and by Mercer et al (2002) who demonstrated that peak tibial impact acceleration values increased by 79% when running speed increased from 50-100% maximum running velocity. This is supported by evidence suggesting that the probability of developing a tibial stress fracture from 3 miles of daily running increases from 17%, to 19%, and to 33% as running speed increases from 2.5m/s, to 3.5m/s, and to 4.5m/s, due to an increase in peak tibial strain of 28% from the lowest to highest speed (Edwards et al., 2009). This increase in impact force and loading rate associated with increased running speed may be explained due to associated increases in stiffness, and a subsequent decrease in the duration of the impact phase. This is supported by Armpatzis et al (1999) who demonstrated that as running speed increased from 2.61-5.59 m/s there was an associated increase in leg stiffness of 39% and total vertical stiffness of 200%, which may subsequently lead to decreased contact time.

Thus, altering running speed appears to alter variables within the impulse-momentum relationship that ultimately leads to an increased magnitude and rate of loading, and increased risk of injury development. However, is running speed actually a modifiable characteristic? If performance is the goal then reducing speed to reduce injury risk is not an option. Similarly, if fitness is the goal of running, reducing running speed to avoid injury seems an unlikely and unattractive option given the associated decrease in metabolic stress and thus cardiovascular adaptation. However, this may not be an issue when employing a more compliant running style as there can be up to a 50% increase in energy expenditure (McMahon et al., 1987). Thus reducing running speed may not be an issue for the recreational runner, if it is associated with such compliant mechanics.

2.2.2.8. The effect of foot-strike on stiffness, impact forces, and injury development

During running, initial contact of the foot with the ground can occur at varying angles of either plantar or dorsiflexion. This factor has subsequently been highlighted as an important aspect of running kinematics (Lieberman et al., 2010) that may influence injury development (Lieberman et al., 2010; Daoud et al., 2012). Although the present study does not aim to directly influence foot-strike, it is the subsequent increase in compliance as a result of a forefoot strike that is important to understand. Runners with a forefoot strike have been shown to present plantar flexion and knee flexion angles 257% and 88% respectively larger than habitual heel-strikers (Lieberman et al., 2010). The subsequent result is increased ankle and knee joint range of motion resulting in a 75% greater drop in COM (increased ROM) during contact and a decrease in vertical stiffness of 17%. Thus, in forefoot runners the perpendicular distance of the moment arm from the point of rotation about the joint is increased both at the ankle and at the knee, subsequently reducing effective mass via mechanisms discussed earlier (in section 2.2.2.1). Effective mass was therefore calculated as 1.7% for forefoot strikers versus 6.8% for heel-foot strikers (calculated relative to body mass). As a result vertical loading rates were shown to be 50% (AVLR) lower for barefoot forefoot runners than shod heel-strikers (Lieberman et al., 2010). Thus it appears that increased lower limb joint flexion results in subsequent reductions in stiffness, reduced effective mass, reduced load, and potentially reduced injury risk. This relationship has been supported via retrospective association between foot-strike pattern and injury development (Daoud et al., 2012). Injury occurrence was assessed in runners over a 9-month period and revealed that 69% of the athletes were habitually RFS, 31% FFS, with no MFS present in this cohort. Examination of injury rates indicated that RFS were 2.5 times more likely to have developed mild-moderate repetitive stress injuries than the FFS, and 1.7 times more likely to have developed moderate-severe stress injuries than the FFS. However, considering that these athletes would complete a significantly larger volume of running, at faster speeds, which would subsequently increase both the magnitude of impact forces (Munro et al., 1987) as well as the number of impact forces experienced by the athlete, in comparison to that experienced by a recreational runner, these findings may not be generalizable to non-competitive/recreational populations.

2.2.3 Neuromuscular Fatigue and Impact loads during running

Given that altering kinematics may bring about an increase in energy expenditure, it is also logical to assume there may be a subsequent increase in fatigue, or a decrease in the time taken to reach a fatigued state. It is therefore critical to discuss the physiology of neuromuscular fatigue and the effect of fatigue on the risk of injury development.

During physical activity, a decrease in performance associated with an increase in actual and/or perceived effort is termed fatigue (MacIntosh, Gardiner, & McComas, 2005). However, the initial condition of the neuromuscular system is altered as soon as exercise begins causing fatigue to gradually progress until the muscle is no longer able to perform the required task (Boyas and Guevel, 2011). According to Bigland-Ritchie and Woods (1984) neuromuscular fatigue is therefore defined as any reduction in force or power irrespective of whether the task can be sustained or not. Although the same functional processes involved in muscle force generation extend to the entire neuromuscular system, many different elements may be responsible or involved in the manifestation of neuromuscular fatigue (Boyas and Guevel, 2011). Within the neuromuscular system functional alterations at certain sites are responsible for the progression of fatigue. These sites can be categorised into central (CNS) and peripheral (PNS) fatigue and can be seen in figure 2.7.

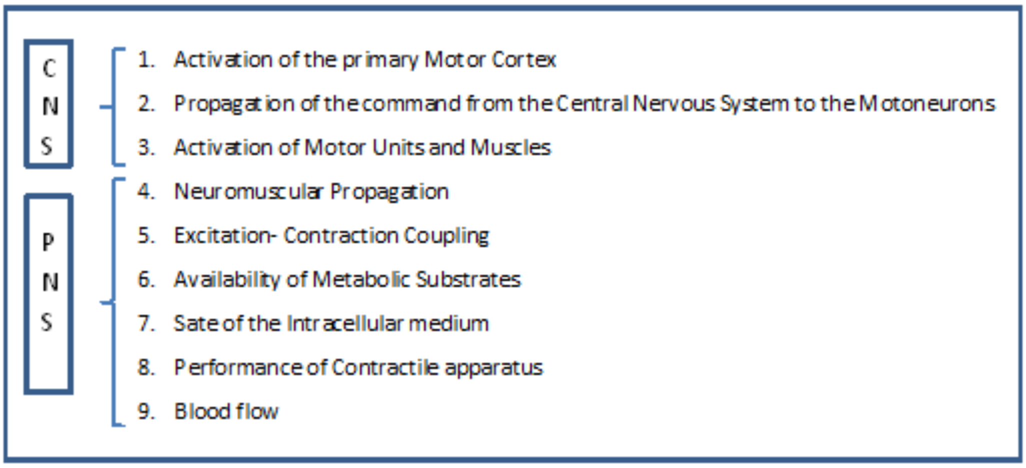

Figure 2. 8: Central and Peripheral contribution to fatigue (adapted from Boyas and Guevel, 2011)

Central fatigue causes a decrease in the voluntary activation of muscle as a result of a decrease in the number and discharging rates of motor units recruited at the start of muscle contraction whereas peripheral fatigue manifests as a decrease in the contractile strength of muscle fibres with a change in the mechanisms underlying the diffusion of muscle action potentials (Boyas and Guevel, 2011; Gandevia 2001; & Gandevia, Allen, and McKenzie, 1995).

Central fatigue incorporates all supraspinal and spinal physiological phenomena capable of provoking a decrease in motoneuron excitation, and thus voluntary muscle activation (Boyas and Guevel, 2011). Over 25% of the drop in force associated with sustained contractions, such as in running, can be accounted for by central fatigue (Gandevia, 2001; McNeil, Martin, Gandevia and Taylor 2009; & Taylor and Gandevia 2008). Weaker central command during prolonged exercise results from a decrease in excitation supplied by the motor cortex (Gandevia, 1998; & Taylor, Todd, and Gandevia, 2006). The causes for this are poorly understood however it has been suggested that depletion of brain neurotransmitters, in particular serotonin, may be an underlying factor. Newsholme, Acworth and Blominstrand (1987) suggest that prolonged physical activity increases the brain's serotoninergic activity, causing a reduction in central command, and thus a reduction in recruitment of motor Units. Because serotonin cannot cross the blood-brain barrier it must be synthesized in the brain from its pre-cursor tryptophan (TRP). At the blood-brain barrier plasma free tryptophan and branch chain amino acids (BCAA) compete for access into the brain. Therefore as the ratio of TRP to BCAA increases so too does the synthesis of serotonin (Newsholme et al., 1987). This increase occurs during prolonged exercise as BCAA's are used by the muscle to provide energy, thus reducing the level of free BCAA's circulating in the plasma (Boyas and Guevel, 2011). Other research confirms that serotonin plays a significant role in the commencement and continuation of exercise and the development of central fatigue (Davis and Bailey, 1997; & Meeuson et al, 2006). Glycogen may also play a role in central fatigue as brain activation is linked to a drop in brain glycogen (Dalsgaard et al, 2003; & Nybo, 2003). As the brain has low glycogen stores they are quickly exhausted during exercise. The depletion of these stores can have a significant effect on brain function and serotonin activity, thus influencing central fatigue (Bequet et al, 2002). At a spinal level, a decrease in motoneuron activity involves inhibitory afferents from intramuscular receptors (Boyas and Guevel, 2011). According to Bigland-Ritchie et al (1986) motoneuron discharge rate can be controlled by peripheral reflexes in reaction to fatigue induced metabolic variations within the muscle. Metaboreceptors are stimulated by ischemia (Lagier-Tessonnier, Balzamo, Jammes, 1993), hypoxemia (Arbogast et al, 2002) and the accumulation of lactate (Darques, Decherchi, Jammes, 1998). Stimulation of these intramuscular metaboreceptors during fatigue may inhibit the activity of motoneurons at the spinal level (Martin et al, 2006). Other receptors, such as neuromuscular spindles, which run parallel to muscle fibres, may also play a role in restricting motoneuron activity (Gandevia, 2001). These afferent receptors provide the nervous system with information on muscle length and change in length during exercise (Proske and Gregory, 2002). According to Macefield et al (1991) discharge rates from these spindles progressively decrease with fatiguing contractions, thus limiting motoneuron activity (Bongioanni and Hagerbarth, 1990; & Gandevia, 2001). Sensitivity of neuromuscular spindles can become reduced with structural changes within the muscle, which can result from fatiguing contractions and stretching (Avela, Kyrolainen, and Komi, 1999), adding to decreased motoneuron activity. In addition, changes in muscle stiffness which accompany fatigue can lead to alterations in muscle electrical activity and contribute to fatigue; however mechanisms underlying this are not fully understood (Nordez et al, 2009).

As explained, there are many anatomical sites and physiological processes involved in the reduction of motoneuron activity, leading to a decrease in voluntary muscle activation, and the development of central fatigue.

Peripheral fatigue involves changes in neuromuscular transmission, muscle action potential proliferation, excitation-contraction coupling and contractile mechanisms resulting in a reduction in the contractile strength of muscle fibres (Boyas and Guevel, 2011). Muscle action potentials are generated when presynaptic potentials (nerve potentials) exceed the muscle cells excitability threshold. Neuromuscular transmission includes the conversion of nerve action potential into muscle action potential and occurs at the neuromuscular junction (Boyas and Guevel, 2011). Fatigue can alter this process through a number of mechanisms such as: inadequate transmission of the nerve potential at the nerve endings, failure of the coupling between excitation and neurotransmitter secretion in the synaptic gap, reduction in the levels of available neurotransmitter, reduced neurotransmitter secretion, and reduction in the sensitivity of post-synaptic acetylcholine receptors (Boyas and Guevel, 2011; Allen, Lamb, and Westerblad, 2008; & Sieck and Prakash, 1995). All of these mechanisms ultimately lead to a decrease in the contractile ability of muscle fibres and force generation.

During exercise, both central and peripheral fatigue will result in an effort from the neuromuscular system to adapt in order to sustain force generation (Boyas and Guevel, 2011). This involves a change in motor unit activity resulting in an increase in motor unit recruitment and alteration of motor unit discharge rates (Boyas and Guevel, 2011).

In order to understand how fatigue may effect impact accelerations and the presentation of overuse injuries in running it is important to have a clear picture of the processes involved in neuromuscular fatigue.

The effect of neuromuscular fatigue in running, on impact loading has been examined on numerous occasions. Verbitsky et al (1998) demonstrated that tibial impact acceleration values increased significantly (48%) following a 30min run to fatigue. This was associated with a subsequent decrease in stride rate (increased stride length); possibly explaining, in part, the increase in impact acceleration (as discussed in section 2.2.2.6) (figure 2.9). This is supported by Voloshin et al (1998) who demonstrated impact accelerations at the tibia and sacrum increased significantly by 60% and 35% respectively, following a run to fatigue.

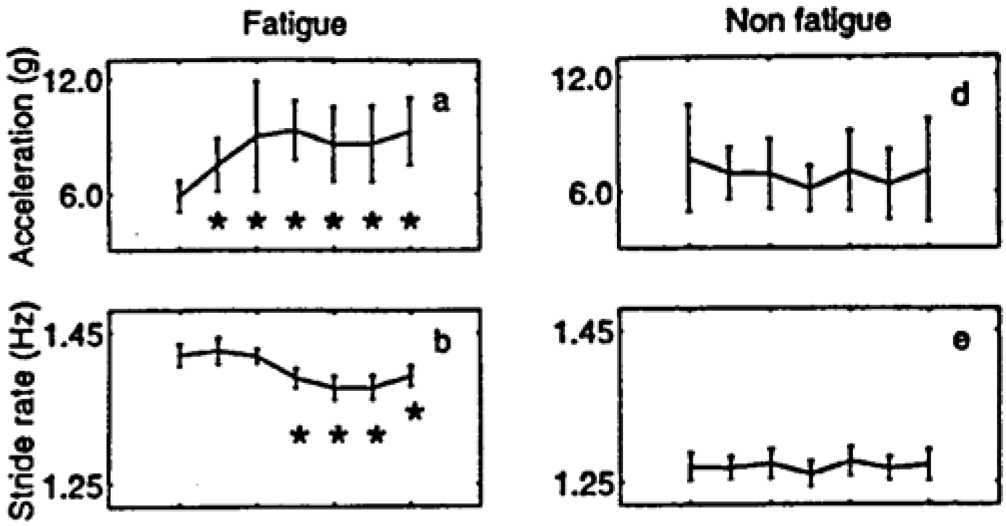

Figure 2. 9 Effect of fatigue on impact accelerations and stride rate (Verbitsky, 1998)

Mercer et al (2000) also demonstrated an increase in tibial impact accelerations following a run to fatigue (increased by 140%). This increase in impact acceleration was associated with a decreased stride rate (by 5%), decreased knee flexion angle (by 46%), increased vertical oscillation (by 19%), and decreased knee joint deformation during impact (by 51%). Therefore, the increased loading experienced as a result of fatigue may be explained by kinematic changes caused by the fatigued condition; the largest of which appears to be an increase in knee joint stiffness (evidenced by decreased join deformation) and an increase in vertical oscillation. Further supporting this, Clansey et al (2012) demonstrated an increase in VALR, VILR, and peak head accelerations (21%, 23%, and 25% respectively) following a run to fatigue. This increased loading was associated with increased hip-extension and a more plantar-flexed ankle, however no change in kinematics was observed at the knee (Clansey et al., 2012).

From the above evidence it is clear that fatigue may increase the risk of injury development during running. It is therefore important to understand how a novel running style that may increase energy expenditure might react under fatigued conditions.

2.3 Rating of Perceived Exertion

Rating of Perceived Exertion (RPE) is a psychophysiological measure that attempts to rate physical strain on a 15-point scale containing verbal anchors (Borg, 1970). The overall rating represents conscious perception of effort and integrates numerous sources of information such as signalling from peripheral working muscles, central respiratory and cardiovascular response, and central nervous system signalling (Borg, 1982; Crewe et al., 2008). Maximum RPE has subsequently been associated with fatigue (Crewe et al., 2008). Fatigue is therefore not purely a physiological event but is a conscious sensation that results from evaluation of subconscious regulatory processes (such as those mentioned above) that occur in the brain (Noakes et al., 2005). It is the conscious perception of effort that links the physiological parameters affected by exercise and the subsequent behavioural change that determines volitional fatigue (Crewe et al., 2008). Numerous studies have displayed a linear relationship between RPE and exercise duration whereby maximum RPE is representative of volitional fatigue and exercise termination (Crewe et al., 2008; Hortsman et al., 1979). Furthermore, Crewe et al (2008) demonstrated that time to volitional exhaustion is inversely related to the absolute rate of RPE increase; whereby increased time to volitional exhaustion is indicative of a reduced absolute rate of RPE increase (Crewe et al., 2008). Higher ratings of perceived exertion therefore act to discourage the continuation of exercise to the point where a catastrophic failure of homeostasis and subsequent damage to the human body would occur (Noakes and St Clair Gibson, 2004, Noakes et al., 2005; St Clair Gibson and Noakes, 2004).

The main physiological processes associated with central signals of exertion, and therefore RPE, are heart rate, pulmonary ventilation, respiratory rate, and oxygen consumption (Robertson, 1981). Each of these variables have been correlated with RPE: heart rate displaying correlation coefficients of 0.42-0.94 (Smutok et al., 1980; Borg, 1970; Robertson, 1982; Bar-Or et al., 1986; Borg et al., 1987), pulmonary ventilation and respiratory rate displaying correlation coefficients ranging from 0.61-0.94 (Borg et al., 1987; Robertson, 1982a; Robertson 1982b; Van De Burg et al., 1986), and oxygen consumption displaying correlation coefficients ranging from 0.76-0.97 (Borg et al., 1987; Robertson, 1982; Van De Burg et al., 1986). RPE has also been associated with blood lactate concentration (Steed et al., 1982), and core temperature (Crewe et al., 2008) and has been demonstrated as an accurate tool for prescribing exercise intensity in running and cycling (Dunbar et al., 1992).

2.4 Factors affecting the energy cost of running (running economy)

Running economy (RE) is determined by measuring steady-state oxygen consumption at submaximal speeds, and is universally used as a measure to determine the efficiency of runners. Runners with good running economy use less energy and therefore less oxygen than a runner with poor running economy (at the same speed and taking mass into consideration) (Nummela, Keranen & Mikkelsson 2007, Saunders et al. 2004). It can therefore also be used as a measure of the efficiency of different running styles. However, the amount of energy expended per unit time (cost of locomotion (COL)) and the amount of energy expended per unit distance (cost of transport (COT)) provide specific ecological meaning, and may therefore be more appropriate for providing information with regard to the energetic cost of running, than simply detailing the amount O_2 consumption at a set speed for different styles (Steudel-Numbers & Wall Scheffler, 2009). O_2 consumption data can be used to calculated calories (kcal) by using Weir's (1949) standard calculation, thus facilitating the measure of COT and COL.

Determining the relative efficiency of a novel, more compliant, running style is important for a number of reasons. Firstly, it may have direct implications in how likely a person may be to adopt such a style; if it is less efficient runners targeting improved performance will likely not consider such a change, regardless of potential reduced loading and injury risk. However, depending on the goal of exercise, increased energy expenditure may yield additional health benefits. In fact, research indicates that increasing energy expenditure by 1000 kcal a week may increase life expectancy by 20 % (Warburton et al., 2006). Furthermore, an average of 2000kcal expended during physical activity (a week) is associated with a decrease in morbidity and mortality of 20-30% (Lee & Skerret, 2001). Therefore, determining the energetic cost of a running style has important implications that must be considered.

In order to assess the energetic variance between different running styles it is necessary to understand all factors that may contribute to any change in running economy, and thus COT and COL. Considering the aim of the present project is to manipulate the kinetics and kinematics (biomechanics) of an individual's running style, these factors must be clearly understood. Given that both COT and COL are determined by a participants running economy, any kinetic or kinematic variable that may influence RE (and discussed in the following sections), will also influence COT and COL.

2.4.1 Biomechanical factors affecting Running Economy

There are numerous biomechanical factors that influence the amount of energy a runner expends. These include both kinematic and kinetic variables (Anderson 1996) such as stride length, stride frequency, stiffness, running speed, vertical oscillation of the COM, and ground reaction force variables and will thus be discussed in the following sections.

2.4.1.1 Stride Length and frequency and energy expenditure

Cavanagh and Williams (1982) experimentally altered stride length in 10 recreational runners using a metronome while running on a treadmill. Participants completed 6 minutes of running at seven different stride lengths (self-selected, plus and minus: 6.7%, 13.4, and 20%; stride length represented as % of leg length) at 7 minute-mile pace. Results showed a mean increase of 2.6 $ml.kg^{-1}.min^{-1}$ and 3.4 $ml.kg^{-1}.min^{-1}$ at the shortest and longest stride lengths (±20%), respectively. This is supported by Högberg (1952), who demonstrated increases in O_2 consumption of 12% and 19% at longer stride lengths (relative to self-selected) and an increase of 4% and 6% at shorter stride lengths, at speeds of 14 and 16km/hr, respectively. Similarly, Heinert et al (1988) found that altering stride length by ±8% yielded significant increases in RE by 2.1% and 3.8% at shortened and lengthened stride lengths, respectively (Heinert, Serfass & Stull 1988). Thus, current literature indicates that increasing stride length beyond that of self-selected has larger energetic consequences than shortening stride length. For the performance athlete maintaining a stride length as close to optimal will be an important factor in determining success. However, if health is the main concern, increasing or decreasing stride length (without introducing aberrant forces) may provide advantageous health benefits, as increased O_2 consumption would yield a greater amount of energy expenditure per unit time. Given that increased stride length is associated with decreased stride frequency (Heidershcheit et al., 2011), it is logical that frequency also plays an important role in determining energy expenditure. In fact, it is alterations of frequency (via a metronome) in the above studies that brought about the changes to stride length. Thus longer stride lengths indicate a reduced stride frequency and shorter stride lengths indicate larger stride frequencies, and subsequently effect energy expenditure as described above. Further evidence for the association between energy expenditure and stride frequency comes from data indicating that children expend more energy (at the same speed) than adults due in part to an increased stride frequency as anthropometric and physiological characteristics limit their capacity to generate increased stride length (Krahenbuhl and Williams, 1992; Viswanath et al 1990).

2.4.1.2 Vertical oscillation of COM and energy expenditure

As previously discussed in relation to injury development, manipulations of stride frequency and stride length can bring about subsequent changes to the magnitude of vertical oscillation of COM (Heiderscheit et al., 2011) and thus may also have implications for energy expenditure. Heiderscheit et al (2011) indicated that as stride frequency increased from -10% to +10% of preferred frequency, stride length decreased by 38%, and vertical oscillation of the COM decreased by 28%. Considering the above literature (section2.3.1.1) this appears to indicate altering vertical oscillation of COM will also alter energy expenditure. This is supported by literature that indicates more economical runners display less vertical oscillation (Cavanagh, Pollock & Landa 1977, Gregor, Kirkendall 1978). However, given that energy expenditure has also been shown to increase as a result of reduced stride length (Heinert et al., 1988, Högberg 1952), which would bring about reduced vertical oscillation, this would suggest there may be an optimal vertical displacement for efficiency and deviation from this optimal may increase energy expenditure. This was supported by McMahon et al (1987) who demonstrated that by almost completely eliminating vertical oscillation of COM (via kinematic alterations to knee angle at contact and during stance) there is an associated increase in energy expenditure of up to 50%. This is probably due to a diminished capacity for maximising the enhancements associated with the stretch shortening cycle [e.g. storage and release of elastic energy via the stretch shortening cycle (section 2.3.1.4 and 23.1.6)].

2.4.1.3 Running speed and energy expenditure

Given that running speed is largely determined by a combination of stride frequency and stride length, it is logical to assume that speed may play a role in determining energy expenditure. A study examining the biomechanical factors affecting running economy examined 17 endurance runners and found that as running speed increased from 3.25-6.25ms^{-1} energy expenditure (measured via steady state O_2 consumption) increased linearly, represented by an increase of 44% between the slowest and fastest speeds (Kyrolainen et al., 2001). This was also associated with an increased stride length of 54% and an increased stride frequency of 38%. Furthermore, Kyrolainen et al (2001) demonstrated that intra-individual differences in energy expenditure increased as speed increased, indicating that inefficient movement patterns may be more costly at higher speeds. Thus, if a running style displays increased energy expenditure relative to "normal" running [e.g. Groucho running (McMahon et al., 1987)] this increase may become exaggerated at faster speeds.

2.4.1.4 Stiffness and energy expenditure

During running, leg stiffness is required in order to effectively utilize the SSC (as described in section 2.3.1.6) (Latash and Zatsiorsky, 1993). Leg stiffness is a measure of the resistance to joint deformation and is therefore influenced by muscle activation, joint angle, and the magnitude of joint moments (Denoth, 1986; Bobbert et al., 1991). Subsequently, it has been demonstrated that increased stiffness is associated with decreased energy expenditure (McMahon and Cheng, 1990; Kerdock et al., 2002; Dutto and Smith, 2002; Heise and Martin, 1998). Kerdock et al (2002) manipulated lower extremity stiffness by altering surface compliance and demonstrated that a 29% increase in stiffness was associated with a 12% improvement in RE. Further support for this comes from research demonstrating significant decreases in vertical stiffness of up to 9% following a run to exhaustion (Dutto and Smith, 2000). This was also significantly and positively correlated with stride frequency (r=0.85), further highlighting the influence on energy expenditure.

As previously discussed (section 2.2.2.4) reduction of lower limb stiffness may be beneficial in regards to injury development. Therefore, decreasing stiffness in order to bring about reductions in the load associated with foot contact may have energetic consequences. Support for this comes from McMahon et al (1989) who demonstrated an increase in energy expenditure by up to 50% as a result of decreased lower limb stiffness. Decreased stiffness has also been associated with increased stride length and decrease stride frequency (Derrick et al., 2000), which have also been shown to increase energy expenditure (Cavanagh and Williams, 1982). However, Farley and Gonzalez (1996) demonstrated that as stride frequency increased from -26% to +36% (relative to self selected frequency) and stride length decreased, there was an associated increase in leg spring stiffness of 80%; indicating that there may be an optimum amount of stiffness that facilitates a balance between stride frequency and length that is most efficient and deviation from this may result in increased energy expenditure. Farley and Gonzalez (1996) also indicated that as stiffness increased, vertical oscillation of the COM decreased by 72% (from the lowest to highest stiffness) and that ground contact time decreased by 32%. Given that runners with better RE have been shown to display smaller vertical oscillation of the COM (Cavanagh, Pollock & Landa 1977, Gregor, Kirkendall 1978) this further supports the role of stiffness in relation to the energy cost of running. Furthermore, consideration of SSC mechanisms indicates that reduced contact time would facilitate more efficient use of the SSC and thus reduce energy expenditure (Bonacci et al. 2009, Divert et al. 2005, Spurrs, Murphy & Watsford 2003).

To conclude, it appears that increased lower limb stiffness reduces energy expenditure by facilitating greater utilization of the SSC. Furthermore, increased stiffness is associated with a number of other kinematic variables (decreased contact time, increased stride frequency, decreased stride length, and decreased vertical oscillation of COM etc) that have also been shown to influence energy expenditure and dictate the efficiency of a running style.

2.4.1.5 Ground reaction force variables and energy expenditure

Given that the magnitude of force applied to the ground during running plays a large role in determining stride length, which subsequently influences energy expenditure, it is logical to assume that the magnitude of the ground reaction force may be an important factor to consider with regard to energy efficiency. It is therefore not surprising that supporting body mass during running and the associated muscle force required to do so has been shown to be a major determinant of the metabolic cost of sub-maximal running (Chang, Kram 1999). Unsurprisingly it has consequently been demonstrated that the vertical ground reaction force is largely responsible for this energetic cost (Taylor et al. 1980, Farley, McMahon 1992, Kram, Taylor 1990b).

Heise and Martin (2001) investigated the role that ground reaction force characteristics play in regards to variations in running economy between participants. Results demonstrated a 27% difference in running economy between the least economical and most economical runners. Total vertical impulse was found to be significantly correlated with running economy (O_2 consumption at a given speed increased with increased total vertical impulse, r=0.62) and accounted for 38% of the variance for between-participant RE. This is in agreement with Farley and McMahon (1992) who demonstrated that RE improved by 25% as a result of a 25% decrease in vertical ground reaction force. Total vertical impulse represents a measure of the magnitude of the vertical ground reaction force and the time course over which it was applied. During foot contact with the ground, muscle activation occurs to ensure stability and maintenance of forward momentum. Total vertical impulse (TVI) may therefore be an indication of the overall muscular contribution during ground contact. Research examining muscle activation patterns during foot contact have suggested that economical runners display greater co-activation between bi-articular (two-joint muscles) muscles of the leg (Heise et al. 1996). This may be due to the fact that bi-articular muscles have been shown to facilitate greater neuromuscular transfer of joint rotations into desired external forces (van Ingen Schenau et al. 1992), such as ground reaction forces. This relationship may be further explained by research displaying improved RE in runners with reduced vertical oscillation of COM (Cavanagh, Pollock & Landa 1977, Gregor, Kirkendall 1978); as reduced vertical oscillation would result in reductions in vertical ground reaction force due to manipulation of the impulse-momentum relationship. However in opposition to this, it should be noted that running styles that act to reduce the magnitude of vertical ground reaction forces by reducing lower limb stiffness may act to increase energy expenditure (as seen in McMahon et al (1989)) despite an associated reduction in vertical oscillation. Therefore, it may be logical to assume that reduced vertical ground reaction forces may be associated with reduced energy expenditure if lower limb stiffness remains unchanged. Net vertical impulse was also found to have a significant positive correlation with RE and results indicate that it may be responsible for 36% of between-participant variance in RE (Farley and McMahon, 1992). Net vertical impulse is a reflection of vertical motion. This relationship would therefore support previous suggestions(Anderson 1996, Williams, Cavanagh 1987) that runners who display less vertical oscillation are more economical.

Very little research exists with regard to the effect of horizontal ground reaction forces on energy expended during running, however Chang and Kram (1999) used applied horizontal forces (AHP) of varying strength (-6% -3%, 0%, +3%, +6%, +9%, am +12%, and +15% of body mass) to examine this relationship. Results indicated that application of horizontal forces caused a significant ($P<0.0001$) effect on RE. This was represented by an increase of 30% in O2 consumption (at the same speed) at -6% AHP when compared to the 0% AHP control group. In contrast RE improved with increased positive AHP by 23%, and 33% for +6% and +15% AHP, respectively(Chang, Kram 1999). This reduction in energy expended was associated with a 70% decrease in horizontal propulsive impulse, indicating that the metabolic cost of generating horizontal force may be costly. These results are somewhat in agreement with data presented by Pugh (1971) who found that negative horizontal forces, in the form of wind, increased O_2 consumption by up to 13% during running and further research that has indicated that the energetic cost of running increases with an increase in external work (as a result of running against an impeding horizontal force via a harness). However, this does not agree with Heise and Martin (2001) who found no correlation between RE and anterior-posterior impulse. To this author’s knowledge, no other research has examined the effect of positive applied horizontal forces on RE.

Interestingly, Chang and Kram also displayed results indicating that the improvement in RE associated with +15%AHP occurred despite an increase in horizontal impulse of 173% during the braking phase of stance. Also, at -6%AHP where metabolic cost of running increased by 30%, horizontal braking impulsive decreased by 51% and propulsive impulsive increased by 47 %; thus, indicating that the generation of horizontal propulsive forces during sub-maximal running is much more expensive in terms of energy expenditure per unit force than that of horizontal braking forces. From this data, Chang and Kram (1999) calculated that generating 1 Newton of horizontal propulsive force on the ground costs 4.6 W, when running at 11.9km/hr under steady-state conditions. In comparison, Farley and McMahon (1992) found that a decrease in average vertical forces of 25% was associated with a 25% improvement in RE. Therefore, assuming the same relationship as found by Farley and McMahon (1992), Chang and Kram (1999) calculated that for their data, generating 1N of vertical force is equivalent to 1.2 W. This suggests that horizontal propulsive forces may be 4 times more energetically expensive than that of the generation of vertical forces.

The preceding paragraphs present information that strongly supports that generating larger vertical and horizontal forces increases the energetic cost of running.

The biomechanical factors that influence running economy are summarised in table 2.1.

Table 2. 1:Biomechanical factors that influence running economy.

Factors	Influence on Running Economy (RE)	Supporting literature
Stride length (SL)	Deviation from optimal: ↑RE* ↑SL =↑RE ↓SL =↑RE (*↑SL increasing RE to a greater extent than ↓SL)	Cavanagh & Williams (1982) Högberg et al (1952) Heinert et al (1988) Heiderscheit (2011)
Stride frequency (SF)	Deviation from optimal: ↑RE* ↓SF=↑RE by 4-20% ↑SF=↑RE by 2-6% (*↑SF increasing RE to a lesser extent than ↓SL)	Cavanagh and Williams (1982) Högberg et al (1952) Heinert et al (1988) Heiderscheit (2011)
Vertical oscillation of COM	↓Vertical oscillation=↓RE* (*When stiffness remains constant)	Heiderscheit (2011) Cavanagh et al (1977) Gregor et al (1978)
Stiffness	↑Stiffness= ↓RE	McMahon et al (1989) Kerdock et al (2002) Dutto et al (2000) Heise & Martin (1998)
Vertical Ground Reaction Force	↑vGRF= ↑RE	Taylor et al (1980) Farley et al (1992) Kram & Taylor (1990) Heise & Martin (2001)
Speed	↑Running speed: ↑RE	Kyrolainen et al (2001)

2.4.1.6 Stretch shortening cycle and energy expenditure

It is well established that the requirement for energy utilization through muscular contraction during the propulsion phase of running is greatly decreased through the body's ability to utilise the Stretch Shortening cycle (Alexander 1991, Lohman III, Balan Sackiriyas & Swen 2011).

The stretch shortening cycle (SSC) is characterized by an eccentric muscle action followed by an immediate concentric action (van Ingen Schenau, Bobbert & Haan 1997). In running the landing phase represents the eccentric action and the propulsion phase represents the concentric action (figure 2.10). The increased force production observed in the concentric action of a SSC contraction (relative to concentric contractions without a pre-stretch) can be explained by numerous mechanisms. Firstly, the load generated via this eccentric contraction is transferred to the series elastic component of the muscle tendon complex and stored as elastic energy. When this eccentric contraction is immediately followed by a concentric action, this stored elastic energy is released also causing an increase in force production (Asmussen, Bonde-Petersen 1974, Komi 1992, Nicol et al 2006). Secondly, during an eccentric muscle contraction the proprioceptive organs within the muscle, called muscle spindles, detect a rapid stretch, resulting in a consequent reflexive muscle action and increased force production (Dietz, Schmidtbleicher, and North, 1978). This reflexive action is also suggested to increase muscle stiffness via increased muscle activity during the eccentric phase (Hoffer and Andraesson, 1999). High muscular action during the eccentric phase is a pre-requisite for efficient storage of elastic energy (Komi and Gollhafer, 1997) and thus reflexive action appears to contribute to the utilization of elastic energy in this regard. Thirdly, it is suggested that the eccentric phase of an SSC provides the muscle time to develop maximum force in comparison to an isolated contraction (Bobbert et al., 1996, Chapman and Sanderson, 1990, Asmussen Sorenson, 1971). For example, it can take up to 300-500ms to reach maximum force output in a leg extension exercise (Bobbert and Ingen Schenau, 1990, Komi 1979). Therefore, if force development is only initiated at the start of the concentric action there is less time to develop maximal force, thus, it is suggested that the eccentric phase allows muscle the time to build up to a more active state (Bobbert and Harlaar 1993, Bobbert and Ingen Schenau). Lastly, the pre-stretch appears to have a potentiation effect on the muscle whereby the properties of the contractile machinery (cross-bridges) are altered, subsequently enhancing force production (Ingen Schenau et al., 1997, Cavagna, 1968). This may be due to the cross-bridges being detached (post eccentric load) to a point whereby they can re-attach more rapidly, relative to an isolated contraction (Woledge and Curtin, 1993). Therefore, any kinetic or kinematic parameters of running style that may influence the SSC will subsequently influence the amount of concentric action required to generate force, and thus the metabolic cost of running.

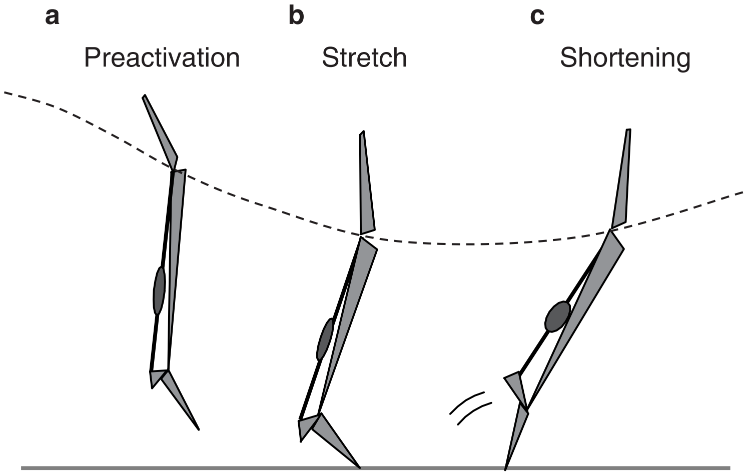

Figure 2. 10: Stages of the stretch shortening cycle.

Key factors that contribute to an effective SSC include pre-activation of muscles prior to eccentric contraction (prior to ground contact in running), a short and fast eccentric phase, and a short coupling period (time between the eccentric and concentric contractions) (Komi, Nicol 2000). Muscle pre-activation largely influences lower limb stiffness thus explaining the association between stiffness and RE (as discussed in section 2.2.2.2). Furthermore, a short coupling period results in a larger transfer of elastic energy as it decreases the amount of energy lost as heat, as well as increasing potentiation of contractile components and reflexive action (Cavagna, 1968, Edman et al., 1978, 1982, Nicol et al., 2006, Komi 1992). The coupling phase is largely associated with the ground contact time in running; thus, reduced contact time has been associated with improved RE. Therefore, appropriate use of the SSC during running can elicit greater force production during propulsion, thus reducing the contractile demand of the concentric phase, improving gait efficiency and reducing the overall energy demand of running(Bonacci et al. 2009, Divert et al. 2005, Spurrs, Murphy &

Watsford 2003).

To conclude, it is clear that the energetic cost of running is not only determined by the muscular contraction necessary to decelerate the body's COM during initial contact and to subsequently propel the body forward, but is also affected by efficient use of the SSC. Furthermore any kinetic or kinematic factors that may influence utilization of the SSC may further influence the overall energetic cost of running.

2.5 Novel running styles and their influence on injury development and energy expenditure

Various different running styles have been suggested throughout the literature to have both injury preventative and/or energetic benefits. So far both forefoot/barefoot running and Groucho running have been discussed in relation to how they act to manipulate the impulse-momentum relationship. However, two other styles that have yet to be discussed are that of Pose running, and Chi running. In order to assess the need for the proposed gait interventions presented in this thesis it is necessary to gain an insight into these other techniques. This section will therefore discuss Pose running and Chi running, and will present further detail on Groucho running.

2.5.1 Pose Running

Romanov and Robson (2003) describe pose running as a style whereby the leg action most closely resembles that of a wheel in that, ground contact is brief and occurs close to the point below the centre of mass (COM), vertical oscillation is minimized, and a slight lean forward produces forward motion. This is achieved by emphasizing certain mechanical characteristics: (1) an S-shaped body pose at impact to encourage the storage and utilization of elastic energy, (2) a mid-foot strike pattern as close to below the COM as possible in order to minimize support time (3) vertical foot removal immediately after foot strike, (4) minimal use of arms, (5) and a small degree of forward trunk lean (Romanov 2002). Examination of the above kinematic characteristics of Pose running present a somewhat unclear picture in terms of the effect this may have on injury development. Considering these characteristics in light of the impulse-momentum relationship would indicate that reduction in vertical oscillation of COM would result in a reduced vertical landing velocity, thus acting to decrease impact force (as described in section 2.2.2.5). However, minimizing contact duration would appear to increase both impact force and loading rate (as described in section 2.2.2.3), thus having an opposite effect of reducing COM oscillation. It would therefore seem that the effect of Pose running mechanics on injury might be largely determined by which of the above factors plays a dominant role. If neither factor dominates it would seem that the kinetic effect of both characteristics would be essentially neutralized by the other, indicating no change in impact force, and thus injury risk. Numerous studies have implemented the above mechanical characteristic in an attempt to understand the effect that such changes may have in relation to both injury prevention and Running Economy (RE).

2.5.1.1 Pose running and variables associated with injury development

Only two studies have examined the effect of Pose running on loading, with Arendse et al (2004) reporting a reduction in loading and Fletcher et al (2008) reporting no effect. Arendse et al (2004) instructed natural heel toe-runners to run with both mid-foot, and Pose running mechanics (on different occasions). Results indicated that heel-toe running presented significantly larger peak vertical impact ground reaction forces than both Pose (heel-toe 43% larger) and mid-foot running (heel-toe 33% larger). This was associated with less vertical displacement of the COM (Pose = 0.05m; heel-toe = 0.09m; mid-foot= 0.08m) and heel marker (Pose = 0.28m; heel-toe = 0.38m; mid-foot= 0.36m), smaller stride length (Pose = 1.48 m; heel-toe = 2.20 m; mid-foot= 2.17 m), and larger knee flexion angle at contact (Pose = 31.5°; heel-toe = 27.3°; mid-foot= 27.2°). Consideration of these kinematic changes suggests that the decreased COM and heel marker displacement would decrease vertical landing velocity, and subsequently impact force and loading rate (via mechanisms described in section 2.2.2.5). Similarly, increased knee flexion at impact may promote joint deformation due to larger moment arms, relative to the point of rotation, subsequently reducing effective mass and impact force (as described in section 2.2.2.1). Literature also supports the fact that decreased stride length is associated with decreased impact forces (Derrick et al, 1998); thus, the kinematic properties associated with Pose running presented by Arendse et al (2004) seem to explain the decreased impact force and suggests that Pose running may decrease injury risk. However, Pose running is described as having a decreased contact time (Romanov and Robson, 2003). Although contact-time was not measured in this study, the above kinematic results are indicative of increased contact time, which would seem logical given the reduction in impact load. This is supported by both McMahon et al (1987) who demonstrated increases in contact time as knee flexion at contact increased and vertical oscillation of COM decreased, and by Derrick et al (1998) who showed that as stride length was reduced there was an associated decrease in vertical landing velocity of the heel, and an increase in impact duration. Thus, the running style examined by Arendse et al (2004) may decrease kinetic variables associated with running injury, however considering that the kinematics are indicative of increased contact time, does this truly represent Pose running mechanics? This may somewhat explain the disparity between results presented by Arendse et al (2004) and Fletcher et al (2008) who also examined Pose running and associated kinetic changes. Fletcher et al (2008) demonstrated that after heel-toe runners completed a Pose running intervention there was an associated reduction in contact time by 15%, and an increase in stride frequency by 15%. However, in contrast to Arendse et al (2004), there were no associated changes in vertical impact forces, stride length, or vertical oscillation of COM. Thus it appears that the association between increase stride frequency and decreased impact load (Habaro et al, 2012) may be negated by the decreased contact time displayed for Pose running by Fletcher et al (2008), resulting in no net change in vertical impact loading. Further explanation of this disparity may come from the different the speed selection employed by both studies. Fletcher et al (2008) standardized speed whereas Arendse et al (2004) employed a self – selected speed protocol; resulting in participants running at slower speeds for Pose running (Pose= 2.9 m/s, Heel-toe= 2.98 m/s, midfoot= 3.06 m/s). It is therefore unclear whether or not the observed kinetic differences observed by Arendse et al (2004) are due to a change in mechanics or a simple change in speed or a combination of the two (section 2.2.2.7).

2.5.1.2 Pose running and running economy (RE)

Only two studies appear to have examined the effect of Pose running on RE. Dallam et al (2005) found that the pose running displayed significantly smaller stride lengths and vertical oscillations by 6% and 20%, respectively. This was associated with an increase in RE of 8%. This indicates that Pose Running is less efficient than each person's "normal" mechanics. Given the strong link between RE and running performance(Anderson 1996, Morgan, Craib 1992), it would appear that making the change to Pose running is not advisable for athletes. However, it has been suggested that Pose running can increase performance without any improvement in RE(Fletcher et al. 2008). Fletcher demonstrated that following a Pose running intervention participants demonstrated kinematics similar to experienced Pose runners (Fletcher, Bartlett & Romanov 2010). These kinematic changes were associated with no significant change in either RE or a 2400m time trial. However, although insignificant, the Pose running group presented a mean decrease of 25 seconds when comparing pre-post intervention time trial results. This is a clear and considerable improvement in performance, despite no change in RE. This indicates that there may be a different factor at play that may increase performance without an associated improvement in RE. Although these results are in disagreement with Dallam et al (2005) who found an increase in RE of 8% with Pose running, there are a number of factors that may explain this. Firstly, Fletcher et al (2008) implemented a study design whereby RE, time trial, and lab trials (kinetic and kinematic) were all conducted on separate occasions, thus it is unclear if the kinematic changes present during the lab trials were also present during both the time trial and during the RE measurement. Secondly, it is well established that running speed directly influences RE (section 2.3.1.3); in the two above studies RE tests were completed at differing speeds (12km/hr (Fletcher et al, 2008) versus 14.8 and 12.9 km/hr (Dallam et al 2005)). Given that both studies examined similar populations (sub-elite runners) this may indicate that Pose running may be less economical at faster speeds, and that the contributing factors to this inefficiency are less apparent at slower speeds. However, this is yet to be examined.

To conclude, clarity within the literature regarding the effect of Pose running on impact force variables associated with running injury appears to be clouded by versions of pose running that do not present the same kinematic characteristics and thus conclusions about its ability to reduce injury development are difficult. If contact time is reduced, there appears to be no effect on impact force variables and thus injury (as in Fletcher et al (2008)), however if contact time increases, as may be the case in Arendse et al (2004)(not directly measured but indicative of other kinematic changes), there may be a reduction in impact force variables and thus injury. However, given that pose running is described by its proposed founders as displaying decreased contact time (Romanov and Robson, 2003), it appears that it may not have an effect on impact force variables associated with injury. A similar cloudy picture is presented in relation to its effect on running performance. Pose running appears to show no positive effect on RE (Dallam et al., 2005; Fletcher et al., 2008), however may improve time-trial performance via mechanisms that are not yet understood.

2.5.2 Chi Running

Chi running is described as the alignment of body, mind and forward movement whereby runners avoid heel-strike, land with a foot-strike anterior to the heel (mid-foot/fore-foot), display a slight forward lean, stride length is decreased and where runners are encouraged to relax their legs (Goss, Gross 2013, Dreyer, Dreyer 2009). A search of Google Scholar on the 18h of April 2015 using the key words “Chi running” returned 134 of which only 1 research paper relevant to Chi Running was discovered. This paper examined lower limb biomechanics in Chi Running in comparison to running with a rear-foot strike (RS) in order to determine the difference (if any) in relation to variables associated with injury development(Goss, Gross 2013). Runners were subjectively judged to run with Chi mechanics if they met the following visual criteria: (1) Postural alignment in mid-stance (shoulders, hips, and ankles aligned), (2) hips slightly ahead of feet during mid-stance, (3) Knees bent on impact with no heel strike or dorsi-flexion, (4) no contraction of calves during terminal stance (i.e. no toe-off), and (5) lifting of ankles and not knees (knees bent but not lifted) (Goss, Gross 2013) . The Chi running group subsequently presented significant kinematic differences for stride frequency (P= 0.01, Chi > RS by 3%), ankle angle at initial contact (P=0.008, RS = dorsi-flexion of 2.6° versus Chi- plantar-flexion of 1.55°), and total knee joint excursion during stance (P=0.03 RS> Chi by 21%). These kinematic changes were associated with larger average vertical loading rates for RS runners (RS> Chi by 46%, P<. 001). However peak vertical ground reaction force, and ankle excursion during stance remained unchanged. Examination of the kinematic and kinetic changes in light of the impulse-momentum relationship somewhat explain the observed changes. Both increased stride frequency and a more forefoot running action have been previously associated with decreased loading rate (Lieberman et al., 2010, Edwards et al., 2009, Habaro et al., 2012). However, greater knee joint excursion for RS runners would indicate decreased joint stiffness, thus reducing effective mass, and impact load (Milner et al., 2007, Derrick 2004). It therefore seems that an increase in stride frequency and more plantar-flexed initial contact act to decrease loading rate to a greater extent than increased knee joint excursion in RS running. However, running speed in the RS group was 9.4% greater than in the Chi running group. Given the association between running speed and impact loading rate (Munro et al., 1987, Mercer et al., 2002) this may somewhat account for the increased loading rate observed in RS runners relative to the Chi running group. Examination of eccentric work at the knee (RS> Chi by 79%, $P< .001$) and ankle (Chi > RS by 39%, $P<0.001$) indicates that Chi running may shift the distribution of overall load from the tibia and knee to the forefoot and ankle. It appears that the increased knee extensor eccentric work and knee excursion in RS runners acts to attenuate vertical ground reaction force variables through the knee joint whereas increased plantar flexion work in Chi runners facilitates attenuation of vertical ground reaction force variables through the ankle joint.

2.5.3 Groucho Running

To date, only study has examined Groucho Running. McMahon et al (1989) sought to investigate how vertical compliance, and/or stiffness of a running gait determines various aspects of running performance such as attenuation of impact accelerations and energy consumption. Vertical stiffness was manipulated by requiring the participants to run with increased stance-leg knee flexion; acquiring the Groucho running style.

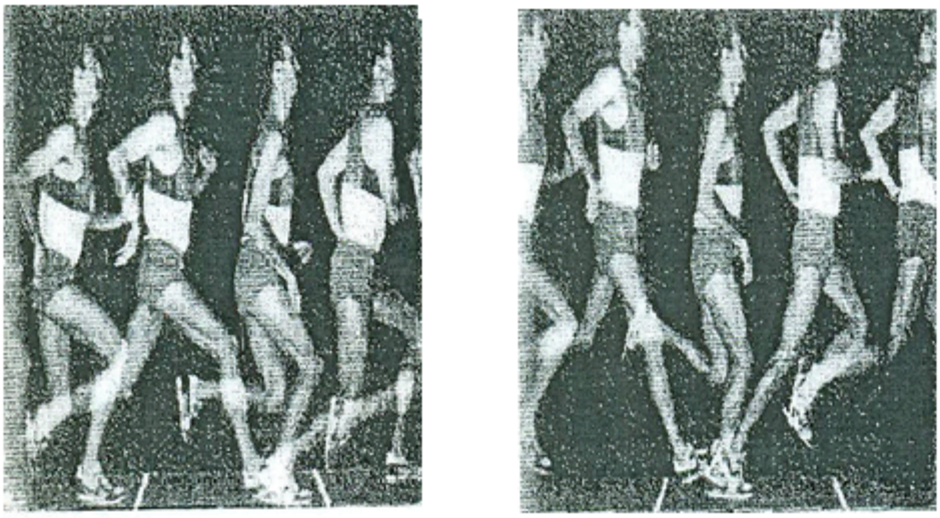

Figure 2. 11: Groucho running (left) versus normal running (right) (adapted form McMahon et al 1987)

McMahon et al (1987) highlighted that the hip follows a much lower trajectory with less up and down motion in Groucho running than normal running. Consequently a longer step length (horizontal distance the body moves over the ground during one foot-contact) is present in the Groucho posture. At lower speeds, in some participants, the aerial phase of each stride disappears completely. These kinematic changes brought about subsequent kinetic changes. In relation to impact accelerations McMahon et al (1987) found that peak vertical acceleration measured at the tibia was somewhat greater in Groucho running than in normal running (exact values not reported), but was smaller at the head. It is likely that the increased tibial accelerations are artificially inflated due to manipulations of effective mass. This is supported by the fact that the impact passive peak for the vertical ground reaction force did not change. Therefore Groucho running appears to be able to reduce the magnitude of load experienced further up the body. Since these impact accelerations have been shown to be a factor in the etiology of many running injuries this suggests a possible injury preventive characteristic of Groucho running. However, because McMahon et al (1987) only measured acceleration peaks at the tibia and head, the magnitude of impact peaks in between these two points and where exactly the found attenuation occurs is unknown; therefore measuring acceleration peaks at a third point (e.g. sacrum) may provide vital information on the injury preventative potential of Groucho running. This may be of particular importance as the hip and back are prevalent areas for running related injuries (Paluska, 2005; McMahon 1987; Lafortune et al, 1996). Interestingly, the kinematic changes associated with Groucho running were also shown to increase energy expenditure by up to 50%. This increase in energy expenditure was clearly associated with increased knee flexion angle at mid-stance; the greater the knee flexion angle at mid-stance the greater the energy expenditure (as can be seen below in figure 2.12).

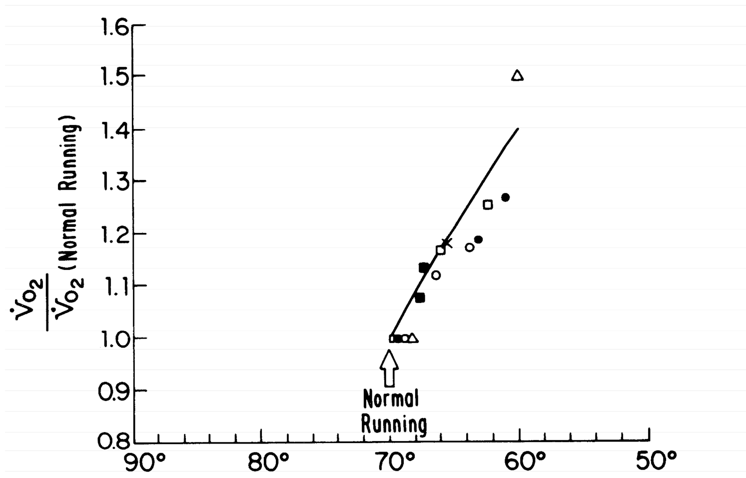

Figure 2. 12: Relationship between knee angle and energy expenditure (adapted from McMahon et al 1987)

This reduced efficiency may be due to increased muscular recruitment necessary to maintain a Groucho posture and a decrease in the utilization of the SSC that results from these mechanics. Although, this presents a possible negative for competitive runners or recreational runners trying to improve their performance, this increase in energy expenditure at any given speed, or for a given time period, may provide additional health benefits. Increasing energy expenditure by 1000 kcal a week has been shown to increase life expectancy by 20 % and burning an average of 2000kcal a week during physical activity is associated with a decrease in morbidity and mortality of 20-30% (Warburton et al, 2006; & Lee and Skerret, 2001). Thus an ability to reach these targets in a shorter time period, while protecting the body from excessive impact loads would be beneficial

2.6 Gait Re-training

In order for any novel running style to be suitably implemented in a practical setting it is imperative that an appropriate intervention is in place to alter a persons kinematics to replicate that of the desired style and facilitate motor learning. To date very few studies have employed the use of an accelerometer based biofeedback system; Crowell et al (2010,2011), Cheung et al (2011), and Clansey et al (2014) employing visual accelerometer based biofeedback, and Wood and Kipp employing auditory accelerometer based biofeedback. However, other forms of biofeedback have been employed, and will be discussed.

2.6.1 Biofeedback

Biofeedback is a method of providing augmented feedback (visual, auditory, or haptic) to a learner that typically involves the use of electronic equipment that will identify, augment, and convey instantaneous physiological/biomechanical processes or functions to which the learner is otherwise unaware (Onate, Guskiewicz & Sullivan, 2001; Femery et al., 2004; Tate & Milner, 2010). Therefore, this becomes particularly useful in situations where intrinsic and extrinsic feedback is lacking or absent (Femery et al., 2004); for example, in running where participants continuously run with kinematics that may produce aberrant or excessive loads, which over-time may result in injury. If a means was present to provide information on excessive loads it may be possible to adopt more compliant kinematics, thereby reducing the magnitude of loading, and potentially reducing injury occurrence. Numerous studies have implemented biofeedback strategies to alter gait mechanics and reduce the magnitude of variables associated with injury development and will therefore be discussed in the following sections.

2.6.1.1 Visual biofeedback and gait alterations

In the first of two studies Crowell et al (2010, 2011) proposed that following an acute bout of gait retraining using real-time tibial acceleration feedback, participants would be able to reduce their peak tibial acceleration immediately and 10 minutes post provision of feedback. To examine this, a uniaxial accelerometer was attached to the anteriomedial aspect of the distal tibia, on the right leg of five participants. Following a warm-up period of 5 minutes, accelerometer and force-plate data were collected for a period of 15 seconds. Immediately after the warm-up was complete a 10-minute bout of biofeedback began. The accelerometer signal was presented in clear view on a monitor situated in front of the treadmill. A horizontal line representing 50% of the mean impact acceleration peak values (individual to each participant) was placed across the accelerometer signal (figure 2.13). Participants were instructed to run softly and to decrease their acceleration peaks to below the line. Results indicated that 3 of the 5 participants were able to decrease the magnitude of impact accelerations experienced by the tibia by 33-50% while the other two participants recorded an increase of 20% and 30%. Following a further 10 minutes of running without biofeedback 4 of the participants showed decreases in tibial impact acceleration peaks ranging from 17-60%, with the fifth participant showing an increase of 6%, relative to baseline. Also, all 5 participants presented decreases in vGRF impact peaks of 6-24% and average loading rates of 16-38% at the end of the protocol, relative to baseline. It is clear from this study that tibial biofeedback can be used to decrease impact acceleration peaks, average loading rates, and vGRF impact peaks and that participants can maintain decreases following 10 minutes of running without feedback. This therefore presents itself as a viable gait-retraining tool that may decrease the development of stress fractures via reductions of variables that have been implicated in the development of such injuries (Davis, Milner & Hamill 2004, Bennell et al. 1999, Knobloch et al. 2007, Milner et al. 2006, Pohl et al. 2008) .

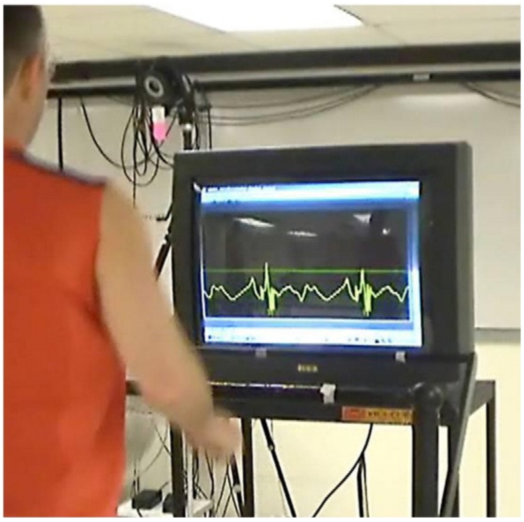

Figure 2. 13: Participants instructed to maintain tibial accelerations below the green line marking 50% of the individual's peak values during baseline measures

The successful use of tibial biofeedback to decrease load by Crowell et al (2010) raises the idea that it may be possible to provide biofeedback from accelerometers at other anatomical/off-body sites such as the sacrum or the treadmill. This has not previously been investigated and may provide important information in regards to the most appropriate method for providing accelerometer-based biofeedback for gait-retraining and subsequent reduction of injury risk. Furthermore, limitations to the work completed by Crowell et al (2010) highlight the need for further investigation of the gait-retraining method. Firstly, the extremely small sample size (5 participants) raises doubts over the generalizability of the found results to larger populations. Also, Crowell et al (2010) did not measure participant kinematics; it is therefore unclear what changes were made to each participants running style in order to decrease impact loads. Furthermore, Crowell et al (2010) only measured impact acceleration values at the tibia. It is therefore unclear what effect any change in kinematics had on impact acceleration values further up the body. Sacral impact accelerations have been implicated in the development of lower back pain (Collins, Whittle 1989) and therefore are an important factor that must be considered when developing a system to reduce impact related running injuries. The acute nature of Crowell et al's work (2010) also raises questions in relation to whether or not motor learning, as indicated by "*relatively permanent changes*" (Schmidt et al. 1989) to gait mechanics, can be produced.

The issue of motor learning was examined in a second study where 10 runners, all with baseline tibial impact acceleration values larger than 8g's, completed a two week intervention using tibial impact acceleration biofeedback (Crowell, Davis 2011). During the two-week intervention each participant completed 8 sessions, increasing in length incrementally from 15 minutes of running to 30 minutes of running. Tibial impact acceleration biofeedback was provided continuously for the first four sessions then decreased gradually, employing a faded feedback design. Impact acceleration, and ground reaction force data were collected before the start of the two weeks, at the end of the intervention, and 1-month post intervention. Results indicate a decrease in tibial impact acceleration values of 48% and 44% from pre-intervention measures to post-intervention, and pre-intervention to 1-month post-intervention measures respectively; thus indicating that motor learning may have occurred, as participants did not revert back to pre-intervention tibial impact acceleration values (Crowell, Davis 2011) . However similar to Crowell et al's previous study (2010), kinematics were not measured; it is therefore unclear what kinematic strategies participants utilized in order to facilitate the observed decreases in impact loading or if the kinematic strategies implemented during the two week intervention were the same as the strategies that presented similar decreases in loading at the 1-month follow-up. A further limitation to this study is that Crowell et al (2011) excluded participants that did not present tibial impact acceleration peaks larger than 8g's. It is therefore unclear if this gait-retraining protocol is applicable to participants that present baseline impact acceleration values less than 8g's, which appears to be common.

A further study implementing the same biofeedback protocol as Crowell et al (2011) was carried out in an attempt to identify some of the common kinematic characteristics adopted as a result of this biofeedback based intervention (Cheung, Rainbow, Altman, and Davis, 2011). Similar significant decreases of 45%, 32%, and 31% were observed for tibial peak positive acceleration (PPA), vertical average loading rating (VALR), and vertical instantaneous loading rate (VILR), respectively. These changes were associated with significant reductions in vertical stiffness during the impact phase of 28% and vertical touchdown velocity by 5%. Furthermore, decreased impact phase stiffness was significantly and strongly correlated with reduced tibial PPA($r=0.76$, $P<0.001$), VILR ($r=0.890$, $P<0.001$), and VALR ($r=0.903$, $P<0.001$). No other common kinematic strategies were identified. This may indicate that regardless of joint kinematics participants appear to modulate their centre of mass dynamics in order to reduce vertical stiffness and subsequently tibial PPA.

A similar study completed by Clansey et al (2014) implemented a slightly different tibial biofeedback system whereby participants received visual and audio based biofeedback, with regard to the magnitude of peak tibial acceleration values, every fifth stride, as opposed to the continuous nature of feedback implemented by Crowell et al (2010, 2011). The biofeedback protocol employed a traffic light system whereby high peak acceleration values (> 75% baseline) were portrayed by a red- light and high pitched sound, medium values (between 50% and 75% base) represented by an amber light and a low pitched-sound, and acceptable values (< 50% base) represented by a green light and no sound. After a 3-week intervention period of 2 sessions a week, participants displayed a reduction in peak tibial acceleration by 31%, VALR by 18%, and VILR by 19%. At a 1-month follow-up only peak tibial acceleration remained significantly reduced (1-month follow < baseline by 22%). It therefore appears this method of feedback and intervention design was less effective at implementing motor learning as that presented by Crowell et al (2011) who demonstrated only a 4% drop in the amount of tibial acceleration reduction (in comparison to post-intervention) relative to baseline at a 1-month follow-up. This may be due to the larger amount of absolute practice employed in Crowell et al (2011) (8 sessions across 2 weeks Vs. 6 session across 3 weeks). Furthermore, the magnitude of reduction presented by Crowell et al (2011) appears to be larger (- 48% Vs. -31%). Clansey et al (2014) further demonstrated that the reductions in load, associated with this biofeedback intervention (as above), were brought about by a change from a rear-foot strike pattern to a mid-foot strike pattern and a reduction in heel vertical velocity (-47%). However, no change was observed at either the knee or hip. Given the positive association between increased knee flexion and reduced loading (as described in section 2.2.2), the current thesis aims to direct participants towards a more compliant style via alteration to knee and hip mechanics by providing feedforward information in the form of simple written instruction before completing the biofeedback protocol. Interestingly, Clansey et al (2014) demonstrated no significant change in running economy associated with the above changes in loading and kinematic variables. Finally, it is unclear what effect the kinematic alterations associated with the 31% reduction in peak tibial acceleration, presented by Clansey et al (2014), may have on loads further up the body such as at the sacrum.

Examination of the literature surrounding gait re-training and re-education (in regards to both running and walking) appears to indicate that only four authors to date (Crowell, Davis 2011, Crowell et al. 2010, Cheung et al, 2011, Clansey et al., 2014) have specifically examined the use of visual accelerometer-based biofeedback to reduce impact loading in running. These studies have been discussed in detail above. However, other evidence to support the use of biofeedback to alter gait is available; particularly in relation to reduction of symptoms associated with patellofemoral pain syndrome (PFPS) and knee osteoarthritis.

Noehren, Scholz & Davis (2011) investigated the use of real-time biofeedback to improve running mechanics and reduce pain associated with PFPS. Ten runners diagnosed by a physiotherapist with PFPS, and hip adduction values greater than 1 standard deviation above the mean of a group of healthy recreational runners, entered a 2-week gait-retraining intervention. Excessive hip adduction has been prospectively (Noehren, Davis 2007) and retrospectively (Willson, Davis 2008) associated with the development of PFPS, thus explaining this inclusion criterion. Participants completed 8 retraining sessions over a two-week period. Each participant's hip adduction curve was displayed in real-time on a monitor in front of the treadmill. A shaded region on the real-time hip adduction curves represented ± 1 standard deviation of the mean hip adduction value for healthy runners. Participants were asked to keep their hip adduction angle within the shaded region by contracting their gluteal muscles, and keeping their knee's pointing forward, while maintaining a level pelvis position (Noehren, Scholz & Davis 2011) . Biofeedback was provided via a faded feedback design. Results indicated that participants were able to decrease hip adduction values by 23% post intervention, with subsequent decreases in pain of 86% (as according to the Lower Extremity Functional Index). A similar decrease of 20% (when compared to baseline) was observed for hip adduction values and 100% for pain values at a 1-month follow up, thus indicating that motor learning had occurred and the underlying mechanics implicated in the development of PFPS have been altered. Interestingly, given that hip adduction values increased slightly from post intervention to the 1-month follow up, whereas reduction in pain improved by 14% (to 100% when compared to baseline), this may indicate that there are other variables associated with the reduction of pain in PFPS. In fact, a significant decrease in impact peaks of 10% was observed, suggesting a possible association with impact loading.

In a research project specifically targeting running related injuries, Davis (2005) carried out a number of pilot/case studies utilizing real-time visual feedback/biofeedback in an attempt to reduce measures associated with participant specific injuries. The first of these studies examined a 40-year-old female runner suffering from plantar fasciitis. Gait analysis revealed that the injured runner presented hip adduction, hip internal rotation, knee abduction, and knee internal rotation values larger than that presented by a group of healthy runners. Davis (2005) hypothesized that the plantar fasciitis was caused by the internally rotated hip and medially deviated position of the knee, placing increased stress on the arch of the foot. The participant subsequently completed an 8-week gait-retraining intervention where visual feedback was provided by a mirror placed in front of a treadmill. The participant was verbally instructed to keep her knees apart, and keep her patella facing forward. Feedback was incrementally decreased across the 8-week period. Immediately post intervention and in a 6-month follow up, the participant presented significantly decreased hip internal rotation, hip adduction, knee abduction and increased knee internal rotation. As a result of these gait alterations the participant was running pain free for 30 minutes 3-4 times weekly (Davis 2005) .

The second of these studies completed by Davis (2005) examined a 46-year-old female runner with PFPS. Completion of a gait analysis revealed excessive hip internal rotation. The participant completed a 10-week gait-retraining intervention where visual biofeedback, detailing hip internal rotation curves, was provided. The participant was asked to lower her internal hip rotation curve without altering foot placement. Upon completion of the intervention the participant was able to reduce the amount of hip internal rotation during stance (details of amount not present). Subsequently the patellofemoral pain dissipated completely.

Biofeedback has also been employed to alter walking mechanics in participants suffering from knee osteoarthritis. Excessive joint loading has been implicated in the development of medial knee osteoarthritis (Andriacchi 1994) . Specifically, knee adduction moments (KAM) is used as a measure of the severity of the disease (Schipplein, Andriacchi 1991, Thorp et al. 2006, Prodromos, Andriacchi & Galante 1985) . In an attempt to alter gait mechanics, reduce knee joint loading and risk of medial knee osteoarthritis, Shull and Besier (2011) provided participants with either visual or tactile biofeedback. Visual biofeedback was presented in front of the treadmill in the form of a stair-step plot where a dotted line represented baseline measure of KAM (as can be seen in figure 2.14). KAM for each step and the previous 9 steps were displayed on the screen. Tactile feedback was provided in the form of a vibration. If peak KAM was 80% of baseline KAM the participant received a large amplitude vibration, if it was 60-80% baseline the participant received a low amplitude vibration, and if it was below 60% the participant received no vibration. On both accounts participants were asked to alter mechanics to reduce KAM values as much as possible while maintaining a comfortable gait. Participants were also provided with examples of written kinematic strategies and asked to experiment with each. Results indicated that both forms of feedback reduced peak KAM by 27%, thus reducing the risk of developing knee osteoarthritis. However, visual feedback appeared to facilitate faster uptake of consistent mechanics.

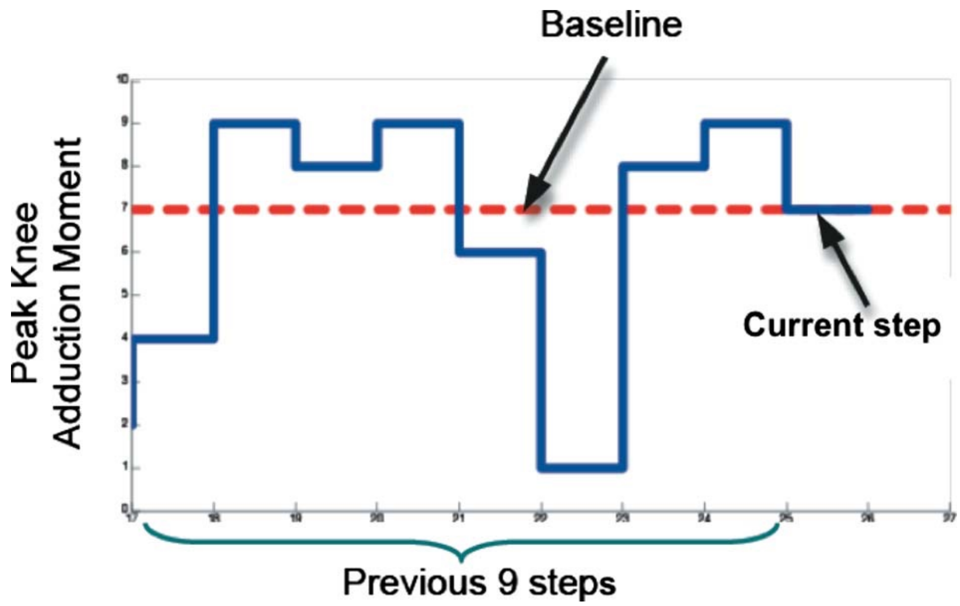

Figure 2. 14: Visual Biofeedback system used by Shull and Besier (2011)

2.6.1.2 Auditory biofeedback and gait alterations

Similar to Noehren et al (2011), Cheung and Davis (2011) examined the use of real-time biofeedback to reduce pain in participants with PFPS. However, the objective of this study was not to decrease hip adduction values (as in (Noehren, Scholz & Davis 2011)) but instead to decrease the magnitude of vertical impact peaks by providing biofeedback that attempted to alter patient's foot strike mechanics from a heel-strike to a forefoot/mid-foot strike. Justification for this lies in the fact that vertical impact peaks and vertical loading rates have both been shown to present higher values in participants suffering from PFPS (Davis, Bowser & Hamill 2010) by roughly 15% and 26% respectively (calculated based on graphs presented by Davis, Bowser & Hamill (2010)) and that forefoot/mid-foot strikers have demonstrated vertical impact peaks and loading rates up to 16% smaller than heel strikers (Altman, Davis 2010) . In order to alter foot-strike a force transducer was placed inside the shoe of each participant under the calcaneus. If the participant landed on their heel the force transducer emitted a warning beep. Participants were asked to eliminate this buzzer noise while they ran by avoiding heel-strikes and shortening their stride length. Participants decreased heel-strikes while running by 98% immediately post intervention and at a 3-month follow-up. Subsequently participants decreased their vertical impact peaks and loading rates by 11%-35%. This was associated with a mean reduction in pain of 75% (as according to the PFPS pain scale) immediately, which was maintained at a 3-month follow-up. This study offers further evidence to support the use of biofeedback to alter running mechanics, however the extremely small sample size (3 participants) limits the generalizability to larger populations. Furthermore, no kinematics assessment was undertaken.

Only one study to date has employed accelerometer based auditory biofeedback (Wood, Kipp 2014). Real-time feedback was provided to participants via a beeping noise that was activated if peak impact acceleration values breached a threshold set at 10-15% below baseline values. Accelerations that exceeded the threshold by a large amount produced a high-pitched beep whereas those closer to the threshold produced a lower pitch beep. Therefore participants were able to determine how close they were to the threshold by the pitch of the beep. If they remained below the threshold no beep was produced. Participants were instructed to alter mechanics while running to eliminate the beep or to reduce the pitch of the beep as much as possible. Baseline data were determined during a 5-minute warm-up. This was followed by 5 minutes of running with biofeedback, 5 minutes without, 5 minutes with, and a final 5 minutes without biofeedback. Participants significantly reduced peak tibial accelerations by 11% during the first biofeedback period and 13% during the second biofeedback period (relative to baseline). This decrease was somewhat maintained during the last no-feedback period as participants displayed a reduction of 9% relative to baseline measures. Therefore, it appears participants were able to respond to auditory feedback, reduce the magnitude of peak tibial accelerations and subsequently potentially reduce the risk of developing running related injuries. Although these reductions are significant, they are not as large as those presented using visual biofeedback (48% reduction) by Crowell (2011), indicating that visual biofeedback may be more effective at bringing about alterations to running style and reducing the risk of developing running related injuries. However, this may be due to the smaller target reduction employed by Wood and Kipp (10-15%) in comparison to Crowell et al (2010) (50%). This highlights a further benefit of visual biofeedback; even if the target reduction has been reached the visual presentation of acceleration peaks may facilitate reductions greater than the target. Similar limitations are present as have already been discussed in relation to the work completed by Crowell et al (2010,2011). Both the kinematic strategies employed by the participants to reduce PPA and the effects these alterations may have on loads further up the body (such as at the sacrum) are unknown.

To conclude, it appears that humans have the ability to respond to visual, auditory, and haptic (vibration) augmented feedback, subsequently altering running mechanics and reducing the risk of injury development or alleviating symptoms of already injured runners. Current evidence suggests that visual feedback mechanisms may be the most effective of the three aforementioned mechanisms. Furthermore, visual biofeedback appears to be able to impart motor learning when implemented via a faded feedback system (Crowell et al, 2011; Noehren, Davis 2007; Cheung et al 2011). However, little evidence is present to indicate common kinematic strategies employed in response to the various feedback mechanisms or what effect these kinematic changes have on other variables associated with injury. These are important considerations as once common kinematic strategies are discovered they may be used in combination with biofeedback to bring about desired changes more efficiently (less trial-and-error). Furthermore, kinematic strategies may not be useful if they simply redistribute load from one location to another, essentially shifting the risk of injury.

2.7 Conclusion

It appears there are numerous factors that may predispose a runner to the development of injuries. From a kinetic perspective an increased magnitude of vertical ground reaction force (passive peak), segmental accelerations, and joint loading appear to increase the risk of injury. Furthermore, it is clear that numerous kinematic factors can directly influence the magnitude of these kinetic variables via manipulation of the impulse-momentum relationship, acting to decrease velocity, effective mass, or increase contact time. Therefore, any running style that implements all or some of these kinematic strategies may influence magnitude of loading and thus serve to decrease the risk of injury development. It is also clear that the same kinematic mechanisms that act to decrease injury risk can also serve to increase the energy cost of running. Although not appropriate for competitive runners, an increase in energy expenditure offers numerous health benefits, including increased life expectancy. Finally, it is evident that humans have the ability to respond to biofeedback, alter kinematics, and reduce variables associated with running injury. Current data suggests that visual biofeedback may be more effective in this regard than either auditory or tactile biofeedback. The use of an accelerometer to provide biofeedback has clear advantages over other measures of loading; (1) it provides loading information specific to a location or segment that may be at risk, (2) it improves ecological validity (running is not confined to a lab), and (3) accelerometers are relatively cheap in comparison to lab based systems. Given the discussed limitations of current research employing accelerometer based visual biofeedback, there is clear justification for further investigation of this area.

Chapter 3: Study 1

The effect of directed compliant running on energy expenditure and impact loading in fatigued and unfatigued conditions.

3.1 Introduction

In Ireland, the popularity of running as a form of physical activity has had a greater than 2 fold increase from 2007-2011 (Irish sports council, 2011). However, based on injury incidence rates and the number of new Irish runners, this will cause an absolute increase of running related injuries of about 126% [based on the number of new Irish runners, population estimates from The Central Statistics Office (2012) and injury incidence rates of 29.5% -54.8% (Van Middelkoop, Kolkman, Ochten, Bierma-Zeinstra & Koes, 2008; Walter, Hart, McIntosh & Sutton, 1989; Buist et al., 2010; Taunton et al., 2003; Marti, Valder, Minder & Abelin, 1988)].

Collision with the ground during running generates high impact forces that travel through the foot and up the musculoskeletal system (Lafortune, Lake, Hennig, 1996). This force has been implicated in the development of numerous overuse injuries such as degenerative joint disease, spinal injuries, tendinitis, muscle tears, and stress fractures (Whittle, 1999; Lafortune et al, 1996; McMahon, G. Valiant, Frederick, 1987). Stress fractures are among the most severe (Brubaker & James, 1974) and most common (Taunton et al., 2002) running related injuries sustained, resulting in exclusion from running or any impact related activity for an average of at least eight weeks (Beck, 1998; Bennell and Brukner, 2005). Research has shown that significant decreases in musculoskeletal and cardiovascular function are observed following eight weeks of reduced training (Coyle et al., 1984; Coyle et al., 1985). Given the high incidence of running related injuries, the health benefits associated with running, and the detrimental health effects of inactivity, developing a method of reducing running injuries is a priority.

Only one study to date appears to have specifically directed participants to run more compliantly by increasing knee flexion. McMahon et al (1989) details a running style that involves a greater (than normal) degree of knee flexion at foot contact, and refers to it as "Groucho Running". Results indicate that increasing knee flexion at foot contact attenuates the transmission of impact accelerations from the ankle to the head by up to 20% more than normal running. However, McMahon et al (1987) only examined the relative transmission of impact accelerations through the body and did not detail the effect this running technique had on impact accelerations measured at specific anatomical sites (e.g. tibia, sacrum). This is a clear limitation as without this information it is not clear if and where any absolute decrease in the magnitude of impact accelerations is experienced, which is extremely important considering the nature of injury development. For example, it is possible to see the same increase in the amount of impact acceleration attenuation between the ankle and head, if the ankle experienced a 20% increase in impact acceleration and the value recorded at the head remained the same. Furthermore, McMahon et al (1989) only used two accelerometers; one on the ankle and one on the head. It is therefore unclear what happens at anatomical sites between these two points.

According to McMahon et al (1987) running using the Groucho technique was found to increase the rate of energy expenditure by up to 50%. Thus, it may have an additional advantage for health and weight loss. However McMahon only collected data on 6 participants, of which, only 4 were able to complete the experimental trial. This therefore warrants further investigation of Groucho running's energy expenditure characteristics, given its potential health benefits. In addition, McMahon et al (1987) did not consider the potential effect of increased rate of fatigue (as a result of increased energy consumption) on the magnitude and transmission of impact accelerations. Given that running results in fatigue, which has been shown to increase impact acceleration magnitude in normal running (Verbitsky et al, 1998), the effect that fatigue may have on the ability of Groucho/compliant running to attenuate impact accelerations is unclear. This may be a key consideration in relation to Groucho running's suitability as an injury preventive tool.

It is clear from McMahons work that humans have the ability to run with a more bent knee, compliant action. However, limitations to McMahon's study provide uncertainty over the protective capacity of this running style. This study will use three accelerometers (tibia, sacrum, and head) to detail the effect of a more compliant running style on the magnitude of impact accelerations at these sites. Furthermore, a full kinetic and kinematic analysis will provide greater information on joint loading and technique.

Aims of study 1

- To investigate the affect of directed compliant running on:
    - Impact accelerations, Joint kinematics, and joint kinetics in both fatigued and unfatigued conditions, in comparison to normal running.
    - Energy expenditure and other physiological responses in comparison to normal running.

3.2 Methodology

3.2.1 Study Design

This study implemented a randomized experimental repeated measures design to study the effect of running technique (directed compliant running versus normal running) on kinetic, kinematic, physiological variables. All participants completed the same experimental procedure and running trials in random order (figure 3.2.1). A Myomonitor wireless system (DELSYS, USA) was used to measure impact accelerations at the start and end of a fatiguing protocol. Fatigue was determined once an RPE of 17 was reached. A Vicon motion analysis system (Vicon Oxford Metrics, UK) and AMTI force-plate (AMTI, USA) were used to determine kinetic and kinematic differences between the two running techniques in both fatigued and unfatigued conditions. Energy expenditure of the two running styles over a six-minute bout was determined using a Vmax gas flow sensor and analyser (Vmax system, Sensor Medics, VIASYS Healthcare, Netherlands). All equipment was calibrated according to standard protocol.

3.2.2 Participants

Twelve healthy, male participants between the ages of 18-31 were recruited from a university population (height, 177cm ±6.5cm; mass, 78kg ±6.5kg). All participants had been involved in running activities for 6 months or more and took part in running activities at least three times a week at the time of recruitment. Convenience based sampling was employed to recruit participants therefore limiting generalisability to wider populations. Participants were excluded from the study if they had a history of lower limb injury in the previous six months. Informed consent was obtained from all participants in accordance with university guidelines, and Dublin City University's ethical committee granted ethical approval.

Table 3.2 1: Participant height and mass

Participant	Mass (kg)	Height (cm)
1	77.0	182.0
2	76.7	175.0
3	81.9	184.5
4	85.6	176.0
5	78.5	182.0
6	63.4	174.5
7	80.5	179.0
8	88.0	169.0
9	73.0	164.0
10	73.5	180.5
11	79.0	176.5
12	82.9	187.0
Mean	78.3	177.4
Standard Deviation	(±6.5)	(±6.5)

3.2.3 Recruitment and familiarization

Involvement in the study required participants to visit the biomechanics laboratory a total of 5 times which involved three 30-minute familiarization sessions and two 2 hour visits to complete the experimental protocol (figure 3.2.1). All participants read and signed the plain language statement, general health questionnaire, and PARQ prior to being tested in line with ethic committee's policy statement. Participants were informed that they were allowed to dropout of the study at any time.

Three weeks before the experimental trials began participants were shown on three occasions how to run using the compliant style. Each instruction session was separated by a week and the participant was required to practice compliant running a minimum of two times between each. An attempt to monitor these unsupervised practice sessions was made by asking each participant to fill out a diary detailing the duration of each practice session and how they felt (represented by an RPE value). If participants could not adequately perform the required compliant technique within three practice sessions, further practice was performed until competency was reached. Competency was subjectively judged by this researcher, and was determined if appropriate technique was maintained for 8 minutes treadmill running at 8km/hr. Participants were instructed to "drop their hips slightly, to keep their feet close to ground (reducing aerial phase of gait), and to run with flexed knees" as naturally as they could. At the end of each familiarization session a self-selected pace, deemed as each participants normal running pace, was determined. Each participant was asked to start running on the treadmill and speed was increased every minute until a pace was reached that participants considered their normal running pace; for both compliant and normal styles. Participants self selected pace were determined during the familiarization phase by taking an average of the paces they selected at each supervised familiarization session.

Participants were asked to refrain from vigorous exercise for 24hrs prior to each testing day, to wear the same running shoes for each test, and to follow the same pre-test nutrition routine. Also, both tests were carried out at the same time on different days. These restrictions were introduced to reduce test-to-test variability. Participants completed all experimental procedures using both compliant running and normal running techniques in random order.

3.2.4. Experimental procedure

The 12 participants involved in the study took part in the following experimental procedure. Compliant and normal running trials were completed in random order.

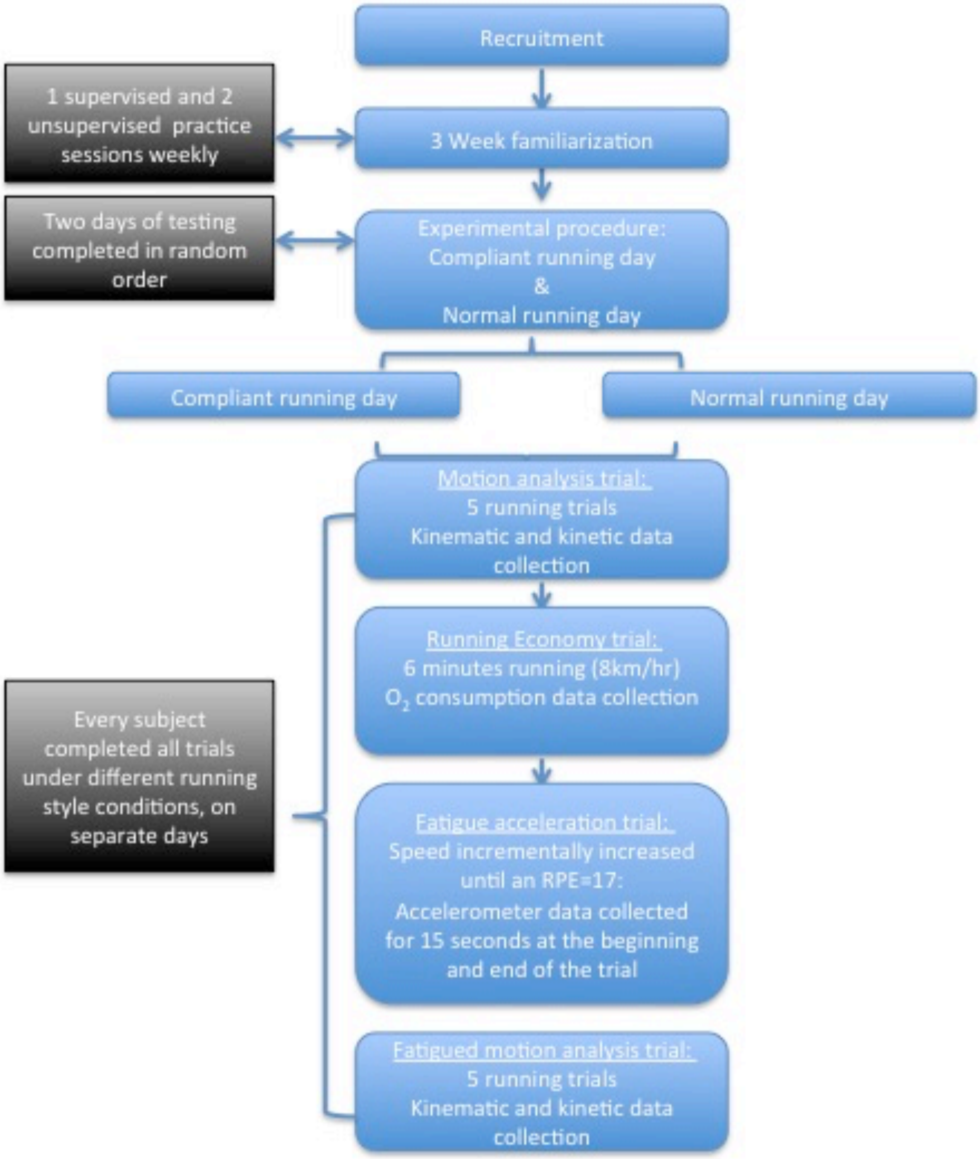

Figure 3.2. 1: Experimental procedure

3.2.5 Motion Analysis

Motion analysis was used to determine differences in joint moments, ground reaction forces, and joint kinematics between running conditions (compliant versus non-compliant; fatigued versus non-fatigued). This involved five runs at a self-selected pace, which was determined in the familiarization phase of testing. Running speed was measured using speed gates set over a five-meter distance.

Ground reaction force data were collected with an AMTI force-plate (USA) that was longitudinally orientated and inserted level with the ground (Kinsella and Moran, 2007). Kinetic and Kinematic information was captured using 12 ME high speed cameras (Vicon Oxford Metrics, UK) by tracking the position of retrofelective markers attached to 21 specific anatomical sites on each participant's body. The 21 markers were placed on each participant in accordance with the Vicon lower body and torso Plug-in-Gait model (Vicon Oxford Metrics, UK). These anatomical landmarks can be seen in table 3.1 and pictures 3.1, 3.2, and 3.3. Light from these markers reflects back into the cameras striking a light sensitive plate, creating a video signal, which along with force plate data (AMTI, USA) was collected and recorded by Vicon Datastation, as described by the Vicon user manual (Vicon Oxford Metrics, UK). Vicon Workstation was then used to take the two-dimensional data from each camera, and combine it with calibration data to convert the equivalent digital motion information into three dimensions (Vicon user manual, Vicon Oxford Metrics, UK). Joint moments, angles and ground reaction forces were calculated using Polygon software (Polygon, Vicon Oxford Metrics, UK) and anthropometric data. Anthropometric measurements included body mass (78kg ±6.5kg), height (177cm ±6.5cm), leg length (95.3cm, ±4.7), knee width (109mm, ±4.4), and ankle width (72.9mm, ±3.6).

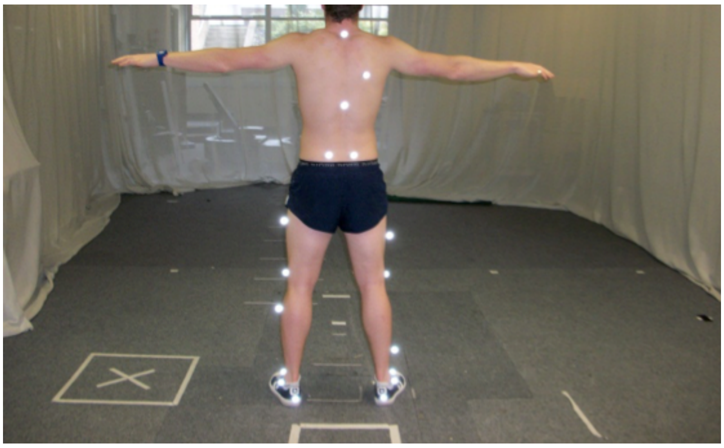

Figure 3.2. 2: Posterior view of anatomical marker positions

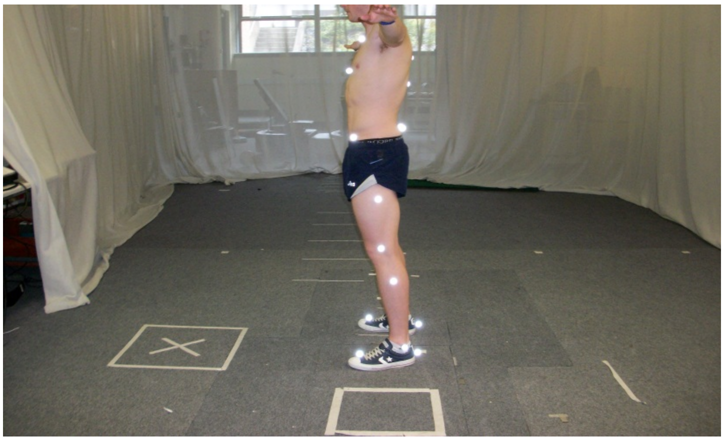

Figure 3.2. 3: Lateral view of anatomical marker positions

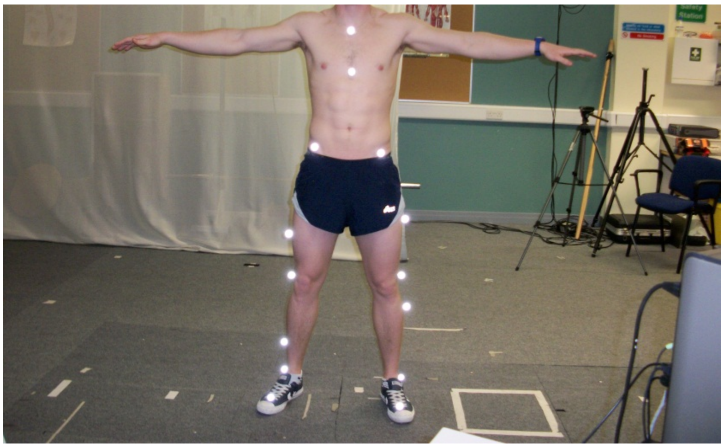

Figure 3.2. 4: Anterior view of anatomical marker position

Table 3.2 2: Anatomical description of marker positions

Anatomical Landmark	Description
7^{th} Cervical Vertebrae	Spinous process of the 7^{th} cervical vertebrae
10^{th}Thoracic Vertebrae	Spinous process of the 10^{th} thoracic vertebrae
Clavicle	Jugular notch where the clavicles meet the sternum
Sternum	Xiphoid process of the sternum
Right Back	Middle of the right scapula: no symmetrical marker on the left side to help determine right and left sides of the body.
Left & Right Anterior Superior Iliac Spine	Placed directly over the right and left anterior superior iliac spine
Left & Right Posterior Superior Iliac Spine	Placed directly over the right and left posterior superior iliac spine
Left & Right Knee	Placed on the later epicondyle of both knees
Left & Right Thigh	Placed the marker on the lateral surface of the thigh at different positions (Right = low, Left = high) on both sides to help determine right and left sides of the body
Left & Right Ankle	Placed on the lateral malleolus along an imaginary line down the centre
Left & Right tibial wand maker	Placed the marker on the lateral surface of the tibia at different positions (Right = low, Left = high) on both sides to help determine right and left sides of the body

Left & Right Toe	Placed on the second metatarsal head, on the mid-foot side of the equinis break between fore-foot and mid-foot
Left & Right Heel	Placed on the calcaneous at the same height above the plantar surface as the toe marker

3.2.6 Running Economy

To determine the energy cost of compliant running in comparison to normal running a six-minute treadmill running economy protocol was used (McMahon et al., 1987). This involved each participant running sub-maximally, in metabolic steady state conditions, where oxygen consumption ($\dot{V}O_2$) is truly representative of energy expenditure per unit time (Hausswirth and Lehenaff, 2001). For the duration of this test participants ran at a pace of 8km/hr. This speed was selected as it has previously been demonstrated to produce steady state conditions for both compliant and normal running (Skime and Boome, 2011). Even at this slow speed participants were deemed to be running based on the observation of a flight phase (subjectively determined). Further support for this comes form McMahon et al (1989) who employed a similar compliant style, and determined that compliant running at slow speeds, where no flight phase was present, was still representative of “running” based on the energetic criterion that the COM was still lowest at mid-stance, as in normal running. Although not measured in this study, this is confirmed in study 2,3,and 4.

According to Mayhew (1977) the oxygen cost of running and running speed have a linear relationship with a strong correlation (R=0.917) once steady state is maintained, thus differences in oxygen consumption and energy expenditure found at 8km/hr should be representative of differences at faster and slower paces. This linear relationship was later supported by Di Prampero (1986). For the duration of the six minute run the treadmill was set to a 1% gradient level (to account for wind resistance), which is suggested to most accurately reflect oxygen cost of outdoor running (Jones and Doust, 1996).

Heart rate and rate of perceived exertion were recorded every two minutes during the test. Heart rate was measured using a Polar heart rate monitor (Polar Electro, USA) and rate of perceived exertion was measured using a 15 point Borg scale where six represents no exertion and 20 represents maximum exertion. Both were used as measures of exercise intensity in order to further highlight differences in the physiological response to both running styles.

Breath-by-breath oxygen consumption was measured for the duration of the six-minute bout using the Vmax system (Vmax system, Sensor Medics, VIASYS Healthcare, Netherlands). The gas flow sensor and analyser were calibrated before testing each participant using a 3 litre syringe for flow volume across a wide range of flow rates and calibration gases, as according to standard protocol, and the sensor was attached to each participant using a head and mouth piece as seen in picture 3.4. Volume of oxygen consumed over the six-minute bout was averaged over 20 second periods and recorded. Values for the last two minutes of each test were averaged to determine O_2 consumption and energy expenditure. Energy expenditure is reported relative to mass. The standard Weir equation (1949) was used to calculate caloric expenditure from steady sate data (as seen below).

$$Calorie\ expenditure\ (kcal \cdot min^{-1}) = 3.94 * \dot{V}O_2\,(\mathrm{l} \cdot min^{-1}) - 1.1 * \dot{V}CO_2\,(\mathrm{l} \cdot min^{-1})(\text{Weir, 1949})$$

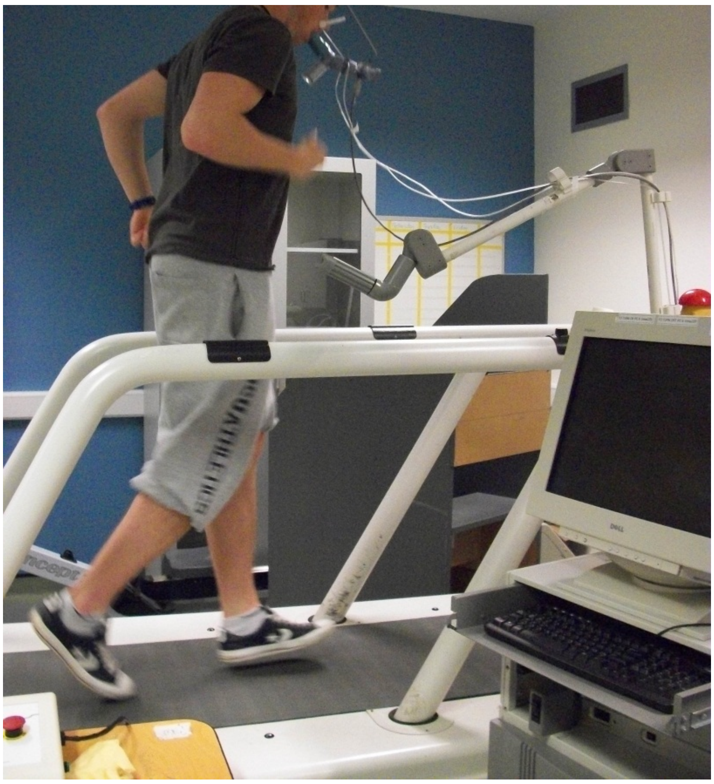

Figure 3.2. 5:Gas flow sensor and monitor

3.2.7 Fatigue trial and Impact acceleration measurements

An incremental running protocol was used in order to determine the effect of neuromuscular fatigue on impact loading in running. Participants began running at a self-selected pace (compliant= 10km/hr± 1.4, normal= 12km/hr ± 1.4) and the speed was increased by 1km/hr every 4 minutes. An RPE of 17, determined as “very hard” according to the Borg scale, was chosen to determine the end of the test (Moran and Marshall, 2006). Following a similar running protocol, an RPE of 17 has previously been shown to be associated with increased tibial accelerations in drop jumps (Moran and Marshall, 2006) due to neuromuscular fatigue. Furthermore, RPE has been used widely as a tool for prescribing physiological intensities (Lamb, Eston, and Corns, 1999; Moran and Marshall, 2005) during exercise and has been shown to be a valid measure for this purpose (Steed, Gaesser, and Weltman, 1994).

During the test, impact accelerations were measured under fatigued and unfatigued conditions using a lightweight, portable Myomonitor system (DELSYS, USA). Two fifteen-second windows (1 minute into the test, and prior to termination at an RPE of 17 when fatigue is reached) were selected to measure impact accelerations. When measuring impact accelerations when fatigued, the pace was reduced to the original self-selected pace at which the test began. This was done to eliminate the effect of running speed on impact acceleration magnitude.

Accelerometers were attached to the skin at the tibia (overlying the proximal, anterior-medial aspect and aligned along the longitudinal axis of the tibia) (Moran and Marshall, 2005), the sacrum (mid-way between the posterior superior iliac spines, aligned along the longitudinal direction of the spine) (Voloshin et al, 1998), and on the forehead (anterior aspect of the skull) (Mercer et al, 2003). Accelerometers were fixed in position using double sided tape and prewrap (Durapore, 3M, Bracknell, UK), and secured with an elastic strap wrapped around the tibia, waist, and head pressing the accelerometers onto the skin as tightly as comfortably allowed (Voloshin et al, 1998). Although skin mounted accelerometers have been shown to underestimate the magnitude of impact accelerations (Lafortune, Hennig, and Valiant, 1995), the effect is considered to be consistent across conditions (fatigued versus unfatigued) and running style (compliant versus normal), and does require an invasive procedure. Previous running studies have used skin-mounted accelerometers to investigate impact accelerations (Derrick, Dereu, and Mclean, 2002; Mizrahi et al, 2000; and Voloshin et al, 1998).

3.2.8 Dependent Variables

- Peak impact accelerations (tibia, sacrum, and head)
- Peak vertical ground reaction force
- Joint Angles (at foot strike and toe off)
- Net peak joint moments
- Oxygen consumption
- Heart rate
- Energy expenditure
- RPE

3.2.9 Statistical Analysis

To examine the effect of running style and fatigue on impact acceleration values measured at the tibia, sacrum, and head as well as other kinetic and kinematic variables (section 3.2.8) multiple 2(compliant Vs. Normal running)*2(fatigued Vs. unfatigued) repeated measure ANOVA's were completed. Furthermore, in order to determine the effect of the compliant running style on O2 consumption, energy expenditure, RPE, and average heart rate multiple paired sample t-tests were completed. An alpha value of $p< 0.05$ was used to indicate statistical significance. All results are presented in the next chapter as means with standard deviations. Normality of data was determined using the Shapiro-Wilk test for normality. Mauchleys test was used to examine sphericity. In cases where the assumption of sphericity was violated a Greenhouse-Geisser correction was employed.

3.3 Results

3.3.1 Peak accelerations

Statistical analysis revealed that all data was normally distributed and that there was no significant interaction effect between running style and fatigue at the tibia (F=(1,9)= 1.82, p=0.210, partial eta squared= 0.168), sacrum (F (1,9)= 1.752, p=0.218, partial eta squared= 0.163), or head (F (1,9)=2.251, p=0.172, partial eta squared=0.220).

There was no significant difference in peak accelerations at the tibia for either running style (F (1,9)=0.794, p=0.396, partial eta squared=0.081) or fatigue (F (1,9)=0. 717, p=0.419, partial eta squared= 0.074).

However, significant differences were observed in peak sacral accelerations for both running style (F (1,9)=8.77p<. 05, partial eta squared=0.493, normal > compliant by 41%) and fatigue (F (1,9)=13.94, P<. 05, partial eta squared=0.603, fatigued > unfatigued by 28%). A similar pattern was observed at the head with significant differences displayed for both running style (F (1,9)=7.28, p< .05, partial eta squared= 0.477 normal > compliant by 28%) and fatigue (F (1,10)=6.930, p<. 05, partial eta squared=0.464 unfatigued > fatigued by 12%).

Retrospective analysis with regard to examining a fatigue*running style interaction reported observed power of 0.259 for tibial accelerations, 0.171 for sacral accelerations, and 0.298 for head accelerations. With regard to the effect of running style power analysis indicates that observed power was 0.166 for tibial accelerations, 0.66 for sacral accelerations, and 0.780 for head accelerations. Finally, for determining an effect of fatigue retrospective analysis indicates an observed power for tibial accelerations of 0.174, 0.291 for sacral accelerations, and 0.426 for head accelerations. This therefore indicates that the sample was underpowered with regard to each measure.

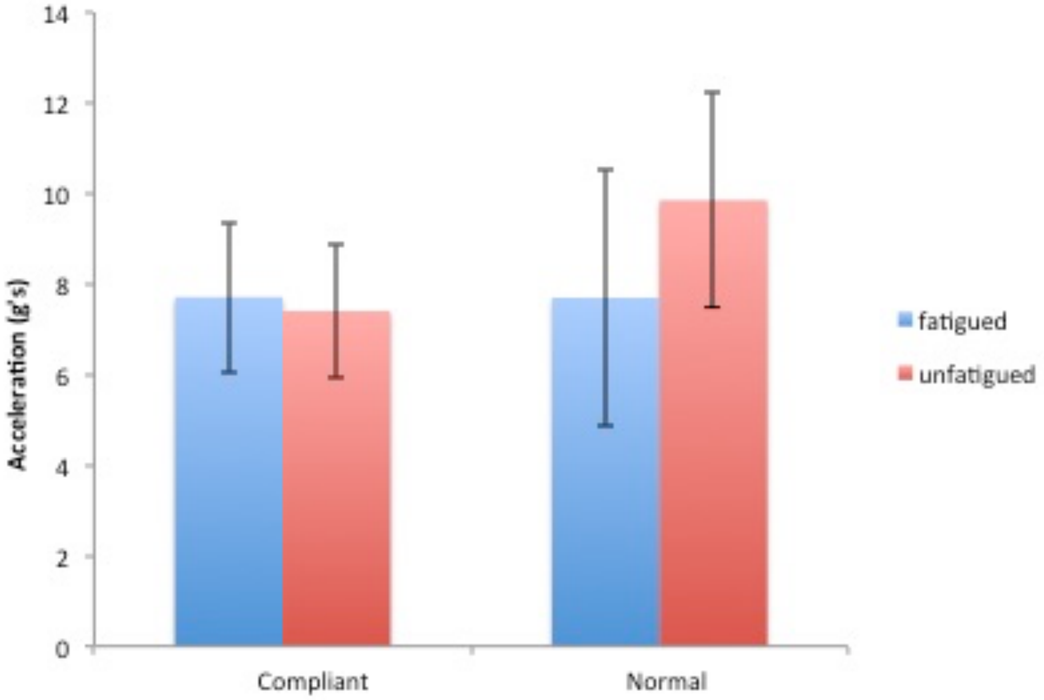

Figure 3.3. 1: The effect of running style and fatigue condition on peak tibial accelerations (Mean + Standard deviation).

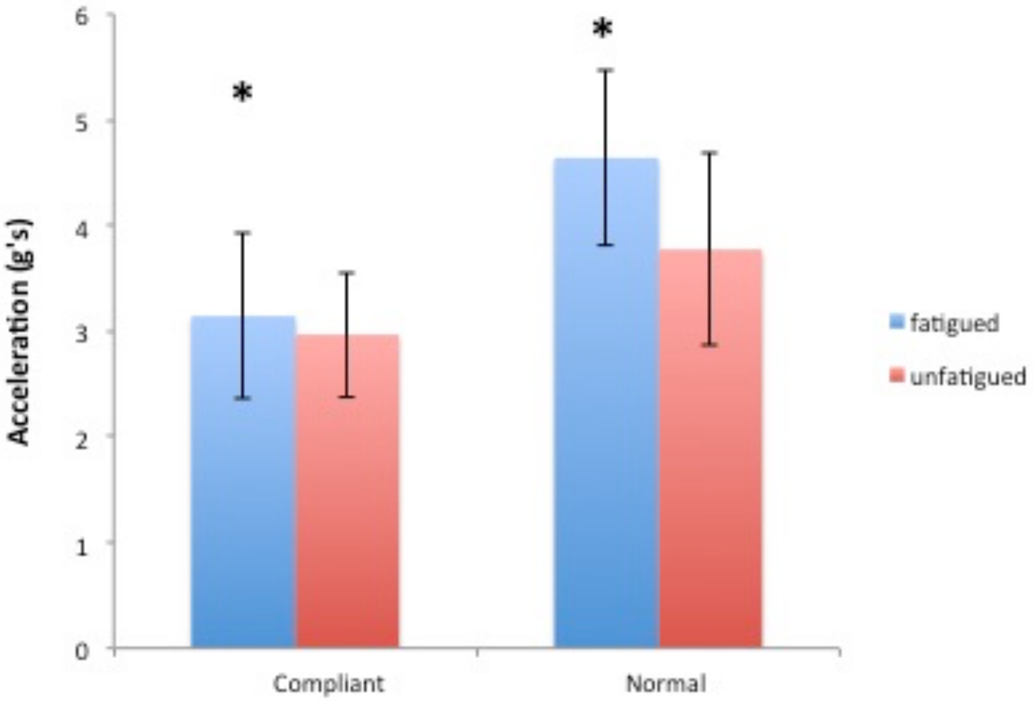

Figure 3.3. 2: The effect of running style and fatigue condition on peak sacral accelerations (meand+SD) (* indicates significantly greater accelerations for Normal running and fatigued running, P<0.05).

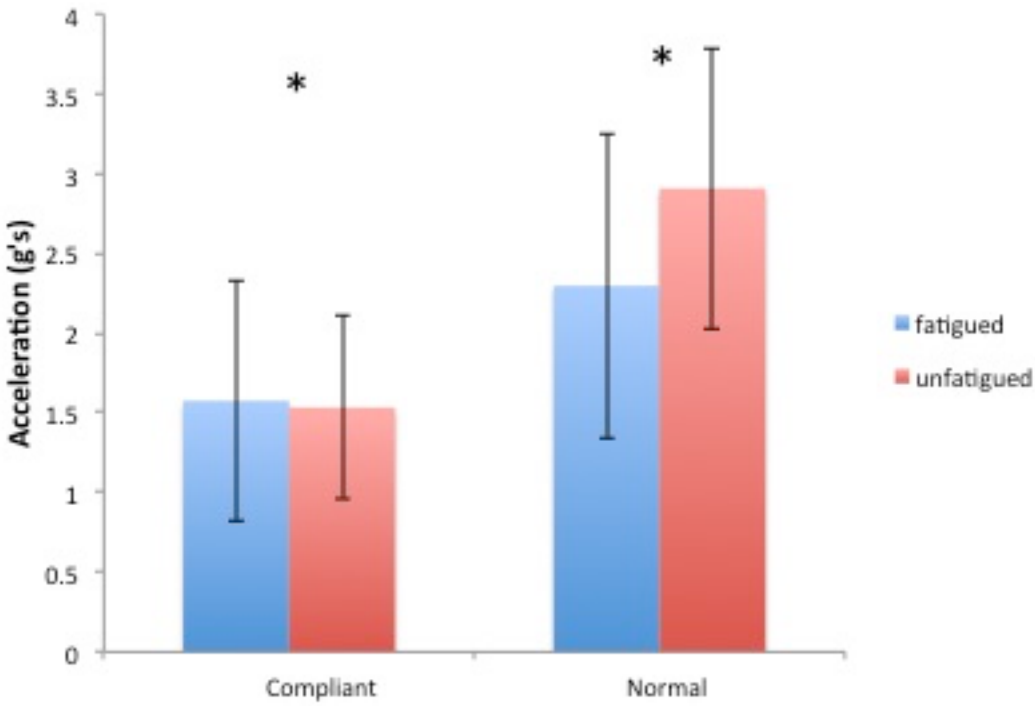

Figure 3.3. 3: The effect of running style and fatigue condition on peak head accelerations (Mean + SD) (* indicates a significantly greater values for normal running and unfatigued running., p<0.05).

3.3.2 Energy expenditure results

Statistical analysis revealed a significant effect of running style on calories expended per unit time (Kcal.kg^{-1}.min^{-1}) (T (11)= 2.64, $p<0.05$ compliant > normal by 21%), oxygen consumption per unit time ($\dot{V}O_2$ /ml/kg/min) (T (11)=2.55, $p<0.05$, compliant > normal by 24%), and rate of perceived exertion (T(11)=2.97, $p<0.05$, compliant > normal 20%). However, no effect of running style was found for average heart rate (T (11)=2.14, $p>.05$).

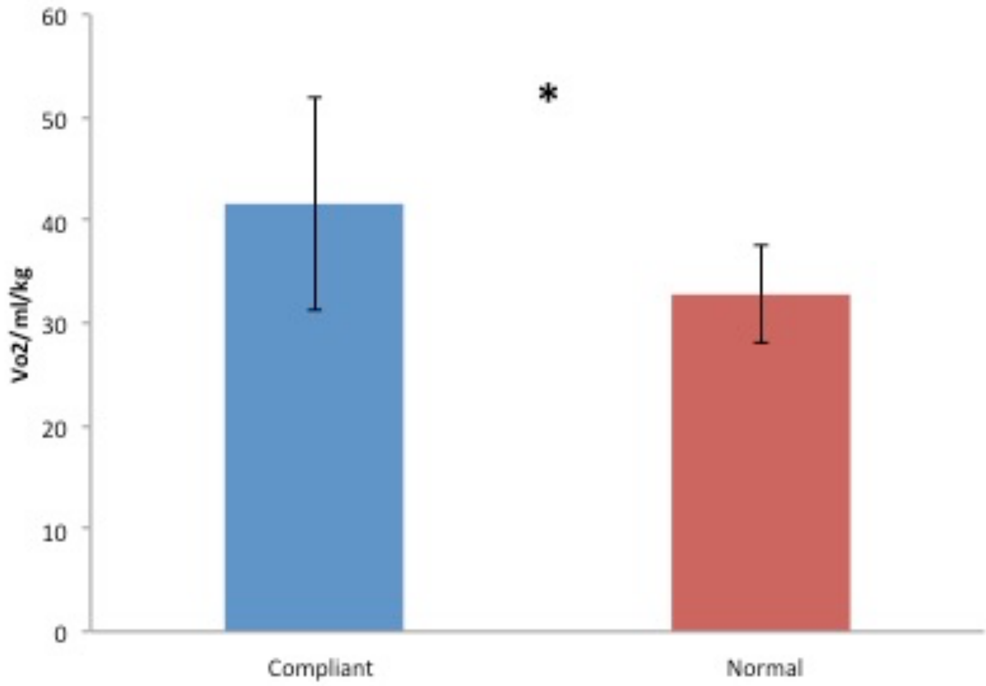

Figure 3.3. 4: The effect of running style on Oxygen consumption (* indicates a significant difference between running styles, p<0.05) (Mean + Standard deviation)

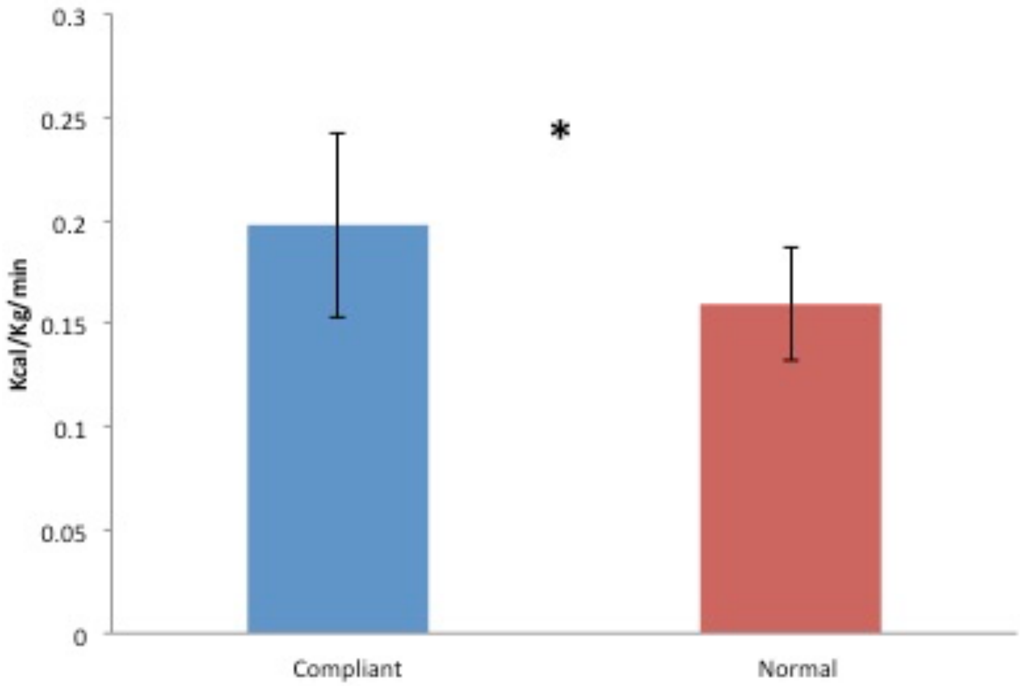

Figure 3.3. 5: The effect of running style on calories expended per unit time (Mean + Standard deviation) (* indicates a significant difference between running styles, p<005)

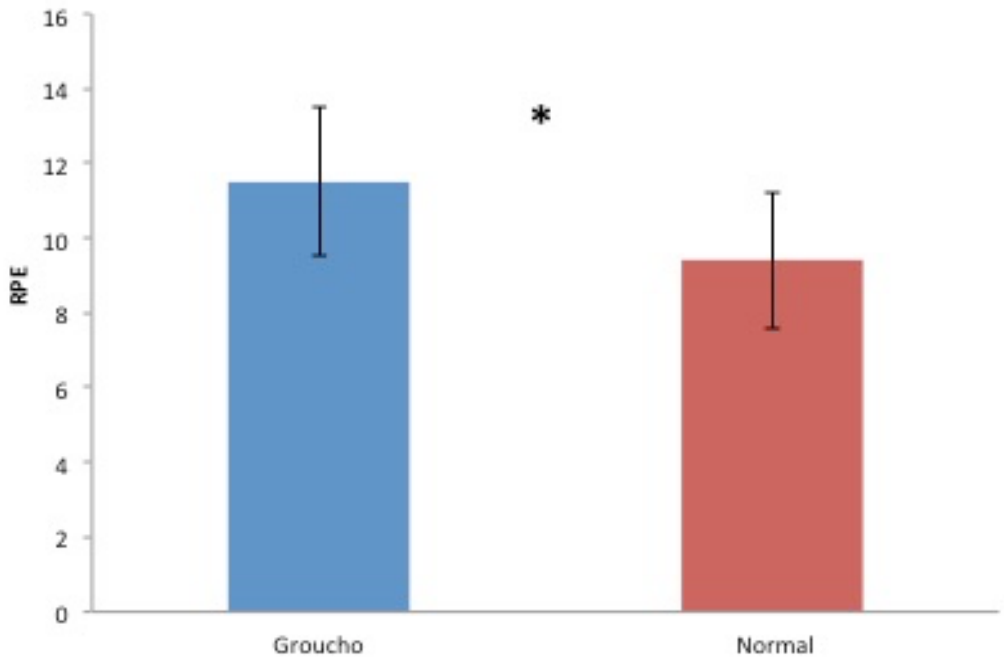

Figure 3.3. 6: The effect of running style on rate of perceived exertion (Mean + Standard deviation) (* indicates a significant difference between running styles, p<0.05)

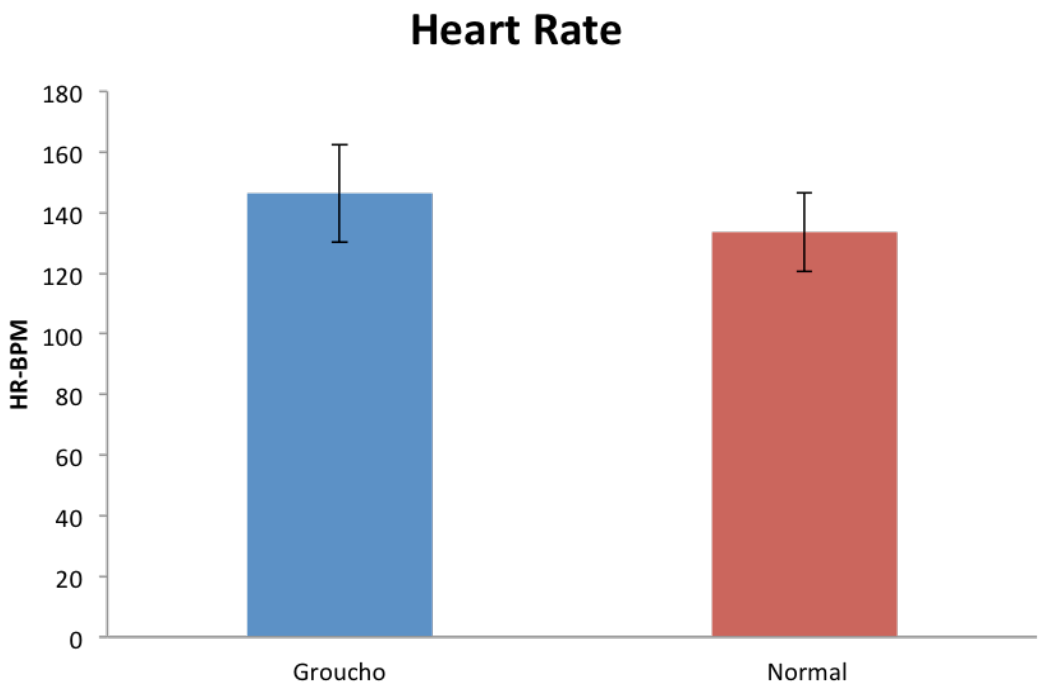

Figure 3.3. 7: The effect of running style on average heart rate(Mean + Standard deviation) (* indicates a significant difference between running styles,p<0.05)

3.3.3 Kinetics and Kinematics

Statistical analysis of kinematic variables indicates that at foot strike compliant running displayed significantly larger degree of hip flexion by 19% (F (1,10)=2.224,p<0.05), and knee flexion by 52% (F (1,10)=38.5,p<0.05). A similar pattern was observed at toe-off with compliant running displaying larger hip flexion by 26% (F (1,10)=7.0,p<0.05) and knee flexion by 19% (F (1,10)=28.6,p<0.05).

Examination of kinetic variables indicated that compliant running displayed significantly smaller vertical ground reaction forces by 17 % (F (1,10)=9.12, p<0.05), smaller peak plantar flexor ankle moments by 43% (F (1,10)=55.2, P<0.05), and smaller peak knee flexor moments by 49% (F (1,10)=6.64, p<0.05). However, compliant running displayed larger peak hip flexion moments by 52% (F (1,10)=38.5, P<0.05). There was no significant interaction effect for any of the kinetic or kinematic results.

Table 3.3. 1: The effect of running style on kinetics and kinematics

Variables	Compliant	Normal	Signif difference	% Diff.
Hip flexion: foot strike (degrees)	58 (±12.06)	48 (±7.7)	(F= 22.4,p<0.05)	19% (C>N)
Hip flexion: toe off (degrees)	43 (±14.64)	33 (±6.96)	(F=7.0, p<0.05)	26% (C>N)
Peak hip moment (flexor) (Nm/Kg)	1.58 (±0.914)	.90 (±0.47)	(F=5.4, p<0.05)	54% (C>N)
Knee flexion: foot strike (degrees)	29 (±9.04)	17 (±7.05)	(F=38.5, p<0.05)	52% (C>N)
Knee flexion: toe-off (degrees)	58 (±7.30)	48 (±5.67)	(F=28.6, p<0.05)	19% (C>N)
Peak knee moment (extensor) (Nm/Kg)	12.92 (±5.12)	21.22 (±8.93)	(F=6.64, p<0.05)	49% (C<N)
Peak ankle moment (plantar flexor) (Nm/Kg)	25.3 (±4.18)	39.06 ±(7.63)	(F=55.2, P<0.05)	43% (C<N)
Ground reaction force (Passive impact peak) (N/Kg)	6.78 (±1.41)	8.01 (±1.39)	(F=9.12, P<0.05)	17% (C<N)

3.4 Discussion

The primary aim of this study was to investigate the effect of directed compliant running on impact accelerations, joint angles, and joint moments under both fatigued and unfatigued conditions and to determine the energy cost of this running style in comparison to normal running. Impact loads have been implicated as a central factor in the development of running injuries (Davis 2010, 2011), furthermore it is clear that joint orientation can influence impact loading (Derrick, 2004, Gerritson 1995), thus these variables provide useful information with regard to the likelihood of injury development when adhering to a more compliant style. Given that a change in kinetics or kinematics can also influence energy expenditure via numerous mechanisms (section 2.3.1), this will also be discussed.

3.4.1 Impact accelerations

A significant decrease in peak impact acceleration values at the sacrum in compliant running compared to normal running (normal > compliant by 41%) suggests that compliant running may decrease the risk of lower back pain and Osteoarthritis as increased impact accelerations have been implicated as factors associated with the development of both (Collins and Whittle, 1989; Folman et al, 1986; Sandover, 1983; Hawes et al, 1979; &Wosk and Voloshin, 1985). Evidence also suggests that the significant decrease in peak impact acceleration values measured at the head in compliant running (normal > compliant by 28%) may decrease the incidence of headaches and further protect the skull, and neck from bone or soft tissue damage, degenerative joint change, or stress fracture development (Whittle, 1999; Lafortune et al, 1996; McMahon et al, 1987). Indication for a potential increased protective capacity for compliant running, resulting from an improved ability to reduce the magnitude of location specific peak impact accelerations (in comparison to normal running) is clearly evident. Lafortune et al (1996) describes increasing knee flexion in running as the most effective and suitable method of protecting the head and back from overuse injury. Results of this study support this statement; an increase in knee flexion at foot-strike of 52% was associated with significant decreases in impact acceleration values at both the sacrum and head (41% and 28% respectively). Similar results were presented by Lafortune et al (1996), where a decrease in impact acceleration values, measured at the head, of 45% was associated with increasing knee flexion angle from 0° to 40°. The smaller decrease in peak head acceleration values in the current study may be due to the smaller degree of knee flexion present in the directed compliant running style (29° compared to 40°) (Lafortune et al, 1996).

Principles of biomechanics indicate that reduction of any force or load results from manipulation of the impulse momentum relationship ($F = (m \cdot \Delta v) / \Delta t$). It is therefore clear that a decrease in the magnitude of force or load experienced by the body results from either a decrease in effective mass, a change of velocity, or an increase in the time interval over which this interaction occurs.

Given that increased knee flexion has been shown to increase contact time by up to 125% (Devita and Skelly, 1992), reduce vertical landing velocity to almost 0m/s(via reduced COM oscillation) (McMahon et al, 1987) and reduce effective mass (Derrick, 2004) it is likely that the observed reductions (in this study) are due to a combination of a number of factors. Increased knee flexion at foot-strike, results in the body being split into different segments that are accelerated as separate bodies, thus causing a decrease in effective mass for each segment (Derrick, 2004). Given that ground reaction forces are proportional to the mass that they act on, this reduction in effective masses may act partly to explain the found decrease in load experienced at the sacrum and the head.

Another mechanism that may explain the decreased impact acceleration values at these sites may be the effect that increased knee flexion has on effective vertical stiffness, and consequently vertical landing velocity and contact time. Although this was not measured in this study, Lafortune et al (1996), found that increasing knee flexion from 0° to 40° was associated with a decrease in effective vertical stiffness from 28.7 KN.m-1 to 7.9KN.m-1. Similarly, McMahon et al (1989) found that effective vertical stiffness decreased to 82% of that experienced in normal running with a mid-stance thigh angle of 50% (compared to 70% in normal running). Therefore, both studies indicate a clear association between increased knee flexion at foot contact and decreased effective vertical stiffness. McMahon (1989) also displayed an associated increase in contact time and a reduction in vertical landing velocity of almost 0m/s (McMahon et al, 1989) with an increase in knee flexion. Furthermore, Blickhan's planar spring model (1989), predicts a negative relationship between vertical spring stiffness and contact time (thus reduced stiffness will act to increase contact time) and running on compliant surfaces (thus decreasing stiffness) has been show to increase contact time 3 fold (McMahon & Greene, 1979).

To summarize, increased knee flexion may decrease effective vertical stiffness, causing an increase in the time period over which the impact force is experienced and reducing vertical landing velocity (via decreased COM oscillation). Therefore, altering the impulse momentum relationship via these mechanisms and decreasing the magnitude of force offers a potential explanation for the decreased impact loads experienced at the head and sacrum for directed complaint running.

Peak tibial impact accelerations were shown to be 10% greater in normal running than in directed compliant running. Although statistical analysis revealed this difference to be insignificant, this still may have important implications in relation to injury development. In a case presented by Milner et al (2006), an insignificantly ($p = 0.051$) larger ground reaction force of 4% was evident in participants who had previously experienced tibial stress fractures. Milner (2006) suggested that due to the repetitive nature of running, this minor, statistically insignificant increase in load might prevail as a central factor in the development of an overuse injury (a stress fracture in this case), as a result of a larger cumulative effect over long periods of time. An average long distance runner will cover 130km per week resulting in 40,000 impacts within 7 days (Holmes, and Andrew (2003). Although a recreational runner will not cover this amount of distance, it has been found that between 15-40 km a week is not uncommon (Mundermann, Nigg, Humble, and Stefanyshyn, 2003), which according to Flynn et al (2003) will account for between 5000 -11,000 impacts weekly, amounting to over 40,000 in one month. Similar numbers would be expected for directed compliant running, as McMahon et al (1989) found no difference in cadence (not measured in this study) when compared to normal running. It is for this reason that a statistically insignificant decrease in tibial accelerations of 10% in comparison to normal running may still be important; especially considering that peak tibial acceleration has been implicated in the development of running related injuries and tibial stress fractures prospectively (Davis et al., 2004, Davis et al., 2010) and retrospectively (Milner et al, 2006b, Pohle et al., 2008, Zifchock et al., 2007).

The insignficant effect of directed compliant running on tibial acceleration values does not agree with other literature that increased knee flexion and subsequently reduced vertical stiffness. In fact, Lafortune et al (1996) and McMahon et al (1989) both found an increase in peak tibial impact acceleration values when knee flexion at foot-strike was increased, of 52% and 18-38% respectively (not presented in McMahon et al, 1989, but calculated based on figures displaying impact attenuation profiles). This may be due to a lesser degree of knee flexion (20% vs. 40% in Lafortune et al (1996)) in the current study, resulting in a smaller change to effective mass (Derrick, 2004). However, the 10% reduction in tibial acceleration may still be an under estimation of the reduction in loading on the tibia.

Impact acceleration values are used to infer force as acceleration is proportional to force, when mass remains constant (F=MA). However, according to Derrick (2003) a change in running technique involving increased knee flexion is associated with a decrease in the effective mass of the tibia and a subsequent increase in impact acceleration values, despite a decrease in vertical ground reaction force. In fact, it has been reported that as initial knee angle during running increases from 5 to 20 degrees (for a 65kg participant), effective mass decreases from 11kg to 5kg (Denoth, 1986 cited by Milner, 2007). Therefore, if this effective mass theory applies to compliant running, it is likely that a reduction in tibial accelerations will reflect larger reductions in vertical ground reaction force (as evident in this study (vGRF decreased by 17%)) and thus a potential deceased risk of injury development.

Finally, The decreased (although insignificant) tibial impact acceleration values for directed compliant running, may be explained by the same mechanisms that manipulate the impact-momentum relationship (increased contact time and decreased vertical landing velocity), as described when discussing the found decreases in impact acceleration values at both the sacrum and head.

Significant differences were found between fatigued and unfatigued conditions at the sacrum and head, in relation to peak impact acceleration values. At the sacrum fatigued accelerations were found to be greater than unfatigued accelerations by 28%, which is in agreement with numerous studies that report an increase in the magnitude of impact accelerations under fatigued conditions (Voloshin et al, 1998; Verbitsky et al 1998; & Mercer et al (2003) and an associated risk in the increase of overuse injuries. This can be explained by a reduction in the functional capacity of muscle associated with neuromuscular fatigue, which results in a diminished ability of muscles to attenuate impact accelerations (Radin, 1986).

However, accelerations measured at the head were significantly lower by 6% under fatigued conditions, which does not agree with the above-mentioned literature. This may be due to some type of protective mechanism present within the body that protects the brain once a fatigue threshold, or acceleration threshold is reached. However no evidence is present in the literature to support this. Another possible explanation is that somewhere between the sacrum and the head, there was an effect of increased localized muscle fatigue, which according to Flynn et al (2004) causes an decrease in the force-generating capacity of the muscle (as a result of peripheral fatigue mechanisms) and subsequently a decrease in muscle tension and stiffness. This decrease in stiffness increases the muscles ability to attenuate impact accelerations, and may therefore be responsible for the greater impact accelerations found under unfatigued conditions in comparison to fatigued conditions at the head.

It is important to note there was no interaction effect for fatigue and running style, thus indicating that compliant running did not react differently under fatigued conditions than normal running. In addition, these findings may only be generalizable to fatigue induced via running.

3.4.2 Ground Reaction Forces and Joint Moments

It was found that compliant running significantly decreased peak ground reaction forces by 17% ($p<0.05$) in comparison to normal running. Ground reaction force is a measure of the force applied to the body by the ground and has been prospectively (Davis, Bowser & Mullineaux 2010) and retrospectively (Zifchock, Davis & Hamill 2006a, Creaby, Dixon 2008, Pohl et al. 2008, Milner et al. 2006a) implicated in the development of running related injuries. More, specifically participants who develop running related injuries, tibial stress fractures, or plantar fasciitis present vertical ground reaction force values 13%, 33% and 40% larger prior to injury than uninjured controls, respectively. Thus, a reduction of 17% associated with compliant running would decrease vertical ground reaction force variables to within “safe” or “normal” levels for those who suffer running related injuries and may decrease the overall risk of injury development. Furthermore, Edwards et al (2010) suggests that small decreases in load may result in significantly larger numbers of impacts required till failure, in relation to fatigue failure of bone structures. Therefore, the found decrease in ground reaction forces in compliant running may increase the number of loads required to cause stress fracture occurrence, and reduce the risk of injury.

As previously discussed in relation to reduction of impact acceleration values, reductions in vertical ground reaction forces are dictated by changes to the impulse-momentum relationship. Therefore, the associated increase in contact time and vertical landing velocity associated with increased knee flexion (compliant> normal by 52%) may partly explain this 17% reduction in vertical ground reaction force. Contributing to this reduction of 17% may also be a reduced effective mass due to increased knee and hip flexion during compliant running (Devita & Skelly, 1992, Derrick 2004). In fact, Gerritson et al (1995) suggests that for every 1° increase in knee flexion there is an associated 68N decrease in vertical ground reaction force.

Compliant running displayed smaller extensor moments at the knee (49%), and smaller plantar flexor moments at the ankle (22%). Increased joint moments result in increased loading across a joint (Winter, 2009), and thus give a representation of the likelihood of injury development. These measures have been implicated either directly or indirectly in the development of numerous overuse injuries and indicate a potential protective capacity of compliant running.

Implementing mathematical modelling, Sasimontonkul et al (2007) suggested that joint moments contribute largely to the magnitude of tibial contact forces, which subsequently were implicated in the development of tibial stress fractures (Sasimontonkul, Bay & Pavol 2007). Similarly, Haris et al (2010) indicated that knee and ankle joint moments contribute to the magnitude of bone bending moments at 11-equidistant points along the tibia, which are also implicated in the risk of stress fracture development. Furthermore, Stefanyshyn (1991) showed that increased knee moments resulted in increased stress within the patella-femoral joint leading to pain and the possible development of patella-femoral syndrome. More specifically, knee extensor moments have been implicated in contributiung to the magnitude of Patella femoral joint stress (Whyte et al. 2010).

There was no significant effect of fatigue on ground reaction forces, rate of force development, or joint moments. This may be explained by an excessive amount of time between fatiguing trials and motion analysis. This increased time was present due to difficulties with removing accelerometers and replacing the 21 retrofelective markers required for motion analysis. High levels of perspiration following the fatigue protocol further increased the difficulty of placing the markers on participants, as the sweat inhibited the ability of the tape to stick to skin.

2.4.3 Energy expenditure

In agreement with McMahon et al (1987), compliant running was found to significantly increase energy consumption (Kcal.kg^{-1} and Kcal.kg^{-1}.min^{-1}) by 21% in comparison to normal running. Although McMahon recorded increases of up to 50%, this magnitude of increase was not present in all of their participants. The increased energy cost of compliant running in comparison to normal running was further represented by an increased level of oxygen consumption of 24% and an increase in the rate of perceived exertion of 20%.

This increase in running expenditure obviously rules out compliant running as a possible running style during performance events. However, increased energy expenditure may yield greater health benefits than normal running, and may therefore be an attractive tool for use by fitness experts or recreational runners. This is due to the fact that for the same time period, running at the same pace, compliant running may expend more energy than normal running for this population. Increasing energy expenditure by 1000 kcal a week has been shown to increase life expectancy by 20 % and an average of 2000kcal expended during physical activity (a week) is associated with a decrease in morbidity and mortality of 20-30% (Warburton et al, 2006; & Lee and Skerret, 2001).

3.5 Conclusion

The aim of this study was to investigate how directing participants to run with a more compliant running style (by increasing knee flexion at foot contact), would effect loading on the body under both fatigued and unfatigued conditions as well the effect this style may have on energy expenditure in comparison to normal running, with an overall goal of determining if this style may have injury preventative characteristics.

Completion of data collection and subsequent analysis indicates that significant reductions in impact loading during compliant running (reflected through significant decreases in sacral impact accelerations values, head impact accelerations values, knee and ankle peak joint moments, and vertical ground reaction forces) may highlight a decreased risk of developing impact related overuse injuries. Furthermore compliant running was shown to have a significantly larger energy cost (Kcal.kg^{-1}.min^{-1}) than normal running.

3.6 limitations

- The time between the fatigue protocol and the fatigued motion analysis was too large, possibly allowing the participants to recover, hindering results.

- Directed compliant technique was subjectively judged during familiarization and experimental procedure.

- $\dot{V}O_{2\max}$ or heart rate were not determine but may have provided a more objective measure of intensity than RPE during the fatigue trial.

- RPE was not anchored and may yield a more objective measure of intensity.

- Convenience based sampling was employed for recruitment of participants and therefore limits generalizability to wider populations.

- Large variance in impact acceleration data may limit generalizability of results to larger populations.

- Retrospective power analysis indicates that the sample is underpowered and therefore may display a type 2 error.

3.7 Future recommendations

- Given the potential high cost required to induce compliant running kinematics via verbal feedback provided by a compliant running expert, other teaching methods, such as the use of biofeedback, should be investigated.

- The use of an anchored RPE or leg specific RPE may provide a better estimation of neuromuscular fatigue.

- In order to determine the true ecological effect of fatigue time taken to reach the fatigued state should be considered.

Chapter 4: Study 2

A comparison of varying accelerometer locations for providing real-time visual biofeedback

4.1 Introduction

As established in the introduction chapter, the development of overuse injuries in running has both detrimental health and economic implications, which clearly justify the need to investigate appropriate means of reducing the risk of developing such injuries. It is clear from study 1 that directing people to run with increased knee flexion, to facilitate a more compliant gait, may decrease the magnitude of impact loads, and thus potentially reduce the risk of overuse injury. This raises the question of how to most successfully bring about the required gait alterations. It is clear that an essential component within the development of a novel running style is the development of an appropriate gait re-training intervention. The idea of gait re-training is not new, and with the use of real-time biofeedback this has been accomplished in numerous clinical conditions: cerebral palsy (Seeger et al., 1981), amputees (Gapsis et al., 1982), total hip replacements (White and Lifeso, 2005) and trans tibial amputees (Dingwell et al., 1996). However, what is new is the idea of using biofeedback from an accelerometer to specifically decrease the magnitude of impact accelerations produced during running. Crowell et al (2010) investigated if providing individuals with real-time visual feedback from an accelerometer attached to the tibia could reduce their tibial acceleration and ground reaction force during a single session of gait retraining. Crowell (2010) showed that 3 out of 6 participants were successfully able to significantly reduce tibial accelerations after a 10-minute bout of continuous feedback. Following the 10-minute bout of continuous feedback, participants were instructed to run for a further 10-minutes, with no feedback present. Participants were asked to maintain the altered running mechanics that were developed as a result of the visual biofeedback. Following this no-feedback period 5 of the 6 participants significantly reduced impact accelerations. Significant reductions ranged from a decrease of 17% to 60% (Crowell et al, 2010). It was concluded that visual biofeedback could successfully reduce magnitude of impact accelerations, which consequently may have important implications for injury prevention. This highlights the potential for providing visual biofeedback from other accelerometer locations both on and off (i.e. treadmill) the body. This has not been previously examined. The present study proposes to use a similar visual biofeedback system as Crowell (2010), to investigate the use of visual biofeedback in three different locations (the tibia, the sacrum, and the treadmill), with the goal of determining which has the most potential for reducing impact loading. Furthermore, it is not clear from Crowell's (2010,2011) studies, what effect this gait retraining method has on loads further up the body (sacrum) or what kinematic changes are made to facilitate the observed reductions in tibial acceleration. Therefore, the present study proposes to examine kinematic changes that participants make to their gait pattern in order to bring about any observed reduction in peak impact accelerations. Given that any kinematic changes made to gait will be self-directed this may provide interesting information regarding compliant strategies that may reduce load, but are currently unknown. This information may then be used for the improvement of future compliant running intervention strategies.

Aims of Study 2

- To compare the use of three different accelerometer sites (tibia, sacrum, and treadmill) for providing visual biofeedback, in relation to their ability to cause gait alterations to reduce impact accelerations.
- To investigate kinematic strategies used by participants to reduce impact acceleration peaks.

4.2 Methodology

4.2.1 Experimental design

This study implemented a three group, randomized, pre-test post-test, experimental design to examine the effect of three different forms of visual, accelerometer based biofeedback (tibial, sacral, and treadmill) on the magnitude of impact accelerations and running kinematics. Participants were recruited from a university population and local sports clubs and were required to attend the Biomechanics lab in the School of Health and Human Performance (Dublin City University (DCU)) on only one occasion. All participants completed a physical activity readiness questionnaire (PARQ) (ACSM, 1997) and informed consent form prior to participation. This study was granted ethical approval by DCU's ethics committee, as required.

4.2.2 Participants

Twenty-seven male participants were recruited (Age: 26.6 ± 7.6yrs, height: 179.8 ± 8.2 cm, Mass: 77.5 ±10.1kg). All participants were involved in running, or running related activities for at least thirty minutes a week, for six months prior to commencement of this study. Participants were excluded if any injuries (defined as any physical impairment affecting running distance, speed, duration or frequency), cardiovascular diseases, or any neurological disorder that may affect their gait were present. Participants were also excluded if they had any lower limb surgery (Devita, Hunter & Skelly, 1992). Volunteers from study 1 were excluded from participating in order to prevent the influence of learning and practice effects.

To increase internal validity, participants were randomly divided into three groups of 9 participants (tibial biofeedback, sacral biofeedback, and treadmill biofeedback) (Campbell & Stanley, 1936 as cited in Thomas, Nelson & Silverman, 2011). Convenience based sampling was employed for participant recruitment, thus limiting generalisability to wider populations.

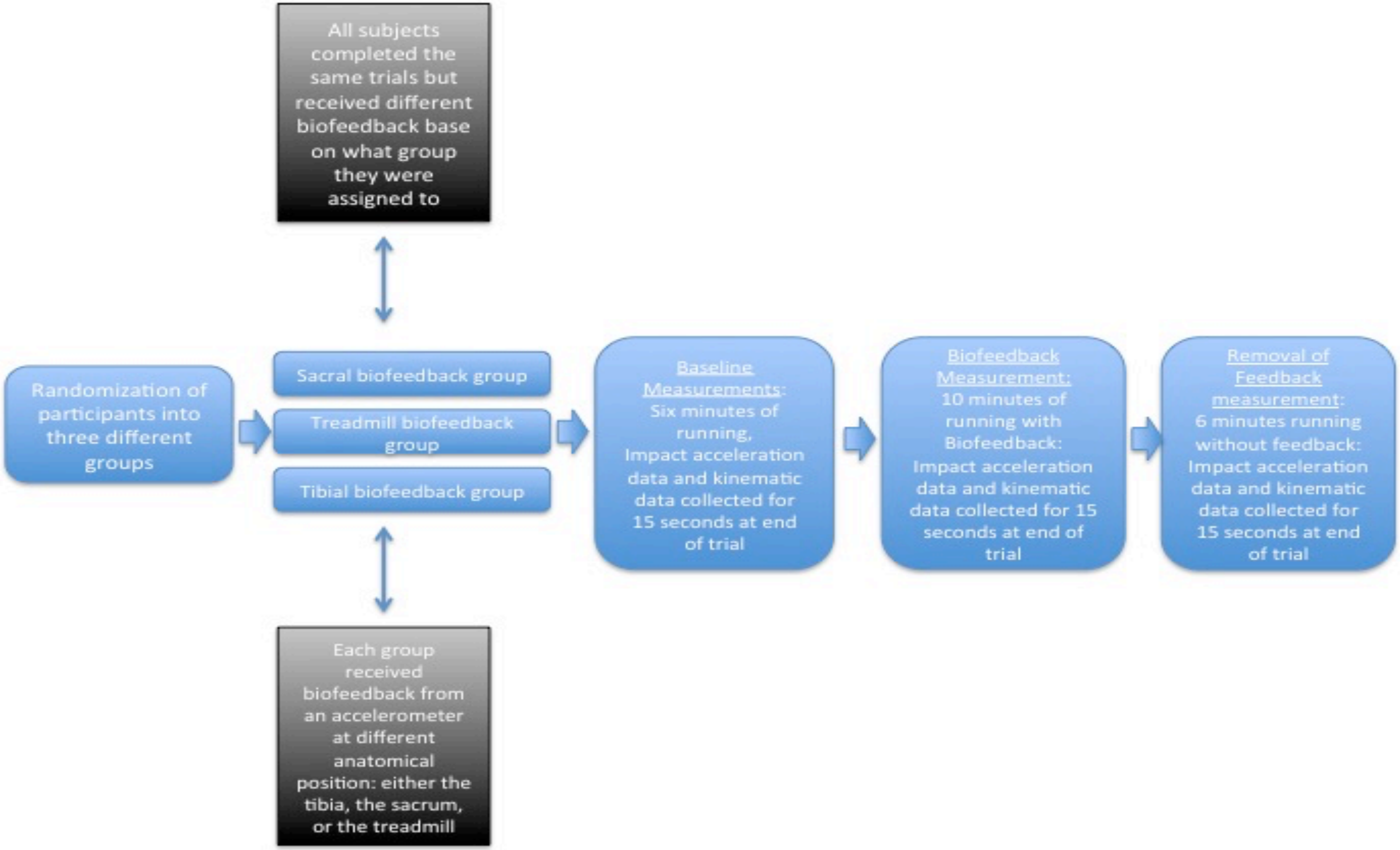

Figure 4.2 1: Experimental procedure

4.2.3 Experimental procedure

The experimental procedure (figure 4.2.1) required participants to attend the biomechanics lab (DCU) for one day of testing. Prior to commencement of testing two uniaxial capacitive accelerometers (Advanced sensors calibration 4421-030, frequency range 1g-200g), mounted to pieces of balsa wood were attached to the proximal anteriomedial aspect of the tibia adjacent to the tibial tuberosity and on the sacrum, horizontally in line with both posterior superior iliac spines. The sensors were attached so that the active axis of each accelerometer was aligned along the axis of the tibia, and along the midline of the spine respectively. A third accelerometer was attached to the treadmill. This accelerometer was positioned with the goal of being as close as possible to the location of each right foot strike, during running. To facilitate this, an area was marked out on the treadmill that represented what was considered an ideal position of foot strike. Participants were not made aware of these markings, however if they veered away from this location during experimental trials, the researcher asked the participant to move forwards or backwards in order to re-enter this area. Tibial and sacral accelerometers were initially attached using double sided tape. Following this, each sensor was pressed against the skin, and while held in this position, an elasticated Velcro strapping was horizontally wrapped around the tibia and sacrum, and subsequently reinforced with zinc oxide tape (figures 4.2.5 and 4.2.6). This attachment method and positioning of accelerometers is similar to methods previously used (Verbitsky, Mizrahi Voloshin, Trieger & Isakov, 1998, Flynn, Holmes & Andrews, 2004; Mizrahi, Werbitski, Isakov & Daily, 2000). Furthermore, this method facilitated appropriate preloading of the accelerometer onto the skin. Preloading reduces measurement errors by minimizing the effect of the accelerometer mass, and ensuring the natural frequency of the accelerometer and attachment equipment are not in the same frequency range as the collected data (Forner-Cordero et al. 2008). Skin mounted accelerometers have been previously shown to significantly overestimate magnitude of accelerations in comparison to more invasive bone mounted accelerometers (Lafortune, Henning & Valiant 1995), however this error is assumed to be consistent across trials, and therefore not a limiting factor. To examine running kinematics participants were marked with a pen at 12 predetermined anatomical positions (as in figures 4.2.3, 4.2.4 and table

4.2.1), retrofelective markers were subsequently attached using double sided tape and 1.5 inch PowerFlex™ self adherent tape (Andover®, PowerFlex™).

For this study, shoes were not standardized across participants. This was done in order to improve ecological validity, as in general and clinical populations participants will wear various different types of footwear (Munro, Miller & Fuglevand 1987).

Finally, running speed has been shown to have an effect on both kinetics and kinematics (Brughelli, Cronin & Chaouachi 2011, Mercer et al. 2005, Arampatzis, Brüggemann & Metzler 1999). This study aims to compare kinematics and kinetics of running under different feedback conditions; speed was therefore standardised.

For all trials, participants ran at a standardized speed of 2.70 m/s (10km/hr) at a 1% incline. This speed was chosen as a result of pilot studies completed in the biomechanics laboratory in DCU, which indicated participants could comfortably complete 22 minutes of running while maintaining unaccustomed running mechanics, developed as a result of an accelerometer based visual biofeedback system. A 1% incline was used, as this has been shown to most closely mimic the effects of air resistance on running economy at a similar speed during outdoor level running (2.92 m/s, 10.5km/hr) (Jones, Doust 1996), thus increasing the generalizability of results to external conditions.

Figure 4.2 2: Anterior view of anatomical marker positions

Figure 4.2 3: Lateral view of anatomical marker positions

Table 4.2. 1: Description of anatomical marker positions

Anatomical marker position	Description
Right and left heel	The calcaneus at the same height above the plantar surface as the toe marker
Right and left Toe	The second metatarsal head, on the mid-foot side of the equinis break between fore-foot and mid-foot
Right and left ankle	The lateral malleolus along an imaginary line down the centre
Right and left knee	The later epicondyle of both knees
Right and left hip	The right and left greater trochanter bilaterally
Right and left shoulder	The right and left acromion process

Figure 4.2 4: Anatomical position of tibial accelerometer (right), preloaded accelerometer attached with elastic Velcro strapping and reinforced with zinc oxide tape (right)

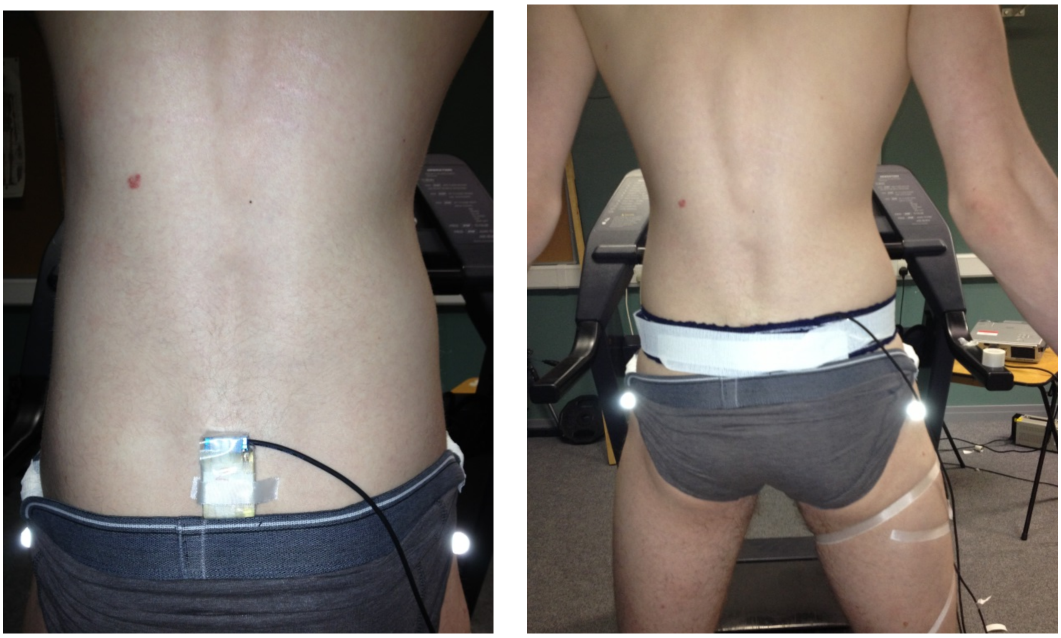

Figure 4.2 5: Anatomical position of sacral accelerometer (left), preloaded accelerometer attached with elastic Velcro strapping and reinforced with zinc oxide tape (right).

4.2.4 Experimental testing

After all sensors were attached appropriately, each participant completed a pre-test, in order to gain baseline data. This involved six minutes of running at 2.70m/s (10km/hr), at 1% incline. Upon completion of these six minutes of running, a 15 second window of kinematic and accelerometer data were collected. Participants were unaware of data collection windows throughout the testing procedure. Six minutes was chosen, as this has been shown to be the required amount of time for kinetic (White, Gilchrist & Christina 2002) and kinematic (Lavcanska, Taylor & Schache 2005) variables, during running, to stabilize. The collection of baseline data allowed each participant to act as their own control. After completion of this six-minute bout, participants were asked to step off the treadmill. Each participant then received a standardized description of the biofeedback protocol that would occur for the next 10-minute bout of running. This informed participants that a screen would be projected in front of them while running (as in figure 4.2.7). This screen contained a live feed of impact acceleration traces from either the tibia, sacrum, or treadmill (dependent on group), where impact acceleration peaks were clearly visible. Participants were informed that they must change the way they run in order to reduce the size of each impact acceleration peak as much as possible. A green horizontal line across the screen represented 50% of the average peak impact acceleration, recorded for that participant during the baseline phase of data collection. This line was present to act as a visual cue, facilitating maximal impact acceleration reduction, as it allowed participants to recognize what technique changes successfully reduced peak impact accelerations. Previous research has used a similar visual biofeedback system (Crowell et al. 2010, Crowell, Davis 2011). Following completion of this ten-minute bout of biofeedback, participants ran for a further six minutes (without rest), without any feedback. Data were collected for two 15-seconds windows, once at the end of the 10-minute biofeedback phase, and once at the end of the 6-minute no-feedback phase.

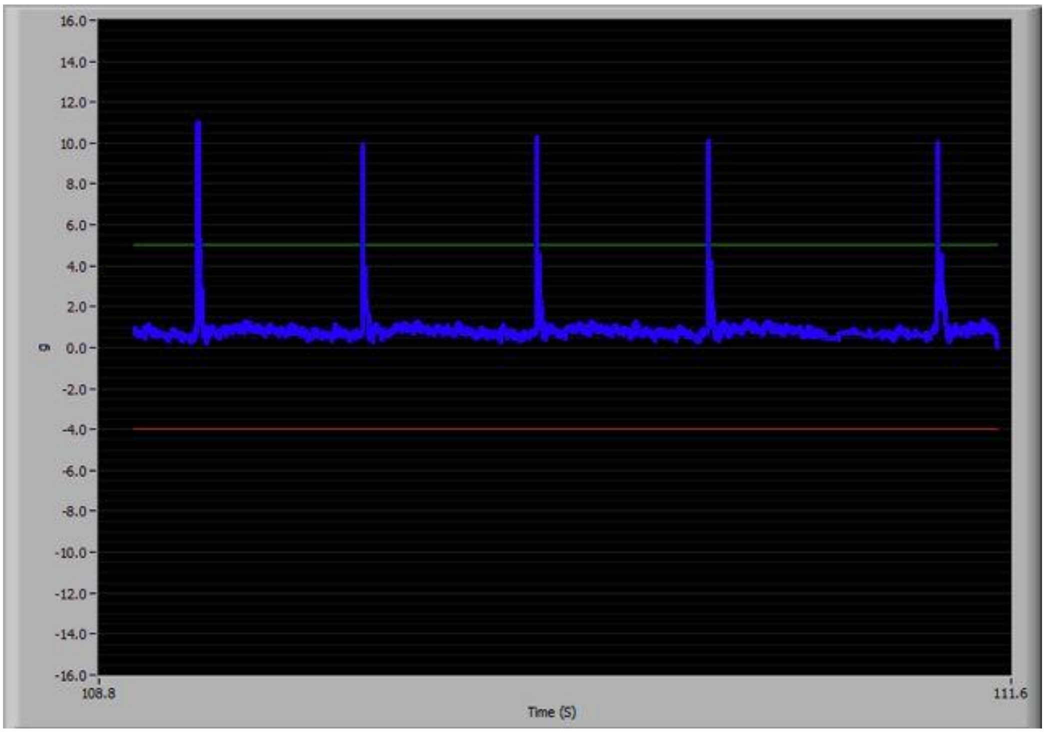

Figure 4.2 6: Example of visual biofeedback screen with the horizontal green line representing 50% of the average impact acceleration peak recorded during baseline phase.

4.2.5 Data collection and processing

Throughout the testing procedure lower extremity kinematics and impact accelerations were measured using a 12-camera 200Hz Vicon motion analysis system (Vicon Oxford Metrics Ltd, Oxford, United kingdom) and three uniaxial capacitive accelerometers (Advanced sensors calibration 4421-030, frequency range 1g-200g), respectively. For the duration of trials, the treadmill was placed in a standardized location and orientation, making contact with a force plate (1200 X 600 mm, 2000 Hz, Advanced Mechanical Technology Inc., Watertown, MA). This facilitated identification of initial contact for each foot strike. Twelve retrofelective markers were tracked (Figure 4.2.3 and 4.2.4, Table 4.2.1) by the 12- camera system. Two dimensional data from each camera is then combined, by Vicon workstation with calibration data to convert the equivalent digital motion information into three dimensions.

A custom MATLAB program (The MathsWorks Inc., United Kingdom) was used to calculate both lower extremity kinematics and to identify impact acceleration peaks. From each 15-second window of data collection, the custom MATLAB program (The MathsWorks Inc., United Kingdom) identified every consecutive foot strike. This data were then averaged fro every participant. 4.2.6 Summary of dependent variables

4.2.6 Dependent variables

For each stride the following variables were measured. Note: data were collected for 15-second windows and averaged for each variable.

- Ankle, knee, and hip angle
- Vertical height of COM
- Peak tibial and sacral acceleration

4.2.7 Data analysis

In order to examine the effect of different forms of feedback across 3 time points on tibial and sacral impact acceleration values two 3*3 (time * biofeedback location) within-between repeated measures ANOVA were completed. The within participant factor was time (baseline *versus* biofeedback *versus* removal of biofeedback) and the between participant factor was biofeedback location (tibial biofeedback *versus* sacral biofeedback group *versus* treadmill feedback group). For all tests, statistical significance was set at $p<0.05$. Normality of data was determined using the Shapiro-Wilk test for normality. Mauchleys test was used to examine sphericity. In cases where the assumption of sphericity was violated a Greenhouse-Geisser correction was employed. Homogeneity of variance (HOV) was examined using Levene's test. Violations to HOV were deemed acceptable as all groups have the same sample size (n=9), as in accordance Stevens (1996).

While traditionally key discrete events in kinematics data are statistically analysed, the present study statistically examined the whole stride data curve. This was completed using an Analysis of Characterizing Phases (Richter et al., 2014). This method detects variation (key phases) within each sample of curves, which are subsequently used to generate scores (similarity scores). These similarity scores are computed for each key phase by calculating the area between each curve and the mean curve across the data set, for every point within each key phase. Key phases were identified using information generated by a principal component analysis; these components described 99.5% of the variance within the data (Richter et al., 2013a). For further explanation of Analysis of Characterizing Phases the reader is referred to Richter et al. (2014). The reader should note that the calculation of participant scores within this thesis differs slightly from Richter et al. (2014). This was done in order to overcome a dependency of the finding on the reference signal chosen. In Richter et al. (2014) the best jump was selected as a reference signal because the participant score calculation used absolute values to measure similarity. This approach assumes that altering a curve towards the reference signal has a positive effect on the dependent variable. However, this might not be true for running as other movement strategies might represent a better movement solution. The score generation approach used in the present paper overcomes this limitation and findings are not dependent on the reference signal. The overall mean was selected as the reference signal because it is commonly used and easy to relate when interpreting the findings. Finally, Key phases were extended if significant differences were found following multiple T-tests. A challenge with continuous curve analysis is that different key phases are identified between each time point (baseline, biofeedback, removal of biofeedback) and between each biofeedback location, therefore multiple paired sample T-tests (with a Bonferonni adjustment) were completed for the within participant comparisons and independent sample T-tests were completed for between participant comparisons.

4.3 Results

4.3.1 Peak acceleration results

A mixed between-within repeated measures ANOVA was conducted to assess the impact of three different biofeedback locations (tibial, sacral, treadmill) on peak tibial and sacral acceleration values across three time points (baseline, biofeedback, no biofeedback). Tests indicate no significant biofeedback location*time interaction at the tibia ($F(3.223, 38.675) = 2.170$, $p = 0.103$, partial eta squared = 0.153) or sacrum ($F(4,48) = 1.605$, $p = 0.188$, partial eta squared= 0.118).

At the tibia (figure 4.3.1) there was a significant main effect for both time ($F(1.612, 38.676) = 10.304$, $p<0.05$, partial eta squared = 0.300) and location ($F(2, 24) = 3.965$, $p<0.005$, partial eta squared= 0.248). Pairwise comparisons revealed a significant location difference with tibial biofeedback 30% greater than treadmill biofeedback at baseline. In addition, baseline measurements were significantly greater than both biofeedback measurements (16%, $p<0.05$), and removed biofeedback measurements (14%, $p<0.05$), for all biofeedback locations.

At the sacrum there was a significant main effect for time ($F(2,48) = 14.029$, $p<0.05$, partial eta squared= 0.369) but not location ($F(2, 24) = 0.347$, $p=0.711$, partial eta squared= 0.028). Pairwise comparisons revealed that baseline measurements were significantly larger than biofeedback measurements (16%, $p<0.05$), and biofeedback measurements were significantly smaller than removed biofeedback measurements (9%, $p<0.05$), for all biofeedback locations.

Retrospective power analysis indicates that an observed power of 0.597 is present for tibial accelerations and 0.457 for sacral accelerations with regard to the determination of a significant interaction effect; thus indicating that the sample may be underpowered to identify such. With regard to within participant differences there was an observed power of 0.96 for tibial accelerations and 0.996 for sacral accelerations, both of which are satisfactory.

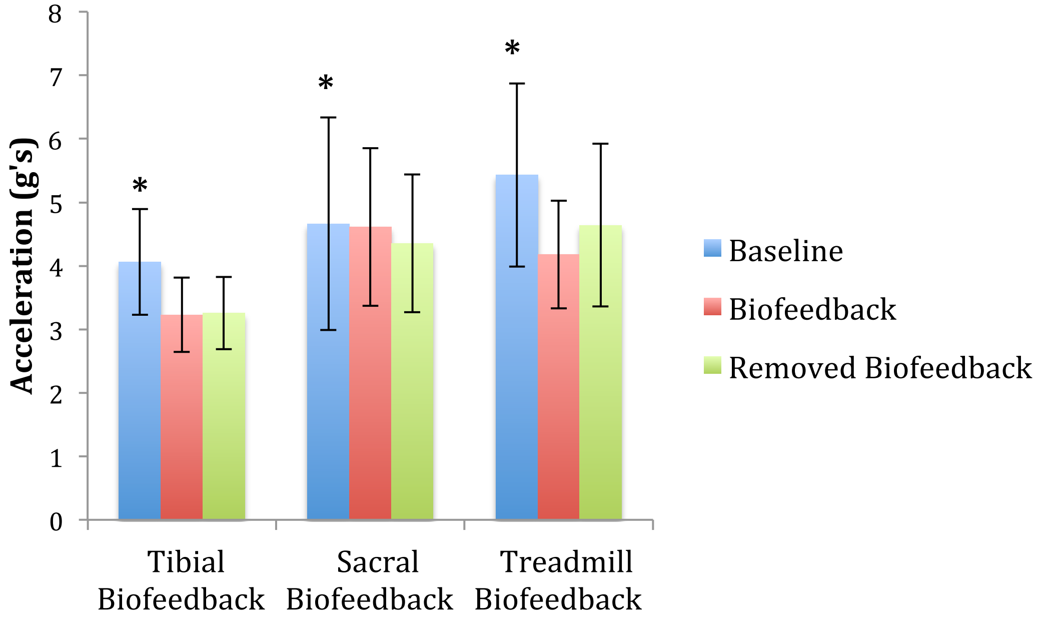

Figure 4.3. 1: The effect of different biofeedback locations on peak tibial accelerations (Mean + Standard deviation) (* indicates significant difference to the other two time points within the same biofeedback location).

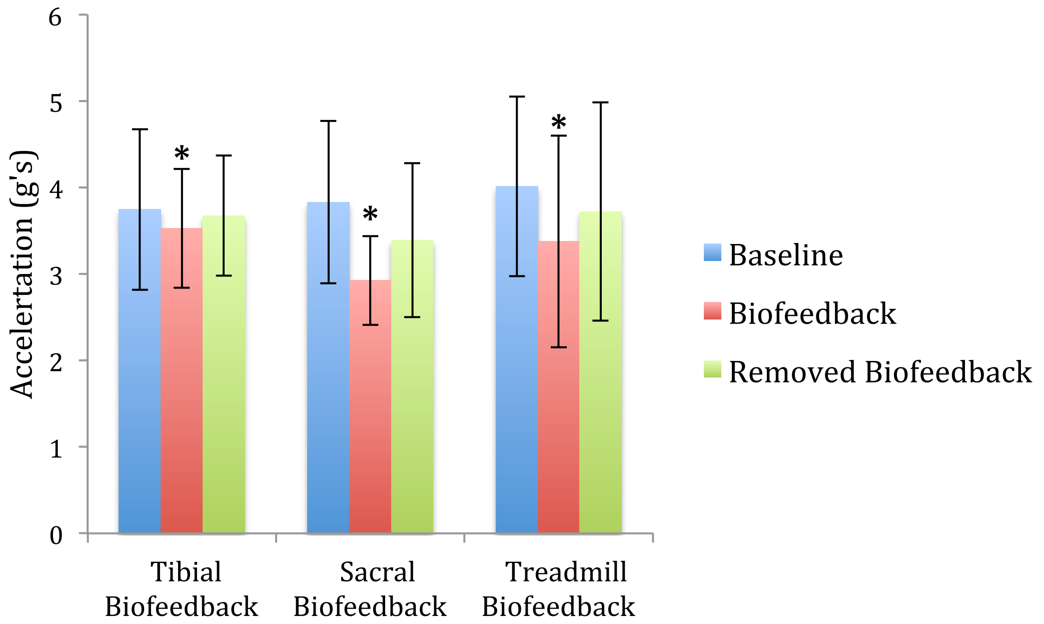

Figure 4.3. 2: The effect of different biofeedback locations on peak sacral accelerations (Mean + Standard deviation) (* indicates significant difference to the other two time points in the same group, p<0.05)

Table 4.3. 1: Differences between conditions for each group (* indicates a significant difference, p<0.05)

Group	Location	Baseline Vs. Biofeedback (% Diff.)	Baseline Vs. Removed Biofeedback (% Diff.)	Biofeedback Vs. Removed Biofeedback (% Diff.)
Tibial Biofeedack	Tibia	23%* (Baseline greater)	22%* (Baseline greater)	1% (Removed Biofeedback greater)
	Sacrum	6%* (Baseline greater)	2% (Baseline greater)	4%* (Removed Biofeedback greater)
Sacral Biofeedback	Tibia	1 % * (Baseline greater)	7%* (Baseline greater)	6% (Biofeedback greater)
	Sacrum	27%* (Baseline greater)	12% (Baseline greater)	24%* (Removed Biofeedback greater)
Treadmill BiofeedbacK	Tibia	26%* (Baseline greater)	16%* (Baseline greater)	10% (Removed Biofeedback greater)
	Sacrum	17%* (Baseline greater)	8% (Baseline greater)	10%* (Removed Biofeedback greater)

4.3.2 Kinematics

4.3.2.1 Tibial Biofeedback group

Participants in the tibial biofeedback group demonstrated a significantly more dorsiflexed ankle during for 1-21% of the cycle at baseline measures compared to biofeedback measures (T=3.19, p<0.016, partial eta squared =0.82).

The tibial biofeedback group demonstrated a less flexed knee during 44-66% of cycle at biofeedback when compared to baseline measurements (T= 2.79, p<0.016, partial eta squared= 0.51). Following removal of biofeedback participants displayed a larger degree of knee flexion during 22-35% of cycle (T=3.51,p<0.016, partial eta squared= 0.49).

The tibial biofeedback group displayed a significantly lower COM during 67-98% of cycle following biofeedback when compared to baseline measurements (T=3.54, p<0.016, partial eta squared= 0.27). Similarly, following removal of biofeedback participants displayed a significantly lower COM during 66-99% of cycle, relative to baseline (T=5.92,p<0.001, partial eta squared= 0.26).

4.3.2.2 Sacral Biofeedback group

The sacral biofeedback group displayed a significantly larger degree of ankle dorsiflexion, during 23-41% of cycle, following biofeedback relative to baseline (T=3.96, p<0.016, partial eta squared =1.49). Similarly, participants displayed a significantly larger degree of dorsiflexion relative to baseline for 25-36% of cycle, following the removal of biofeedback (T=2.72, p<0.16, partial eta squared=0.99). A larger degree of dorsiflexion was also present following biofeedback in comparison to removal of biofeedback for 29-42% of cycle (T= 3.27,p<0.16, partial eta squared=0.95) and 64-85% of cycle (T=-3.63, p<0.016, partial eta squared=0.88).

The sacral biofeedback group displayed a significantly larger degree of knee flexion during 20-40% of cycle (T=3.73, p<0.016, partial eta squared=1.19) following biofeedback, relative to baseline. Following biofeedback participants displayed a significantly larger degree of knee flexion during 85-99% of cycle (T=4.42, p<0.016, partial eta squared=0.49) and 18-40% of cycle (T=3.62, p<0.016, partial eta squared=0.73) when compared to the removal of biofeedback.

Following biofeedback participants displayed a significantly larger degree of hip flexion during 17-40% of cycle ($T=3.33$, $p<0.016$, partial eta squared= 1.28) and 80-100% of cycle ($T=3.05$, $p<0.016$, partial eta squared= 1.26), relative to baseline. Participants also displayed a significantly larger degree of hip flexion during 11-35% ($T=3.15$,$p<0.016$, partial eta squared=0.83) and 80-100% of cycle ($T=3.77$,$p<0.016$, partial eta squared=0.86) following biofeedback, relative to removal of biofeedback.

Following biofeedback the sacral biofeedback group displayed a significantly lower COM relative to baseline measurements for 1-30% of cycle ($T=3.7$,$p<0.016$, partial eta squared= 0.58) and 66-100% of cycle ($T=4.28$,$p<0.016$, partial eta squared= 0.75) and relative to no feedback measurements for 26-100% of cycle ($T=2.93$, $p<0.016$, partial eta squared= 0.36). Following removal of biofeedback participants displayed a significantly lower COM for 2-40% of cycle ($T=4.16$, $p<0.016$, partial eta squared=0.44) and 69-100% of cycle ($T=4.92$, $p<0.016$, partial eta squared= 0.44), relative to baseline.

4.3.2.3 Treadmill Biofeedback group

Following treadmill biofeedback participants displayed a significantly larger degree of ankle dorsiflexion for 21-34% of cycle ($T=3.63$,$p<0.016$, partial eta squared= 1.29), a significantly larger degree of knee flexion for 18-29% ($T=3.11$, $p<0.016$, partial eta squared=0.96), and a significantly larger degree of hip flexion for 6-31% of cycle ($T=4.27$,$p<0.016$,partial eta squared=0.81) when compared to baseline.

4.3.2.4 Biofeedback location comparisons

Comparison of biofeedback locations at each time point (baseline, biofeedback, no biofeedback) indicates that the group using tibial biofeedback demonstrated a smaller degree of dorsiflexion for 20-38% of cycle ($T=3.06$, $p<0.016$, partial eta squared=1.27), a smaller degree of knee flexion for 21-39% of cycle ($T=2.48$, $p<0.016$, partial eta squared= 1.13) and a smaller degree of hip flexion for 68-94%($T=3.45$, $p<0.016$, partial eta squared=1.37) and 1-39% of cycle ($T=3.17$, $p<0.016$, partial eta squared=1.31) when compared to the sacral biofeedback group. Tibial biofeedback also demonstrated a smaller degree of knee flexion for 19-32% ($T=2.41$, $p<0.016$, partial eta squared=1.16) in comparison to the treadmill biofeedback group.

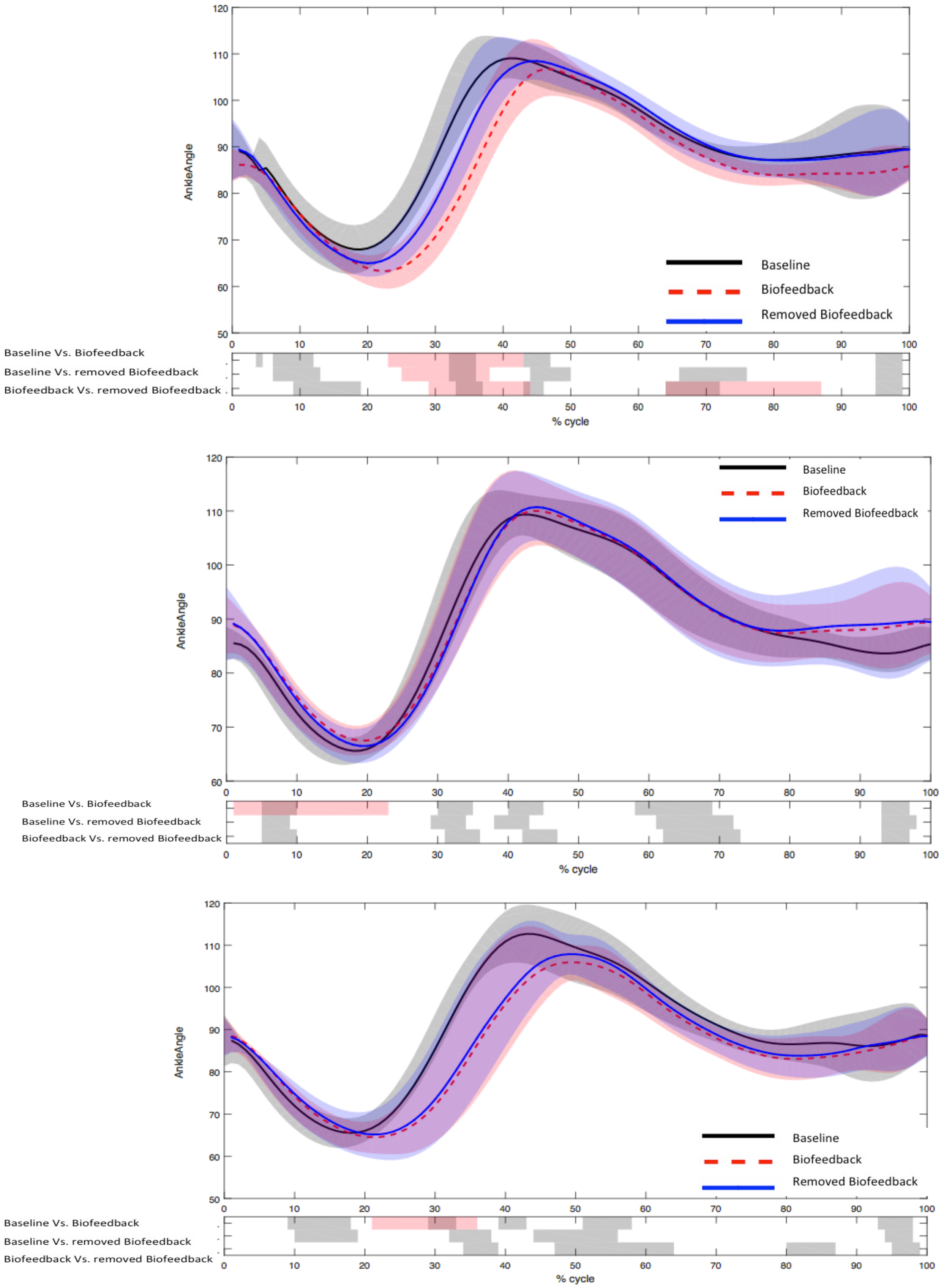

Figure 4.3. 3: Change in ankle angle (degrees) across time points for each group: Top (sacral biofeedback), middle (tibial biofeedback), bottom (treadmill biofeedback). Regions highlighted in red (underneath curve) indicate significance.

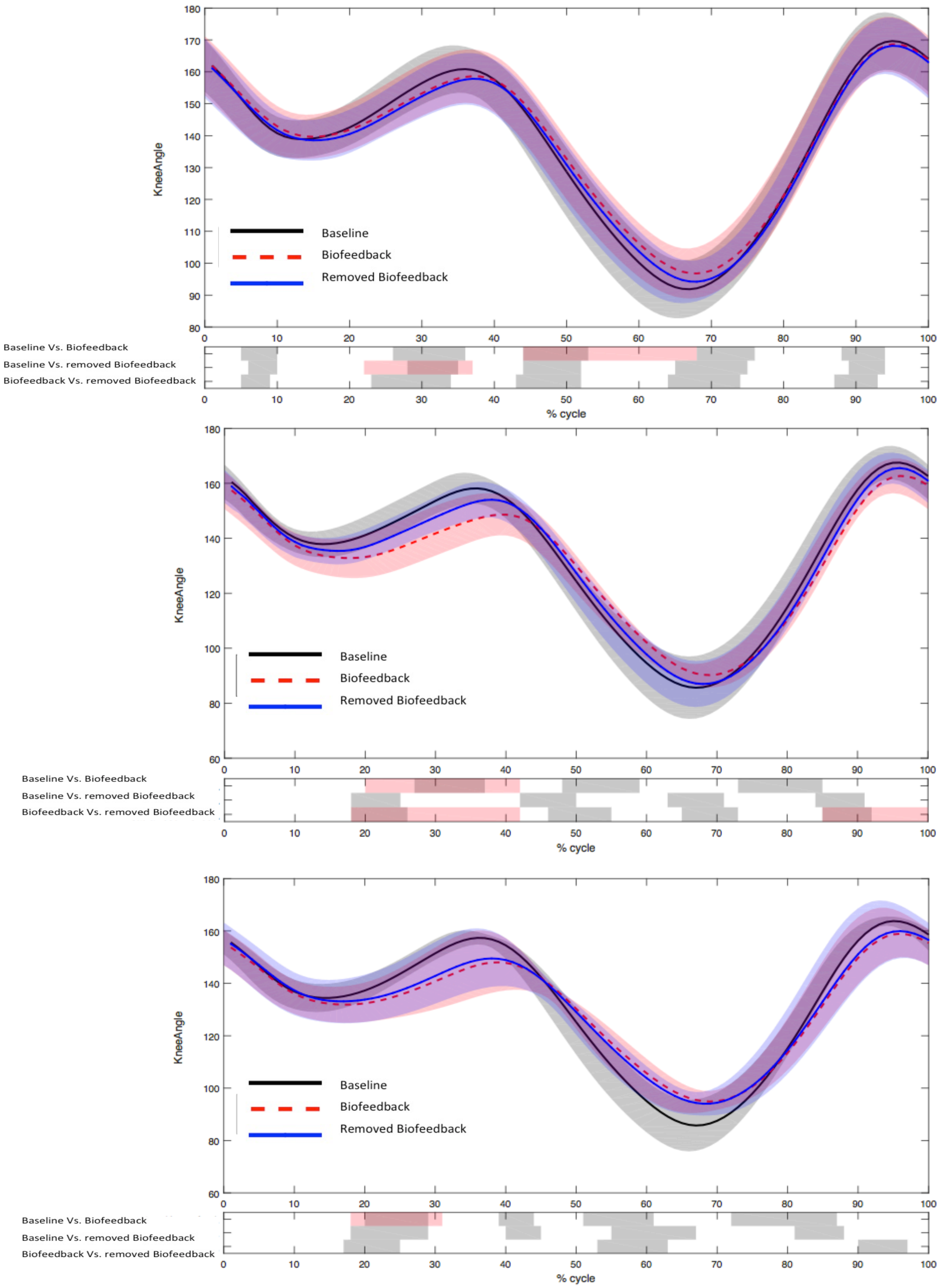

Figure 4.3. 4: Change in knee angle (degrees) across time points for each group: Top (tibial biofeedback), middle (sacral biofeedback), bottom (treadmill biofeedback). Regions highlighted in red indicate significance.

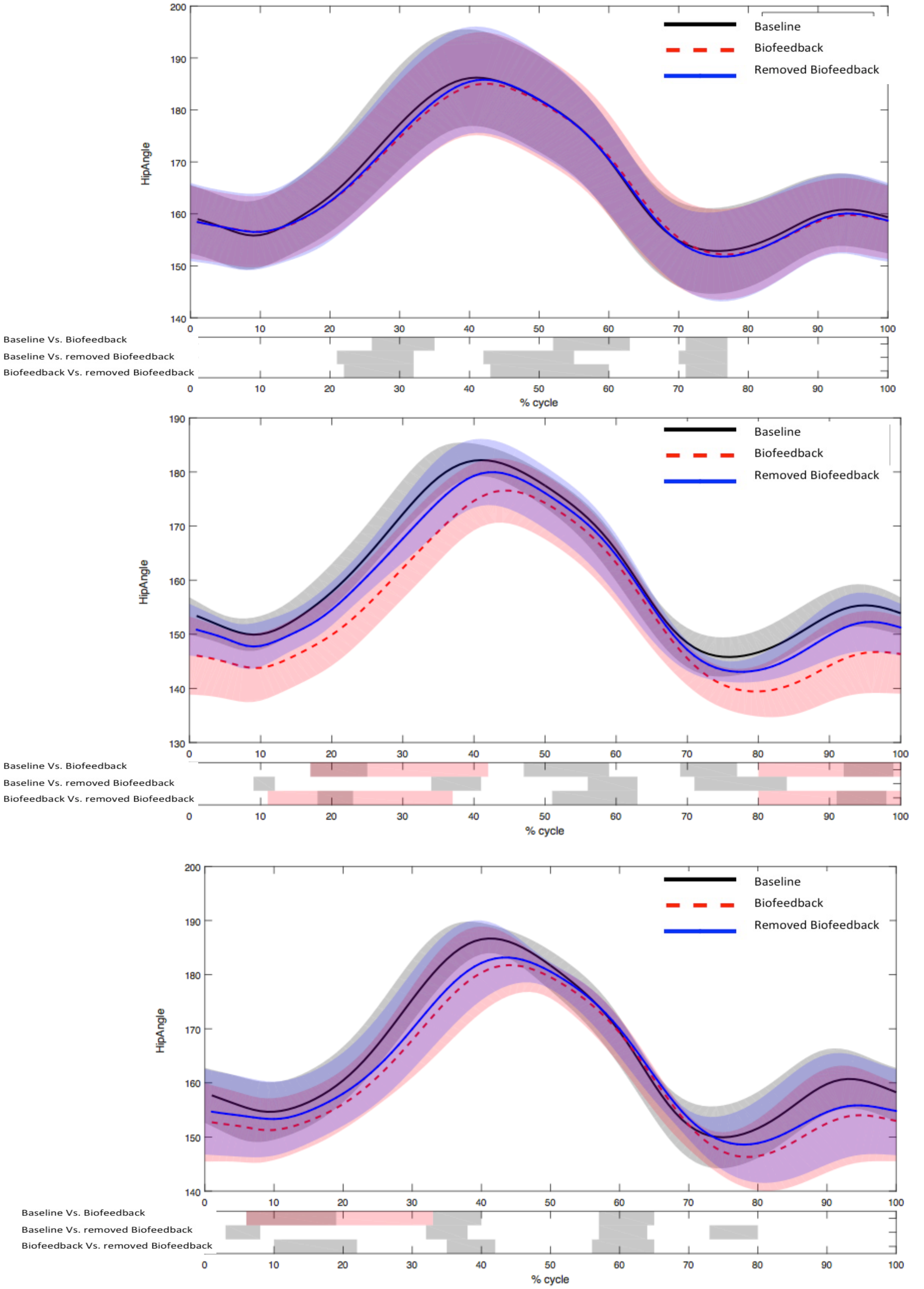

Figure 4.3. 5: Change in hip angle (degrees) across time points for each group: Top (tibial biofeedback), middle (sacral biofeedback), bottom (treadmill biofeedback). Regions highlighted in red indicate significance.

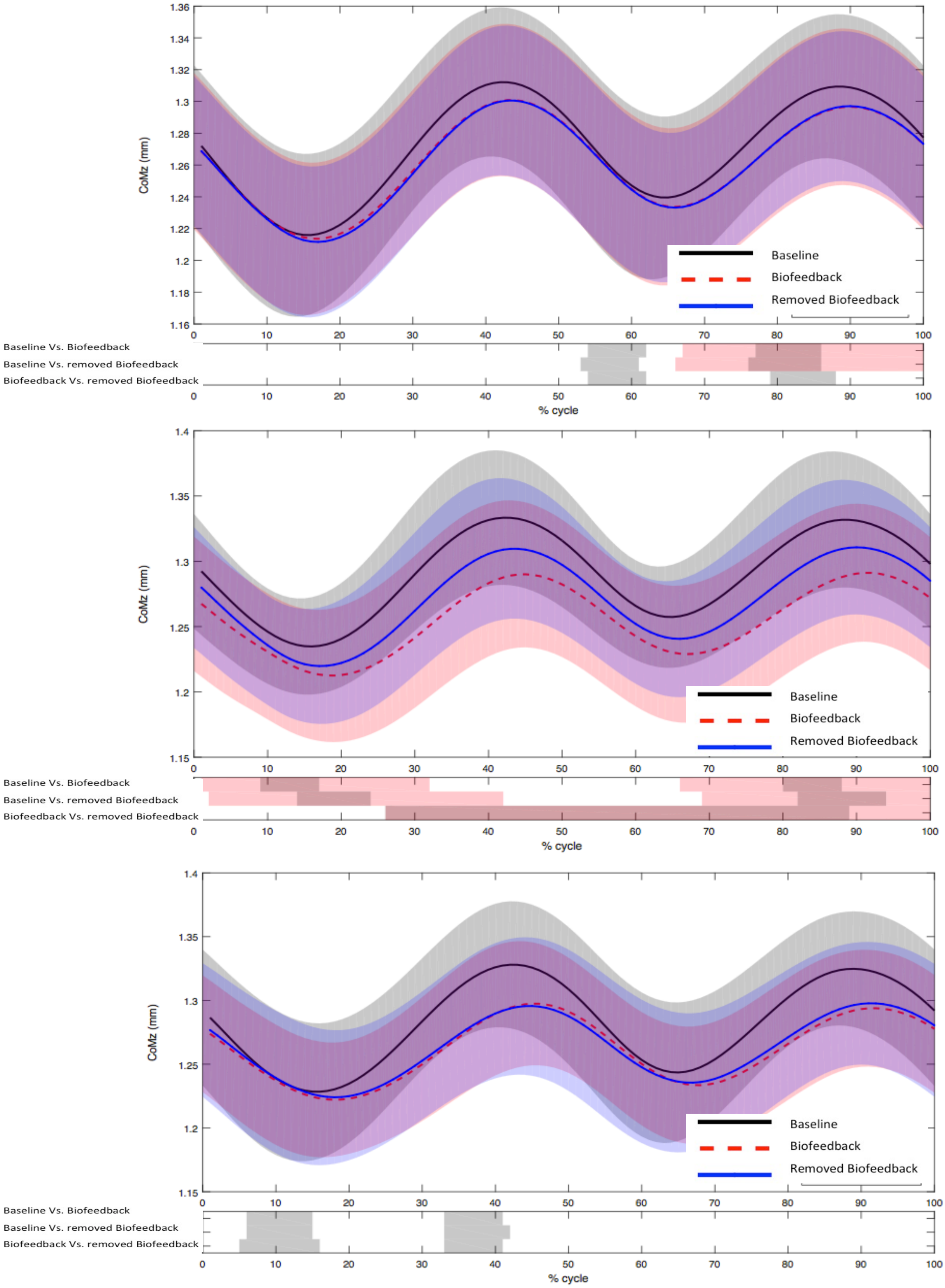

Figure 4.3. 6: Change COM height (metres) across time points for each group: Top (tibial biofeedback), middle (sacral biofeedback), bottom (treadmill biofeedback). Regions highlighted in red indicate significance.

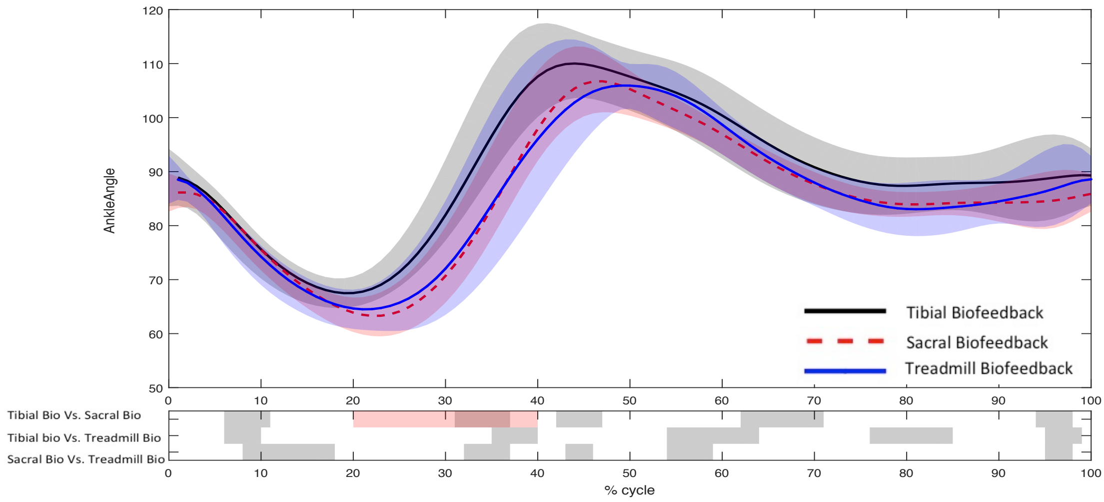

Figure 4.3. 7: Difference in ankle angle (degrees) between groups at feedback measures. Regions highlighted in red indicate significance.

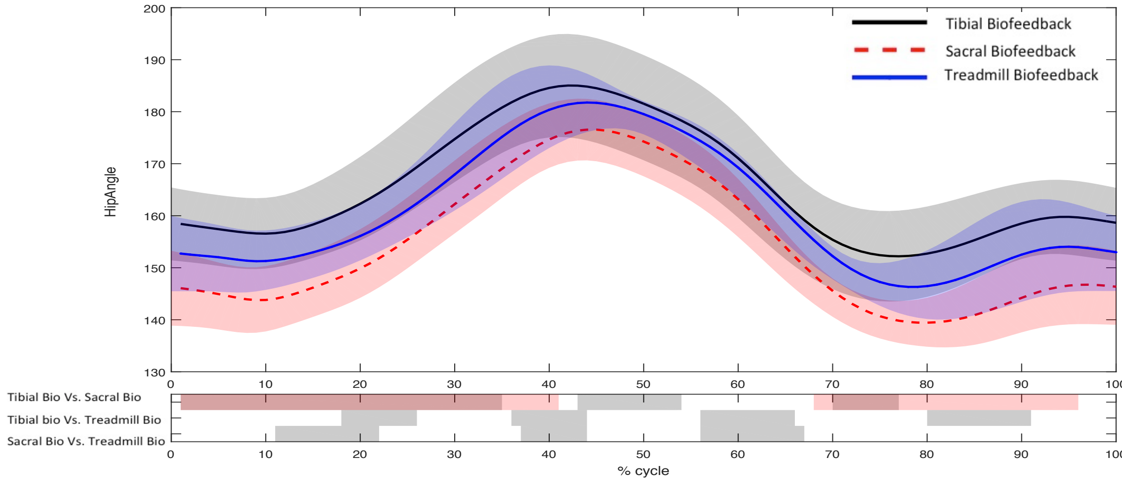

Figure 4.3. 8: Difference in hip angle (degrees) between groups at feedback measures. Regions highlighted in red indicate significance.

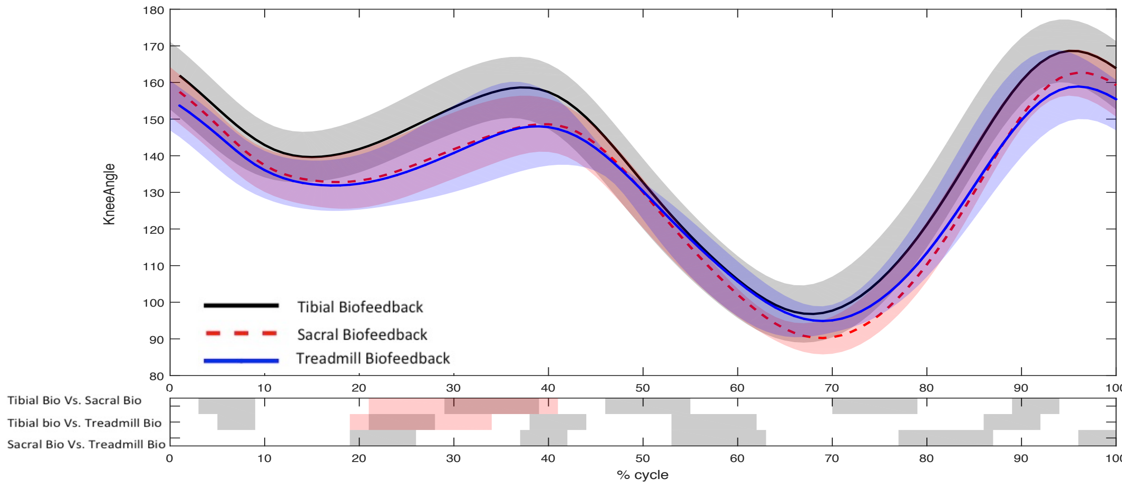

Figure 4.3. 9: Difference in knee angle (degrees) between groups at feedback measures. Regions highlighted in red indicate significance.

4.4 Discussion

4.4.1 Peak accelerations

The primary aim of this study was to examine the use of three different biofeedback locations (tibial, sacral, and treadmill) in relation to their ability to reduce impact acceleration values at the tibia and the sacrum, and to examine what kinematic changes were made to facilitate these decreases.

The use of biofeedback resulted in an average decrease of 16% in peak tibial accelerations (in comparison to baseline). This is slightly larger than that displayed following auditory tibial biofeedback (11%)(Woof & Kipp, 2014), but smaller than a very similar visual biofeedback study by Crowell et al (2010) (26%; calculated based on graphs presented). The larger decrease displayed by Crowell may have numerous explanations including their study having: a very small sample size, difference biofeedback locations, larger baseline acceleration values, and the additional inclusion of verbal feedback.

The small sample size (5 participants versus 27 participants in this study) in Crowell et al's work, allows the extremely large decrease found in two of the five participants (60% and 54%) to have a much larger effect on the average group decrease, thus potentially limiting the generalizability of their results to larger populations. Secondly, Crowell et al (2010) employed verbal feedback in conjunction with the provided tibial biofeedback. Verbal feedback has previously been shown to be an effective method of altering running style (Fletcher et al. 2008, Goss, Gross 2012, Dallam et al. 2005), thus without the presence of a control, it is unclear in Crowell's work whether the decreases were due to the tibial biofeedback, the verbal feedback or a combination of the two. Verbal feedback was not used in the current study as this would require someone to stand next to the runner, adding significant cost to the process, making it less likely to be adopted. Thirdly, all participants in Crowell's study received tibial biofeedback, whereas participants in this study were split into three different groups so that 9 received tibial biofeedback, 9 received sacral biofeedback, and 9 received treadmill biofeedback. Although no main biofeedback location by time interaction effect was evident, examination of percentage differences for each group shows that the tibial biofeedback group presented decreases of 23%, the sacral biofeedback group displayed a decrease of 1%, and the treadmill biofeedback group displayed a decrease of 26%, for tibial acceleration values. Therefore, both tibial and treadmill biofeedback locations displayed similar reductions as Crowell et al (2010). Finally, average baseline tibial acceleration values presented by Crowell et al (2010) are almost double the average baseline tibial acceleration values reported in this study (8.6g's versus 4.7g's), thus this method of providing biofeedback to reduce impact acceleration values at the tibia may be more effective for participants that are outside the normal range (3-8g) for tibial acceleration values (Davis, Milner & Hamill 2004, Hennig, Milani & Lafortune 1993, Milner et al. 2006b).

Another important consideration is the ability of participants to maintain reductions in peak accelerations once the biofeedback has been removed. This is a more ecological measure, since it is unlikely that someone would use the biofeedback long term or would not want to run outdoors, and may be indicative of motor learning. Comparison of tibial impact accelerations at baseline to removal of biofeedback indicates that for all biofeedback locations, participants were able to maintain the decreases produced by biofeedback. Crowell et al (2010) demonstrated a similar pattern for 3 of 5 participants, with the other 2 displaying further decreases in peak tibial accelerations following the removal of biofeedback. This is an important consideration given the association between the magnitude of tibial acceleration values and running related injuries (Davis et al., 2010). Seven of the 26 participants in the present study also reduced the magnitude of tibial accelerations following biofeedback removal. There are 2 possible explanations for this. Firstly, participants were unaware of data collection windows and therefore may have been experimenting with a less effective technique during biofeedback measures. Secondly, the effectiveness of the kinematic strategies employed by participants may have improved due to a longer practice period. This is supported by research indicating that volume of practice is an important factor in skill development and motor learning (Winstein 1991, Salmoni, Schmidt & Walter 1984, Porte et al. 2007).

An important factor, which was not considered by Crowell et al (2010), is the effect that biofeedback may have on loads further up the musculoskeletal system. To account for this, peak positive acceleration was also measured at the sacrum within the present study. Sacral impact accelerations decreased as a result of the biofeedback protocol (-16%), but returned to almost baseline levels once the feedback was removed; therefore not following the same pattern as observed at the tibia. This highlights the importance of measuring impact acceleration values at more than one anatomical location, as changes at the tibia may not be reflective of loading further up the body. This may be due to increased fatigue (above the knee) resulting in a diminished capacity for muscle of the upper leg to attenuate impact accelerations. This is supported by research demonstrating the role of muscle action in the attenuation of impact loading (Shorten, Winslow 1992, Paul et al. 1978, Jefferson et al. 1990) in combination with research indicating an increase in sacral accelerations (but not tibial) following a bout of downhill running (Mizrahi, Verbitsky & Isakov 2000). Downhill running is associated with larger mechanical stress (Nurenberg et al. 1992, Iversen, McMahon 1992) and thus a decreased ability to reduce sacral accelerations. More compliant mechanics (as employed in the present study) have been associated with increased energy expenditure (McMahon et al., 1987 and study 1) and therefore potentially increased muscle damage, and a reduced ability to attenuate sacral accelerations. However it should be highlighted that the increased sacral values are increased relative to biofeedback measures and thus may not pose an additional threat in terms of injury development.

Perhaps the most important finding is the implication of a reduced injury risk due to the displayed reductions in tibial (-16%) and sacral (-16%) accelerations following biofeedback. Prospectively, it has been shown that runners who suffer tibial stress fractures or any running related injury present peak tibial acceleration values 63% greater and 81% greater than uninjured controls, respectively (Davis, Milner & Hamill 2004, Davis, Bowser & Hamill 2010). These results are consistent with retrospective research indicating that participants with history of tibial stress fracture injury display peak tibial acceleration values 13-28% greater than participants with no history of injury(Milner et al. 2006b, Pohl et al. 2008, Zifchock, Davis & Hamill 2006b, Milner, Hamill & Davis 2007). Therefore, an acute bout of biofeedback (regardless of the location of the accelerometer to provide the feedback) may decrease the risk of tibial stress fracture and general running injury development.

A popular method for the reduction of impact loads is the use of shock absorbing insoles, which have been shown to display similar reductions in tibial accelerations as those presented in the current study (11-20%)(Butler et al., 2006, Milani et al., 1997). However, consideration of tibial biofeedback (-23%) and treadmill biofeedback (-26%) alone indicates slightly larger reductions, which are expected to become larger following an extended intervention period (versus the acute nature of the present study), as has been displayed by Crowell et al (2011) (-48% following 8 biofeedback sessions across 2-weeks).

It is clear that tibial impact acceleration values may not reflect changes in load further up the body, and thus consideration of sacral accelerations values may have important implications with regard to injury development. Prospective research has shown that in a 13 week period 14% of male and 15% of female recreational runners may sustain an overuse injury to the lower back/pelvic region. Sacral impact accelerations have been suggested as a measure of impact experienced by the centre of mass (Henriksen et al. 2008), thus potentially representing whole body load, and risk of injury development. Therefore the found decrease in sacral impact acceleration values, for all biofeedback locations, of 16% from baseline to biofeedback measures, suggests that this biofeedback system may be an effective tool for decreasing the likelihood of developing an overuse injury. Sacral impact accelerations may also contribute to the development of pelvic stress fractures, which account for 7% of stress factures developed in normal populations and about 4% in track and field athletes (Hosey, Fernandez & Johnson 2008). Although, this is a relatively small percentage, the incidence of pelvic stress fractures is more prevalent in female military recruits and the elderly, with suggested figures of up to 22% experiencing such injuries (Hosey, Fernandez & Johnson 2008). Research has shown that participants diagnosed with sacral stress fractures can make a rapid return to sport following a rehabilitative programme involving low impact that is gradually increased over a six-week period, facilitating return to sport in the seventh week(Knobloch et al. 2007). This indicates that the biofeedback system employed in the present study may be a useful rehabilitative tool, allowing physiotherapists or healthcare professionals to monitor and control the magnitude of load experienced by a participant, as well as providing a means of altering mechanics to facilitate the required decreased loading.

One of the primary goals of the present study was to investigate what location (tibia, sacrum, or treadmill) for providing visual impact acceleration based biofeedback, most effectively alters running style to decrease load on the body. Following statistical analyses, there was no significant biofeedback location*time interaction, thus indicating that the change in acceleration values at the tibia and sacrum across time did not differ between different biofeedback locations. However, when examining variance in load, it has been suggested that very minor, insignificant differences, may play a major role in the development of overuse injuries, due to the extremely repetitive nature of running and the accumulative effect this may have over time(Milner et al. 2006a) (as discussed in study 1 section 3.4). Therefore, any differences (although statistically insignificant) between the locations with regard to reducing the magnitude of impact accelerations, may have important implications in terms of this biofeedback systems ability to reduce the potential risk for injury development.

Results indicate that the largest reductions in tibial impact accelerations occur as a result of treadmill biofeedback (Treadmill=🡻 26% *versus* Tibial= 🡻23% *versus* Sacral= 🡻1%), however participants in the tibial biofeedback group were able to maintain decreases more effectively following biofeedback removal (Tibia= 🡻 22% *versus* Treadmill=🡻 16 *versus* Sacral= 🡻 7%). With regard to sacral accelerations, sacral biofeedback displayed the largest initial reductions (Sacral= 🡻 27% *versus* Treadmill= 🡻17% *versus* tibial= 🡻 6%) and the greatest ability to maintain reductions following biofeedback removal Sacral= 🡻 12% *versus* Treadmill= 🡻8% *versus* tibial= 🡻 2%).

An important consideration may be how the magnitude of baseline accelerations affects the ability of participants to subsequently reduce these values following biofeedback. In the present study, the treadmill biofeedback group present significantly larger baseline tibial accelerations suggesting that it may be easier to make decreases in impact acceleration values for participants who are at a higher risk of injury development (as a result of larger impact acceleration values).

These results also indicate that it may be appropriate to prescribe different forms of biofeedback to different participants depending on what anatomical locations may be at risk of developing an overuse, impact related, injury. For example, participants at risk of developing tibial stress fractures or have a history of tibial stress fracture may be prescribed a treadmill or tibial biofeedback based intervention whereas a participant presenting with a history of pelvic or sacral stress fractures may benefit more from an intervention providing sacral biofeedback. It should however be noted that participants who suffer general running injuries (overuse impact related injuries resulting from running activities) have been shown to present tibial impact acceleration values 81% higher than an uninjured population (Davis, Milner & Hamill 2004). This suggests that provision of an acute bout of either treadmill or tibial biofeedback may most effectively decrease the risk of overall injury development.

The goal of this study was to determine what form of biofeedback is most effective at reducing whole body loading. With this in mind, observation of tibial and sacral acceleration reductions alone portrays a somewhat ambiguous picture. For this reason, it may be appropriate to examine the change in the mean of both tibial and sacral impact accelerations. Therefore, by averaging the reductions for tibial and sacral accelerations for each group (as displayed in table 4.3.1) between baseline and biofeedback measurements, we can see that the largest decrease occurs following treadmill biofeedback (🡻 22%), followed by tibial biofeedback (🡻 15%), and sacral biofeedback (🡻 14%). Therefore treadmill biofeedback may be the most suitable form of feedback for decreasing whole body loading. However, consideration of reductions following removal of biofeedback indicates that the treadmill and tibial biofeedback group displayed the same decreases in the mean tibial/sacral impact acceleration values (both 🡻 12%) and the sacral biofeedback, again, presented the lowest overall decrease (🡻 10%). This indicates that treadmill biofeedback provides the largest acute decreases in impact loading, but appears to display a poorer ability to facilitate motor learning (to the same extent as the tibial biofeedback group).

4.4.2 Kinematics

Examination of kinematic data displayed for each biofeedback location group offers an insight into the strategies employed by participants to bring about the observed changes to peak accelerations at the tibia and sacrum. It should be noted that kinematic data indicates that toe-off occurs in the region of 30-36% of the cycle, as previously reported (Riley et al., 2008).

Kinematics data indicates that the tibial biofeedback group attempted to reduce tibial accelerations mainly via reduced COM oscillation, as displayed by a significantly lower COM height for 67-98% of the cycle for feedback measures (partial eta squared= 0.27) and 66-99% (partial eta squared= 0.26) following removal of biofeedback; both of which includes the phase of maximum COM height.

Reduced vertical oscillation of COM in Groucho running has previously been shown to reduce vertical landing velocity (0.5-0.8m/s in normal running versus almost 0m/s in Groucho running) (McMahon et al.,1987). Therefore manipulation of the impulse-momentum relationship reduces impact loading and may potentially explain the significant decrease in tibial (-23%) and sacral (-6%) accelerations for the tibial biofeedback group. Reduced COM vertical motion appeared to be aided by a less flexed knee during early swing phase (44-66% of cycle) (partial eta squared= 0.51), potentially allowing participants to keep the foot closer to the ground, and reduce foot vertical touchdown velocity.

Kinematic data indicates that following sacral biofeedback participants also attempted to reduce loading via reduced COM oscillation (66-100% of cycle)(partial eta squared= 0.75), however the strategies employed to do so appear to be different from those employed following tibial biofeedback. As explained, reduced maximum COM height will result in a reduced fall height, ultimately reducing vertical touchdown velocity and impact loading (McMahon et al, 1989). This reduced COM oscillation appears to be a result of a change to kinematics in mid-to-late stance, consequently reducing the magnitude of vertical propulsion. Participants displayed a more dorsi-flexed foot (23-41% of cycle) (partial eta squared =1.49), more flexed knee (20-40% of cycle)(partial eta squared=1.19), and more flexed hip (17-40% of cycle)(partial eta squared= 1.28) in mid-to-late stance/early swing. This is indicative of a reduced vertical ground reaction force active peak, due to a decreased range of motion over which the force is applied manipulating the impulse momentum relationship. Participants also appeared to adopt a more compliant landing strategy as evident by increased hip flexion prior to and at initial contact (80-100% of cycle)(partial eta squared= 1.26) and a decreased COM height during stance (1-30% of cycle)(partial eta squared= 0.58). Although the increased hip flexion is not maintained during early stance/impact the reduced COM height is indicative of reduced vertical stiffness, and consequently reduced impact loading (Lieberman et al., 2010), as evident by reduced tibial (-1%) and sacral accelerations (-27%).

Participants responding to treadmill biofeedback displayed increased ankle dorsiflexion (21-34% of cycle) (partial eta squared= 1.29), knee flexion (18-29%) (partial eta squared=0.96), and hip flexion (6-31% of cycle)(partial eta squared=0.81), during mid-to-late stance (similar to sacral biofeedback). This is indicative of reduced vertical ground reaction force active peak, and thus reduced COM oscillation (as explained previously). However, these kinematic changes appeared not to alter COM height at any point during the cycle. This is somewhat surprising considering the similarity in kinematics to those displayed by the sacral biofeedback group, in conjunction with the found reductions in both tibial (-26%) and sacral (-17%) accelerations. Despite this, COM was trending towards significantly lower values (partial eta squared= 0.76) during late stance/early swing phase (33-41% of cycle). This insignificant difference may be due to large variations in response to treadmill biofeedback, potentially associated with an oscillating treadmill that does not match the natural frequency of the runner.

In summary, the kinematic strategies employed for each biofeedback location appear to indicate that peak sacral and tibial accelerations are mainly reduced via alterations to COM oscillation as opposed to increasing cushioning at impact (as in study 1). However, this does not explain why each group demonstrated different abilities to reduce peak accelerations.

Tibial biofeedback appeared to be the only group that did not make alterations to knee or hip kinematics during stance. This may have two possible explanations. Firstly, it is possible that tibial accelerations may be most easily reduced via a reduced vertical touchdown velocity of the foot, which can be achieved by maintaining the foot closer to the ground throughout swing (as evident by the increased knee extension in early swing). However, this appears not to facilitate large reductions in sacral accelerations (-6%). Secondly, during tibial biofeedback, initial experimentation with increasing knee or hip flexion may have occurred during impact and subsequently displayed larger tibial accelerations values due to a reduction in effective mass (Derrick 2004, Potthast et al., 2010, Denoth, 1986). As a result, participants may have avoided adjustments to hip and knee kinematics during stance. This may therefore explain the relatively larger sacral reductions for sacral and treadmill based biofeedback. Furthermore, treadmill biofeedback may have presented large reductions to both tibial (-26%) and sacral (-17%) acceleration values as it displayed similar knee and hip kinematics as the sacral biofeedback group during stance, and trended towards a more extended knee during the early-to-mid swing phase (51-61%) (Effect size =0.97), thereby displaying aspects of strategies employed by both the sacral and tibial biofeedback locations.

4.5 Conclusion

Given the simplicity and relative cheapness of the biofeedback system used in this study there is huge potential for numerous applications in pre-habilitation and rehabilitation of running related injuries. Treadmill biofeedback appears to provide the largest whole body decreases in impact loading; however there may be a case for prescribing different forms of biofeedback (by varying anatomical location) depending on where an individual may be most predisposed to injury. Moving forward, examination of the implementation of such a feedback system to impart motor learning and permanently alter mechanics to reduce risk of injury development, may have implications within clinical populations (elderly, osteoarthritic, obese), facilitate exercise in patients at increased risk of bone injury, and reduce the risk of running related injuries in general populations. However, longer intervention periods must first be examined in healthy populations to determine if long-term motor learning can be attained through this biofeedback based gait re-training system. Finally, kinematic strategies to reduce sacral accelerations appear to be largely driven by increased knee and hip flexion in the second half of stance, whereas maintaining a lower foot position during swing may most effectively reduce tibial accelerations.

4.6 Limitations

- Kinematic data is limited due to a lab incident resulting in loss of data. Only variables that were already extracted are reported.
- Participants within this study were healthy and may therefore not demonstrate the same response to biofeedback as at risk or injured participants.
- Convenience based sampling was employed for recruitment of participants and therefore limits generalizability to wider populations.
- Large variability in response to feedback may limit generalizability of results to wider populations.

4.7 Future recommendations

- A direct comparison between verbal feedback and biofeedback should be examined.
- In order to provide improved ecological validity it may be beneficial to examine the use of both verbal feedback and biofeedback in conjunction with simple written technical instruction. This stems from the idea that if a biofeedback system was sold as a product it is likely to be accompanied by instructions.
- A longer-term intervention should be examined, as acute changes to kinetics and kinematics may not reflect the chronic effect of such biofeedback system.

Chapter 5: Study 3
A comparison of verbal feedback, visual accelerometer-based biofeedback, and simple written instruction on reducing impact loading during running

5.1 Introduction

It is clear from work by Crowell et al (2010; 2011), and study 2 of this research, that humans have the ability to respond to visual biofeedback from an accelerometer, and make necessary kinematic changes to reduce the magnitude of impact accelerations at both the tibia and sacrum. However, verbal instruction alone has also been effectively used to alter both running mechanics (Williams, McClay and Manal, 2000; Fletcher, Bertlett, Ramanov and Fotouhi, 2008) and jump landing mechanics (McNair, Prapaveiss and Colleagues, 1999; 2000; 2003). The advantage of verbal instruction is that it does not require the use of technology and there is currently no commercially available system. The disadvantage of verbal instruction is that the instructor needs to be able to observe the unwanted movement pattern and repeated access to an instructor would be very expensive.

It is necessary to investigate the ability of verbal feedback in comparison to the biofeedback system used in study 2, with regard to its ability to alter mechanics to facilitate reduced magnitude of impact accelerations and subsequently maintain reductions one the feedback mechanism has been removed. This has not previously been investigated. Furthermore, study 2 indicated that the provision of tibial, sacral, and treadmill biofeedback could effectively alter mechanics to reduce the magnitude of impact accelerations as a result of self-directed change. It is reasonable to consider that prior to the use of a biofeedback system or visiting a 'running re-trainer' the user/runner would access written technical instructions. Such instructions may further improve the ability to alter running mechanics during feedback based gait re-training. However, this has not been previously examined.

Aims

- To compare the success of verbal instruction and visual biofeedback in combinations with a simple information pack detailing compliant running strategies, on their ability to alter running mechanics and decrease impact accelerations at the tibia and sacrum.
- To examine the kinematic changes made by participants to facilitate reduction of impact acceleration magnitudes, with the goal of discovering successful kinematic strategies for impact acceleration reduction.

5.2 Methodology

5.2.1Experimental design

This study utilized a three group randomized pre-test post-test experimental design to investigate the effects of verbal feedback, visual biofeedback, and written technical instructions, on the magnitude of impact accelerations and running kinematics. Participants were recruited from a university population and local sports clubs, and were required to attend the Biomechanics lab in the School of Health and Human Performance (Dublin City University (DCU)) on two occasions. Convenience based sampling was employed for recruitment of participants and therefore limits generalizability to wider populations. All participants completed a physical activity readiness questionnaire (PARQ) (ACSM, 1997) and informed consent form prior to participation. This study was granted ethical approval by DCU's ethics committee, as required.

5.2.2 Participants

All participants were male, between the ages of 18-46, and were involved in running related activities for at least 30 minutes a week for six months prior to commencement of this study. Participants were excluded if they had any injuries (defined as any physical impairment affecting running distance, speed, duration or frequency), cardiovascular disease, or neurological disorders that may affect their gait. Participants were also excluded if they had lower limb surgery (Devita, Hunter & Skelly, 1992). Volunteers from study 1 and 2 were excluded from participating in order to prevent the influence of learning and practice effects. A total of 30 participants (18-46, height: 178.8± 7.2, mass: 79.2 ±8.3) were randomized into three groups of 10 participants (biofeedback group, verbal feedback group, and no feedback group).

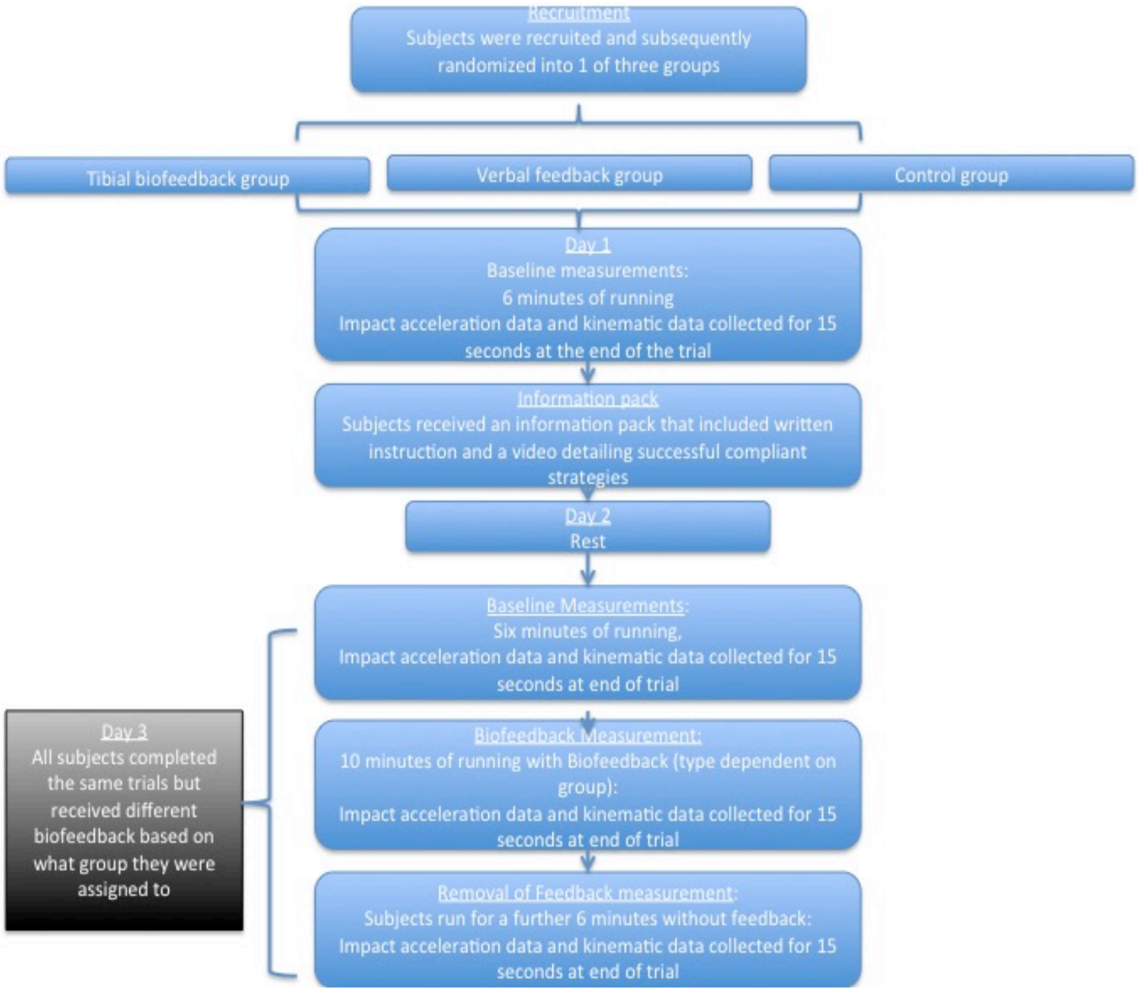

Figure 5.2. 1: Experimental procedure.

5.2.3Experimental Procedure

The experimental procedure (Figure 4.2.1) involved two testing days separated by one rest day. Day one involved baseline measures and day 2 involved the main experimental measures. Prior to commencement of each testing day participants were marked with pen at 12 predetermined anatomical positions (Figures 4.2.2, 4.2.3 and table 4.2.1), retrofelective markers were subsequently attached using double side tape and 1.5 inch PowerFlex™ self adherent tape (Andover®, PowerFlex™). Following this, 2 uniaxial capacitive accelerometers (Advanced sensors calibration 4421-030, frequency range 1g-200g), mounted to pieces of balsa wood were attached to the proximal anteriomedial aspect of the tibia adjacent to the tibial tuberosity and on the sacrum, horizontally in line with both posterior superior iliac spines. The sensors were attached so that the active axis of each accelerometer was aligned along the axis of the tibia, and along the midline of the spine respectively (refer to section 3.2.3 for further attachment details).

For all trials, participants ran at a standardized speed of 2.70 m/s (10km/hr) at a 1% incline (as explained in section 4.2.3 (study 2)).

For the duration of all trials participants wore their own standard running shoe. In order to ensure this, a photo of each person's footwear, worn during baseline measures, was photographed, recorded, and then checked before commencement of day two of testing. This was done in order to improve test-to-test reliability of data. This is important as footwear has been shown to influence both kinematics (Hardin, van den Bogert, Antonie J & Hamill 2004) and magnitude of impact acceleration peaks (McNair, Marshall 1994). However, shoes were not standardized across all participants (as explained in section 4.2.3).

4.2.4 Day 1: Baseline testing

For initial baseline testing and familiarization each participant ran for six minutes at 2.70 m/s (10km/hr) at 1% incline. It has been established that six minutes of treadmill running is required in order to observe stable, kinematics (Lavcanska, Taylor & Schache 2005) and impact kinetics (White, Gilchrist & Christina 2002). Once six minutes of running was complete, 15 seconds of kinematic and impact acceleration data were collected. Participants were unaware of data collection windows. Following completion of this bout of treadmill running each participant received an instruction form that detailed what was required of them between day 1 and day 2 of testing. Each participant was asked to wear the same shoes for day 2, to follow the same hydration and nutrition plan as before baseline testing, and not to engage in any vigorous physical activity until completion of day 2 of testing. This was done in an effort to improve reliability of measures.

Following day 1 each participant received an information pack that contained both written and videotaped instructions on how to run with a more compliant style. Participants were asked to familiarize themselves with this information but not to practice the running style. This was to avoid participants completing varying amounts of practice prior to day 2 testing. This is an important consideration as varying amounts of practice have been show to effect motor performance and skill development(Salmoni, Schmidt & Walter 1984).

4.2.5 Day 3: Experimental measures

The second testing session was carried out at the same time of the day as the familiarization and baseline testing, but 48hours later. Before data collection began, each participant completed a standardized questionnaire that confirmed that; 1) they received the information pack, 2) that they had read the information pack, watched the video and were familiar with compliant running technique, and 3) that they had not completed any running or vigorous physical activity since completion of familiarization and baseline testing. Participants were informed that the present researchers were unable to answer any questions regarding the material presented in the information pack. This was done in order to improve ecological validity, with the idea that compliant running information may be available online.

Before testing commenced each participant was informed of the details of the second testing session and what kind of feedback they would receive (tibial biofeedback, verbal feedback, or no feedback). All participants were instructed to try and implement the compliant running guidelines contained within the info pack, while running. All participants, regardless of group, completed six minutes of running without any feedback. At the end of this six minutes a 15 second window of kinematic and acceleration data were collected. Upon completion ten minutes of feedback was provided to the biofeedback and verbal feedback groups, while running. For the duration of this 10minutes a screen was projected in front of the biofeedback group displaying a live feed of tibial acceleration traces (as explained in section 4.2.4)(figure 4.2.6, and 5.2.2). During this ten minute bout the verbal feedback group received standardized verbal instructions regarding their adherence to the compliance running style outlined in the info pack. An expert on this running style viewed each participant, from a side on view, distally to proximally. Participants were informed that if no verbal instructions were given, to assume they were correctly adhering to the compliant style portrayed in the info pack. Participants in the verbal group were told when the ten minutes of feedback started and ended. After receiving ten minutes of feedback, both feedback groups completed a further six minutes running, with no feedback. Participants were instructed to try and maintain the running style that was most effective during the ten minutes of feedback. Participants in the control group ran for a total of 22 minutes (same as the two feedback groups), receiving no further instructions or feedback, using only the information received in the information pack to attempt to run more compliantly. Running was continuous for each group without any breaks between the three different phases of the test. Data were collected in 15-

second windows after 6 minutes, 16 minutes and 22 minutes (marking the end of each phase).

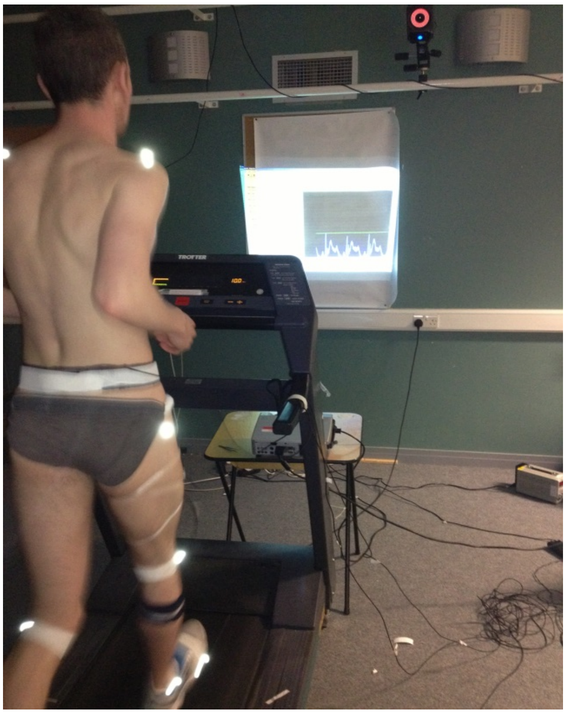

Figure 5.2. 2: Experimental set-up for Biofeedback group

5.2.6 Data Collection and processing

Throughout the testing procedure lower extremity kinematics and impact accelerations were measured using a 12-camera 200Hz Vicon motion analysis system (Vicon Oxford Metrics Ltd, Oxford, United kingdom) and two uniaxial capacitive accelerometers (Advanced sensors calibration 4421-030, frequency range 1g-200g), respectively. For the duration of trials, the treadmill was placed in a standardized location and orientation, making contact with a force plate (1200 X 600 mm, 2000 Hz, Advanced Mechanical Technology Inc., Watertown, MA). This facilitated identification of initial contact for each foot strike. Twelve retrofelective markers were tracked (Figure 4.2.2 and 4.2.3, Table 4.2.1) by the 12- camera system. Two dimensional data from each camera is then combined, by Vicon workstation with calibration data to convert the equivalent digital motion information into three dimensions.

A custom MATLAB program (The MathsWorks Inc., United Kingdom) was used to calculate both lower extremity kinematics and to identify impact acceleration peaks. From each 15-second window of data collection, the custom MATLAB program (The MathsWorks Inc., United Kingdom) identified each consecutive foot strike. This data was then averaged.

5.2.7 Summary of dependent variables

For each stride the following variables were measured. Note: data were collected for 15-second windows and averaged for each variable.

- Ankle, knee, and hip angle
- Vertical height and velocity of the COM
- Peak tibial and sacral accelerations
- Heel Marker vertical velocity

5.2.8 Data analysis

In order to examine the effect of different forms of feedback across 4 time points on tibial and sacral impact accelerations two 3*4 (feedback type * time) within between repeated measures ANOVA were completed. The within participant factor was time-point (baseline, post information pack, feedback, and removed feedback) and the between factor refers to feedback type (biofeedback, verbal feedback, and no feedback). For all tests, statistical significance was set at $p<0.05$. Normality of data was assessed with using the Shapiro-Wilk test for normality. Mauchleys test was used to examine sphericity. In cases where the assumption of sphericity was violated a Greenhouse-Geisser correction was employed. Homogeneity of variance (HOV) was examined using Levene's test. Violations to HOV were deemed acceptable as all groups have the same sample size (n=10), as in accordance Stevens (1996).

In order to do complete full curve comparisons for all kinematic data an Analysis of Characterizing Phases was performed (Richter et al., 2014) as described in study 2 (4.2.7). Multiple paired sample T-tests were completed for the within participant comparisons and independent sample T-tests were completed for between participant comparisons to determine significantly different key phases. A Bonferonni adjustment was employed to reduce the likelihood of a type 1 error.

5.3 Results

5.3.1 Peak acceleration results

A significant time*feedback type interaction for peak tibial accelerations was evident ($F_{(3.451, 46.591)}=3.782$, $p<0.05$, partial eta squared= 0.219) but not for peak sacral accelerations ($F_{(2.813, 37.973)}=0.660$, $p=0.572$, partial eta squared= 0.047).

Subsequent post-hoc repeated measure ANOVA's revealed that only the biofeedback group displayed a significant effect of time ($F_{(3,2)}=8.792$, $p<0.05$, partial eta squared=0.494). Pairwise comparisons revealed that baseline measures were significantly smaller than post information pack, biofeedback, and removed feedback measures ($p<0.05$) (Percentage differences can be observed in table 5.3.1.).

At the sacrum there was a significant main effect for time ($F_{(1.406, 37.973)}=4.174$, $p<0.05$, partial eta squared=0.134) but not feedback group ($F_{(2,27)}=1.186$, $p=0.171$, partial eta squared=0.123). Pairwise comparisons revealed that baseline measurements were significantly larger than post information pack measures (19%, $p<0.05$), biofeedback measures (17%, $p<0.05$), and removed feedback measures (15%,$p<0.05$), for all feedback types.

Subsequent power analysis indicates that for the determination of an interaction effect for tibial accelerations there was an observed power of 0.819, and a within participant observed power of 0.874. Indicating that there was high power with regard to tibial accelerations.

For sacral accelerations there was an observed power of 0.172 with regard to discovering an interaction effect, and 0.838 for within participant effects. This indicates that there may not have been sufficient power to indicate an interaction effect, however there was sufficient power to determine within participant differences, at the sacrum.

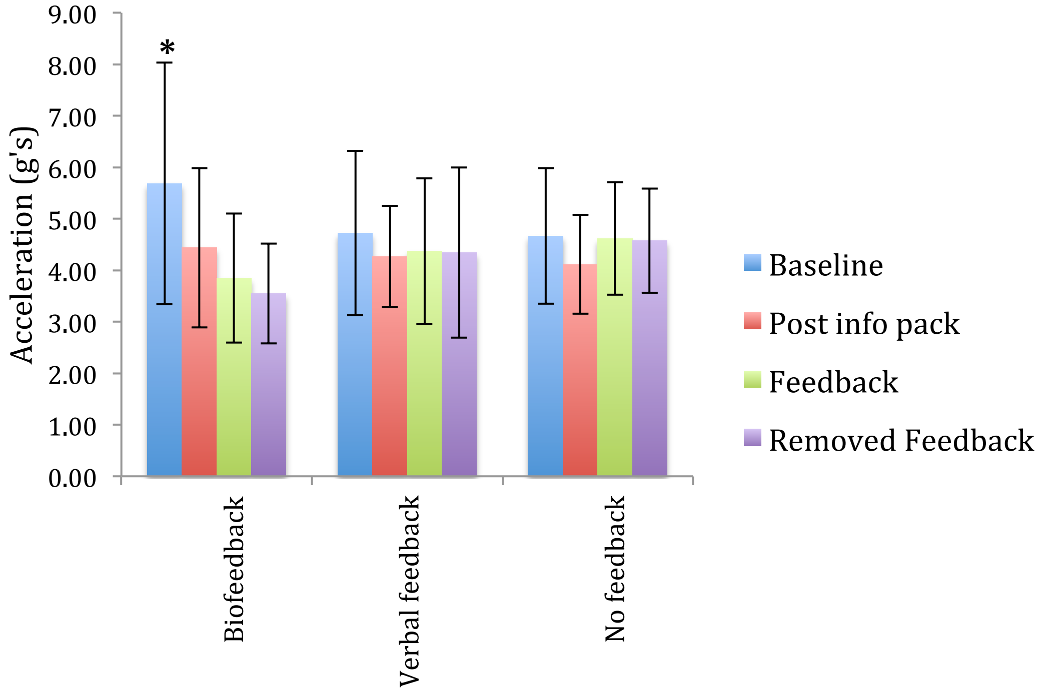

Figure 5.3. 1: The effect of different forms of feedback on peak tibial acceleration (Mean + Standard deviation) (* indicates a significant difference between baseline and all conditions within the same group)

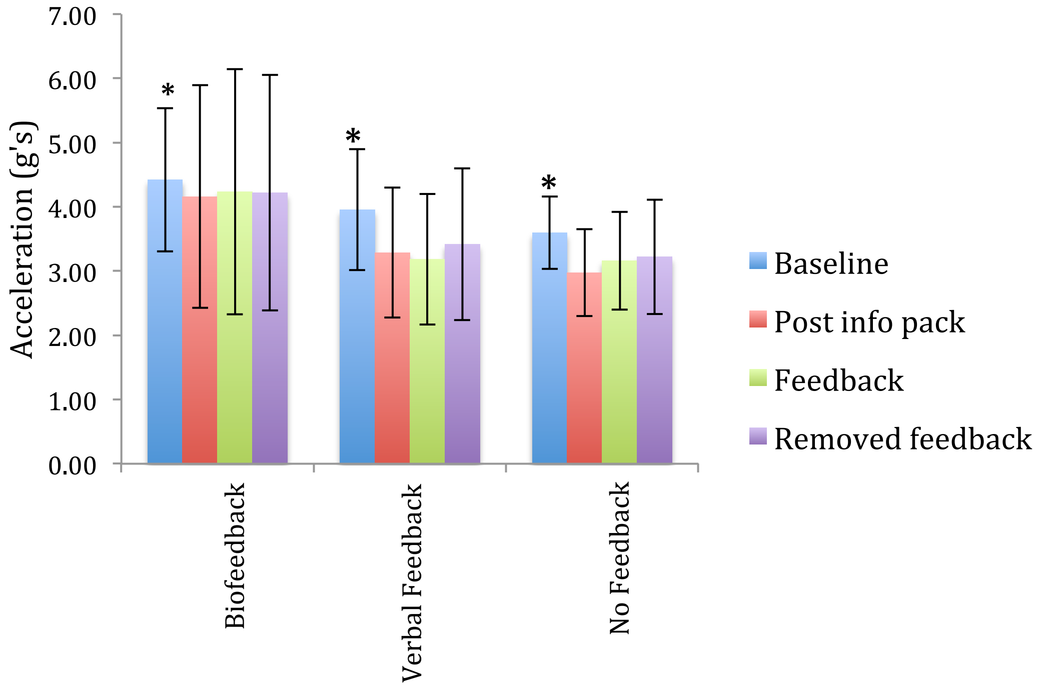

Figure 5.3. 2: The effect of different forms of feedback on peak sacral acceleration (Mean + Standard deviation) (* indicates a significant difference between baseline and all conditions within the same group)

Table 5.3. 1: Percentage difference between conditions within each group (* indicates a significant difference, p<0.05)

Group	Location	Baseline Vs., Post Info pack	Baseline Vs. feedback (% Diff.)	Baseline Vs. Removed feedback	Feedback Vs. Removed feedback	Post Info pack Vs. Feedback	Post Info pack Vs. Removed Feedback
Biofeedback	Tibia	25%*	39%*	46%*	8%	14%	22%
	Sacrum	6%*	4%*	5%*	0%	-2%	-1%
Verbal Feedback	Tibia	10%	8%	8%	-1%	-2%	-2%
	Sacrum	18%*	22%*	15%*	-7%	3%	-4%
No Feedback	Tibia	13%	1%	2%	1%	-11%	-11%
	Sacrum	19%*	13%*	11%*	-2%	-6%	-8%

5.3.2 Kinematics

5.3.2.1 Tibial biofeedback group

The biofeedback group displayed a significantly larger degree of dorsi flexion at post information-pack, relative to baseline measures, for 6-16% of the cycle (T=2.51,p<0.01, partial eta squared=0.62) and 20-42% of the cycle (T=4.53, p<0.01, partial eta squared= 1.44). Following tibial biofeedback participants displayed a significantly more dorsiflexed ankle for 18-41% of the cycle (T=6.15, p<0.001, partial eta squared=1.46), relative to baseline. This pattern continued following removal of biofeedback portrayed by a significantly greater degree of dorsiflexion for 19-38% of the cycle (T=5.27,p<0.01, partial eta squared=1.29), relative to baseline.

The tibial biofeedback group displayed a significantly larger degree of knee flexion for 19-35% of the cycle (T=3.23, p<0.01, partial eta squared=1.31) and a more extended knee for 53-70% of cycle (T=2.86,p<0.01,partial eta squared) at post information pack measures, relative to baseline. Following tibial biofeedback participants displayed a larger degree of knee flexion for 15-36% of the cycle (T=5.37,p<0.001, partial eta squared= 1.40) and a more extended knee for 82-90% of the cycle (T=2.66, p<0.01, partial eta squared=0.88), relative to baseline. Following removal of biofeedback, participants displayed a more flexed knee during 16-38% of the cycle (T= 4.14,p<0.05, partial eta squared= 1.37), relative to baseline. Following the information pack participants displayed a more extended knee for 63-76% of the cycle (T=2.74,p<0.05, partial eta squared=0.44) in comparison to removal of biofeedback. Furthermore, following tibial biofeedback participants displayed a more extended knee for 63-72% of the cycle (T=2.60,p<0.05, partial eta squared= 0.22) in comparison to removal of biofeedback.

The tibial biofeedback group displayed a significantly larger degree of hip flexion following the information pack, relative to baseline, for 17-40% of the cycle (T=3.85,p<0.01, partial eta squared= 1.12) and 79-96% of the cycle (T=2.69, p<0.01, partial eta squared= 1.08). Similarly, following biofeedback participants displayed a significantly larger amount of hip flexion during 18-39% of the cycle (T=3.90,p<0.01, partial eta squared= 1.25), 76-93% of the cycle (T=2.56, p<0.05, partial eta squared=1.15), and 39-45% of the cycle ($T=2.13$, $p<0.01$, partial eta squared=0.75). This pattern continued to no feedback measurements with participants displaying a larger degree of hip flexion (relative to baseline) for 13-35% of the cycle ($T=6.3$,$p<0.001$, partial eta squared= 1.26) and 78-100% of the cycle ($T=3.13$, $p<0.01$, partial eta squared= 1.13).

The tibial biofeedback group displayed a significantly lower COM (relative to baseline) for 20-55% of the cycle ($T=4.24$,$p<0.01$, partial eta squared=0.76) following the information pack, for 20-52% of the cycle ($T=4.28$,$p<0.01$, partial eta squared=0.74) following biofeedback, and for 69-90% of the cycle ($T=3.18$,$p<0.01$, partial eta squared=0.62) after the removal of biofeedback. Following the information pack information participants displayed a significantly lower COM for 30-100% of the cycle ($T=3.56$,$p<0.01$, partial eta squared=0.28) in comparison to removal of biofeedback. Similarly, participants displayed a significantly lower COM height during biofeedback than removal of biofeedback for 23-100% of the cycle ($T=3.4$,$p<0.01$, partial eta squared=0.21).

The tibial biofeedback group displayed a significantly slower COM velocity following the information pack, relative to baseline, for 12- 32% of the cycle ($T=5.92$, $p<0.001$, partial eta squared= 1.39), 9-11% of the cycle ($T=2.4$, $p<0.01$, partial eta squared= 0.33), and 58-79% of the cycle ($T=5.42$, $p<0.01$, partial eta squared= 1.26). Similarly, participants displayed a slower COM velocity following biofeedback (relative to baseline) for 62-79% of the cycle ($T= 6.12$,$p<0.001$, partial eta squared=1.35), 1-6% of the cycle ($T= 4.39$, $p<0.01$, partial eta squared= 0.89), and 82-91% of the cycle ($T=5.55$, $p<0.001$, partial eta squared=1.44). This pattern continued to no feedback measures, with participants displaying a significantly slower COM velocity relative to baseline for 60-78% of the cycle ($T=5.13$, $p<0.01$, partial eta squared= 1.39), 18-32% of the cycle ($T=4.46$,$p<0.01$, partial eta squared= 0.90), and 1-4% of the cycle ($T=2.92$, $p<0.01$, partial eta squared= 0.82). Following the information pack participants displayed a significantly slower COM velocity at 4-7% of the cycle ($T=2.61$,$p<0.01$, partial eta squared=0.74) relative to after biofeedback. Following removal of biofeedback, COM velocity was significantly faster for 1-7% of the cycle ($T=4.42$,$p<0.01$, partial eta squared=0.73) and 72-79% of the cycle ($T=4.6$,$p<0.01$, partial eta squared=0.46) when compared to information pack measures.

The tibial biofeedback group displayed a significantly slower heel velocity, relative to baseline, following the information pack for 16-36% of the cycle (T=5.85, p<0.001, partial eta squared=1.32) and 68-87% of the cycle (T=3.53, p<0.01, partial eta squared=1.15), following biofeedback for 16-36% of the cycle (T=10.18,p<0.001, partial eta squared=1.30) and 62-83% of the cycle (T=3.59, p<0.01, partial eta squared= 1.07), and following removal of biofeedback for 61-74% of the cycle (T=3.85,p<0.05, partial eta squared=0.96) and 15-37% of the cycle (T=8.96, p<0.001, partial eta squared= 1.15). Participants displayed a significantly slower heel velocity following the information pack during 76-87% of the cycle (T=3.34,p<0.01, partial eta squared= 0.56), relative to removal of biofeedback. Participants also displayed a slower heel velocity following biofeedback for 75-87% of the cycle (T=3.82,p<0.01, partial eta squared=0.44) when compared to removal of biofeedback.

5.3.2.2 Verbal feedback group

The verbal feedback group displayed a significantly larger degree of dorsiflexion following feedback, relative to baseline, for 23-42% of the cycle (T=5.09,p<0.01, partial eta squared= 1.65) and following removal of feedback for 23-45% of the cycle (T=9.62, p<0.001, partial eta squared= 1.83).

Participants demonstrated a larger degree of knee flexion, relative to baseline, following the information pack for 19-33% of the cycle (T=2.89,p<0.01, partial eta squared=1.22), following feedback for 83-100% of the cycle (T=4.6, p<0.01, partial eta squared= 1.49) and 15-51% of the cycle (T=8.40, p<0.01, partial eta squared= 1.54), and following removal of feedback for 85-100% of the cycle (T=4.02,p<0.01, partial eta squared= 1.52) and 18-44% of the cycle (T=6.94,p<0.01, partial eta squared=1.41). Furthermore, participants displayed a larger degree of knee flexion following feedback for 45-55% of the cycle (T=3.96,p<0.01, partial eta squared=1.04) relative to after the information pack. This pattern continued with a larger degree of knee flexion displayed removal of feedback (39-57% of the cycle)(T=4.40, p<0.01, partial eta squared= 1.07), relative to after the information pack.

Participants displayed a larger degree of hip flexion, relative to baseline, following the information pack for 23-45% of the cycle (T=3.34,p<0.01, partial eta squared=1.07), following feedback for 1-41% of cycle (T=8.98,p<0.001, partial eta squared=1.47) and 73-100% of the cycle (T=6.80,p<0.01, partial eta squared=1.46), and following removal of feedback for 17-42% of the cycle (T=5.72,p<0.01, partial eta squared=1.39) and 74-100% of the cycle (T=3.82,p<0.01, partial eta squared=1.41). Furthermore, comparisons relative to post information pack, indicates that participants displayed a larger degree of hip flexion following feedback for 1-25% of the cycle (T=4.13, p<0.01, partial eta squared=1.42) and 46-90% of the cycle (T=4.86, p<0.01, partial eta squared= 1.42), and following removal of feedback for 20-44% of the cycle (T=3.85, p<0.01, partial eta squared=0.97) and 60-81% of the cycle (T=4.40,p<0.01, partial eta squared=1.07). Finally, hip flexion was also larger following feedback for 2-7% of the cycle (T=3.22, p<0.01, partial eta squared=1.32) and 48-56% of the cycle (T=9.8, p<0.001, partial eta squared=0.50) in comparison to removal of feedback.

Participants in the verbal feedback group displayed a significantly lower COM, relative to baseline, following feedback for 20-53% of the cycle (T=8.74,p<0.001, partial eta squared=0.89), and following removal of feedback for 23-53% of the cycle (T=7.26, p<0.01, partial eta squared=0.84). Similarly, participants displayed a significantly lower COM height following feedback for 45-62% of the cycle (T=3.88,p<0.01, partial eta squared= 0.45) than following provision of the information pack.

Vertical velocity of COM was significantly slower, relative to baseline, following the information pack for 40-51% the of cycle (T=3.12,p<0.01, partial eta squared= 1.12), following feedback for 3-8% of the cycle (T=7.66,p<0.01, partial eta squared=1.12) and 40-45% of the cycle (T=3.19, p<0.01, partial eta squared=1.44), and following removal of feedback for 2-8% of cycle (T=8.69, p<0.001, partial eta squared= 1.18), 64-80% of cycle (T=4.21,p<0.01, partial eta squared=1.60), and 13-33% of the cycle (T=17.28, p<0.001, partial eta squared =1.63).

Participants in the verbal feedback group displayed a significantly slower vertical heel velocity, relative to baseline, following the information pack for 29-36% of the cycle (T=3.12, p<0.01, partial eta squared= 1.12), following feedback 20-38% of the cycle (T=5.95, p<0.01, partial eta squared= 1.7), and following removal of feedback for 19-37% of the cycle (T=9.70,p<0.01, partial eta squared=1.76).

5.3.2.3 No-feedback group

The no-feedback group displayed no significant change to ankle angle, hip angle, COM height or heel velocity, across time points.

Participants displayed a significantly more extended knee following the information pack measures for 44-54 % of the cycle ($T=3.52$, $p<0.01$, partial eta squared= 0.70), at following feedback (no feedback in this case) for 43-62% of the cycle ($T=2.78$,$p<0.01$, partial eta squared =0.97), and following removal of feedback for 44- 59% of the cycle ($T= 2.52$, $p<0.01$, partial eta squared=0.84).

Comparisons, relative to baseline, indicates that the no-feedback group displayed a significantly slower vertical velocity of COM following feedback for 83-100% of the cycle ($T=2.87$, $p<0.01$, partial eta squared= 1.01), and following removal of feedback for 91-100% of the cycle ($T=2.97$, $p<0.01$, partial eta squared=1.05).

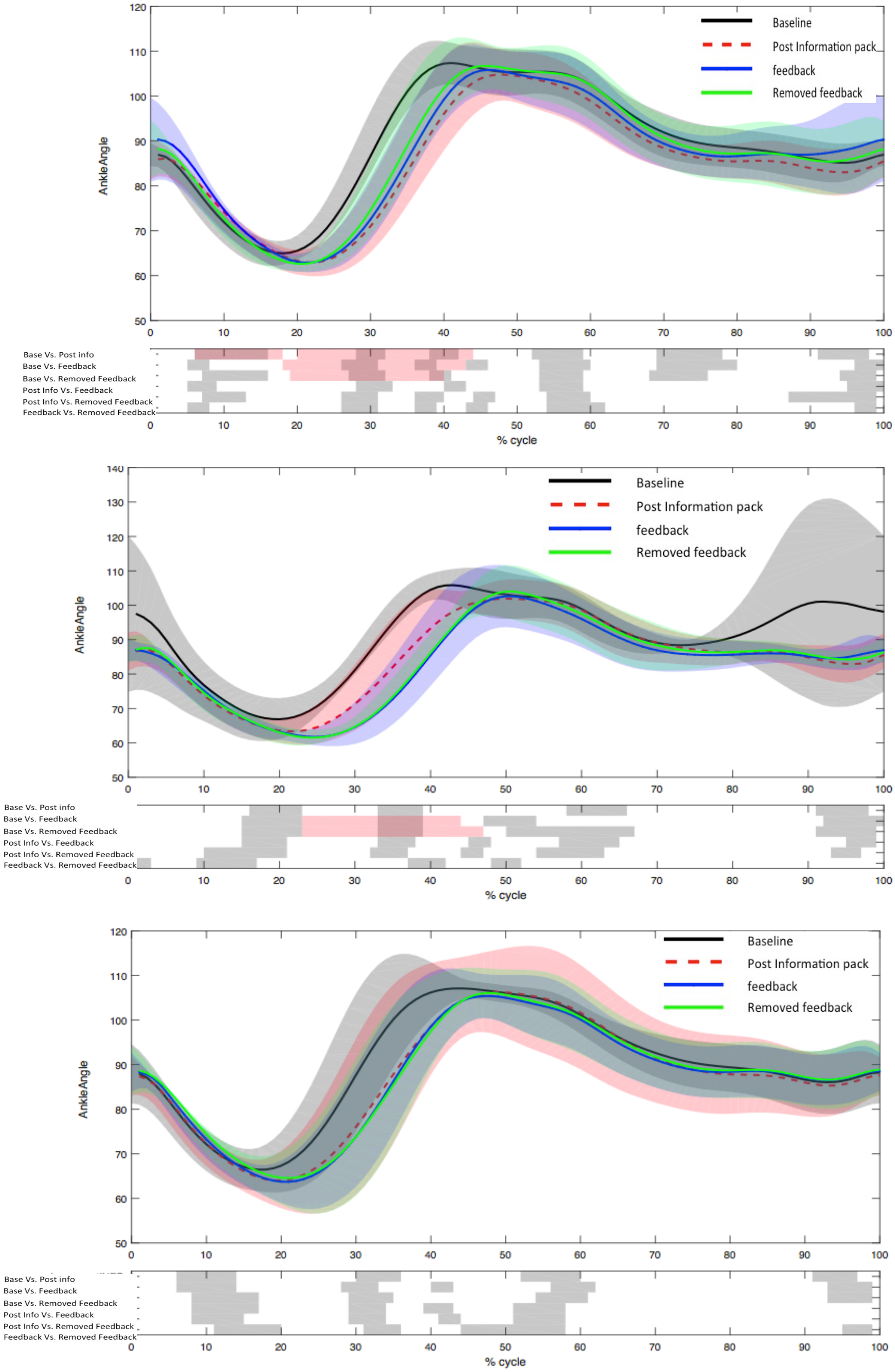

Figure 5.3. 3: The Change in ankle angle (degrees) for each group across time points: top (biofeedback group), middle (verbal feedback group), bottom (no feedback group).

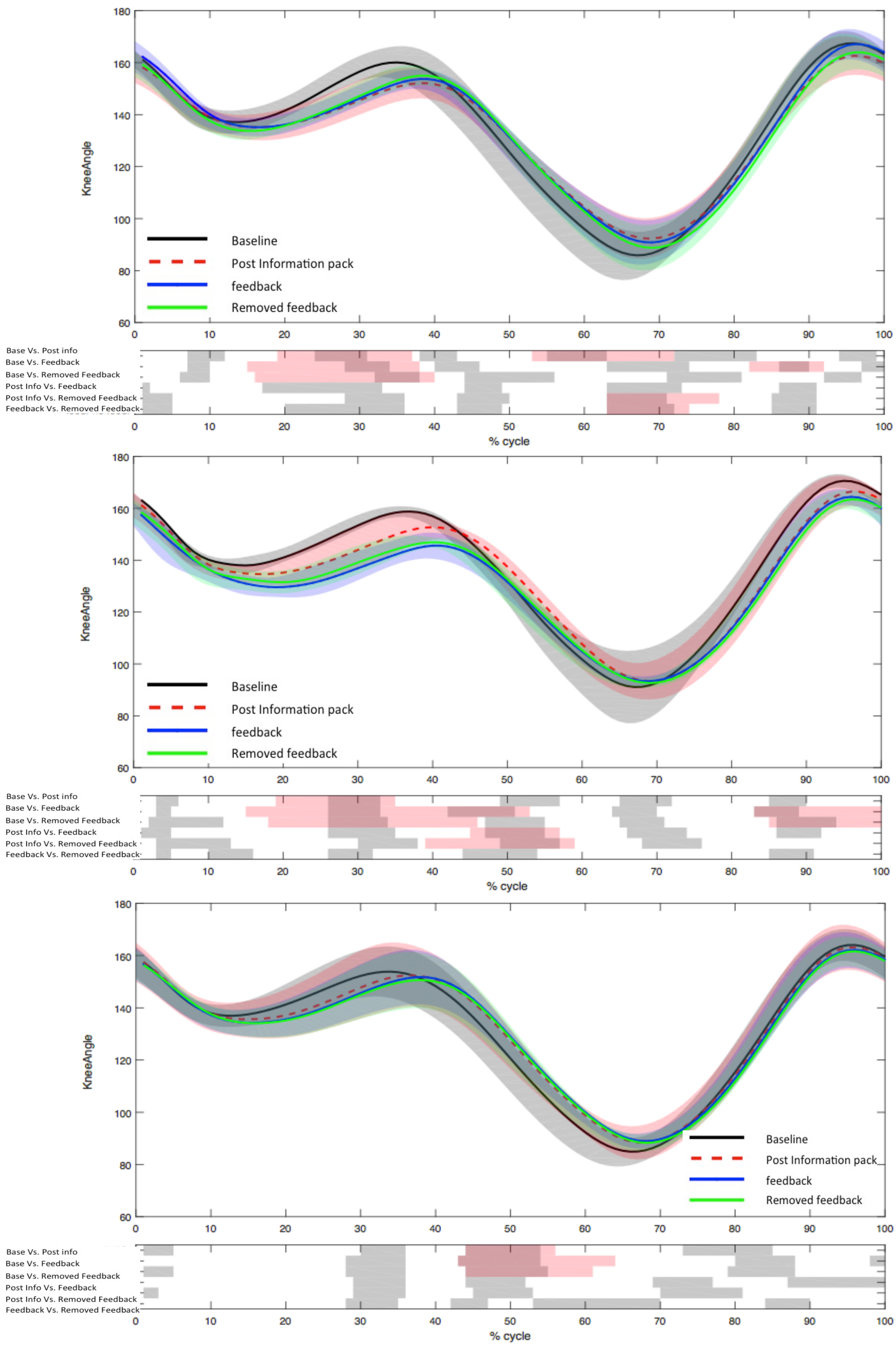

Figure 5.3. 4: The Change in knee angle (degrees) for each group across time points: top (biofeedback group), middle (verbal feedback group), bottom (no feedback group).

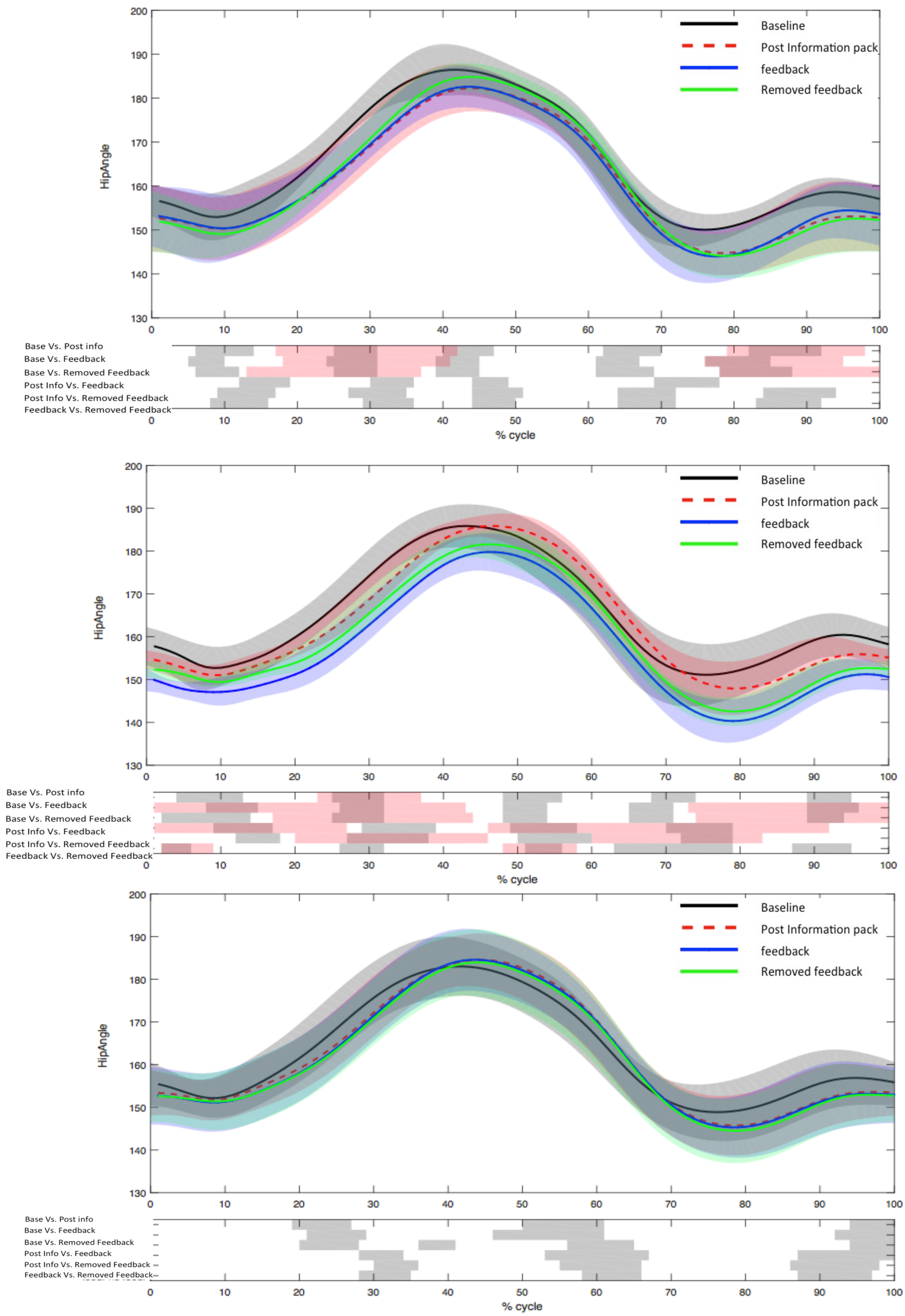

Figure 5.3. 5: The Change in hip angle (degrees) for each group across time points: top (biofeedback group), middle (verbal feedback group), bottom (no feedback group).

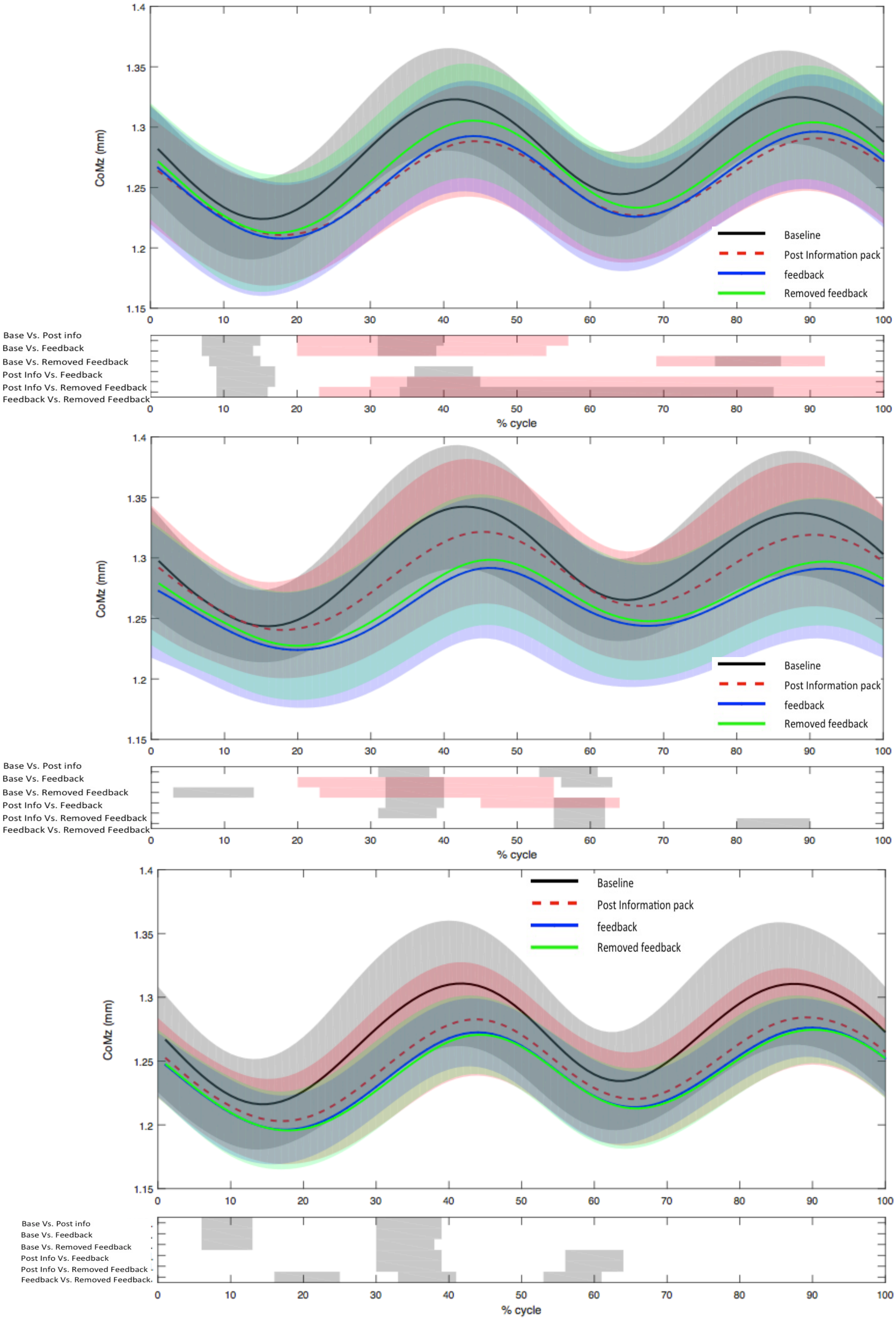

Figure 5.3. 6: The Change in COM height (metres) for each group across time points: top (biofeedback group), middle (verbal feedback group), bottom (no feedback group).

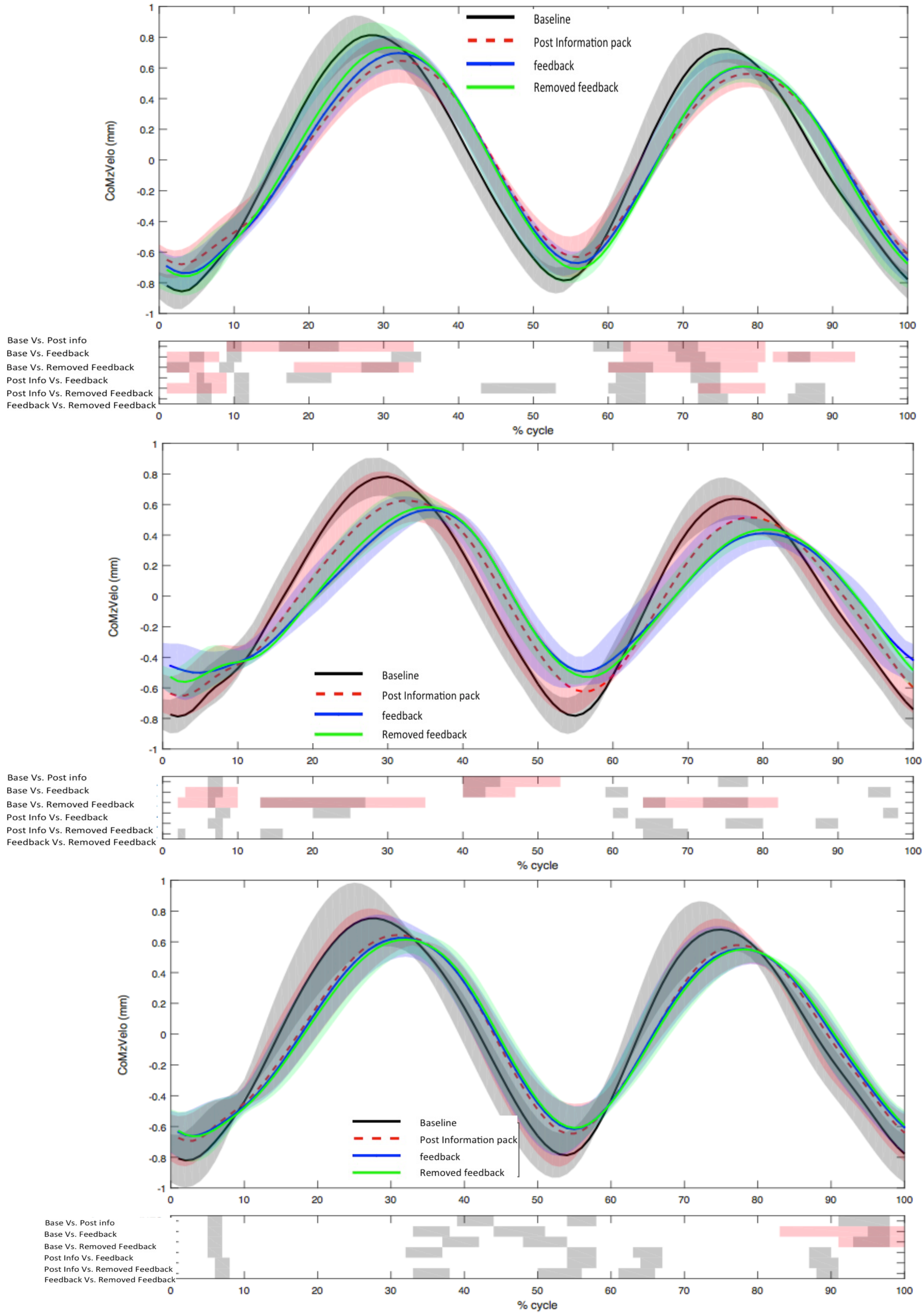

Figure 5.3. 7: The Change in COM vertical velocity (m/s) for each group across time points: top (biofeedback group), middle (verbal feedback group), bottom (no feedback group).

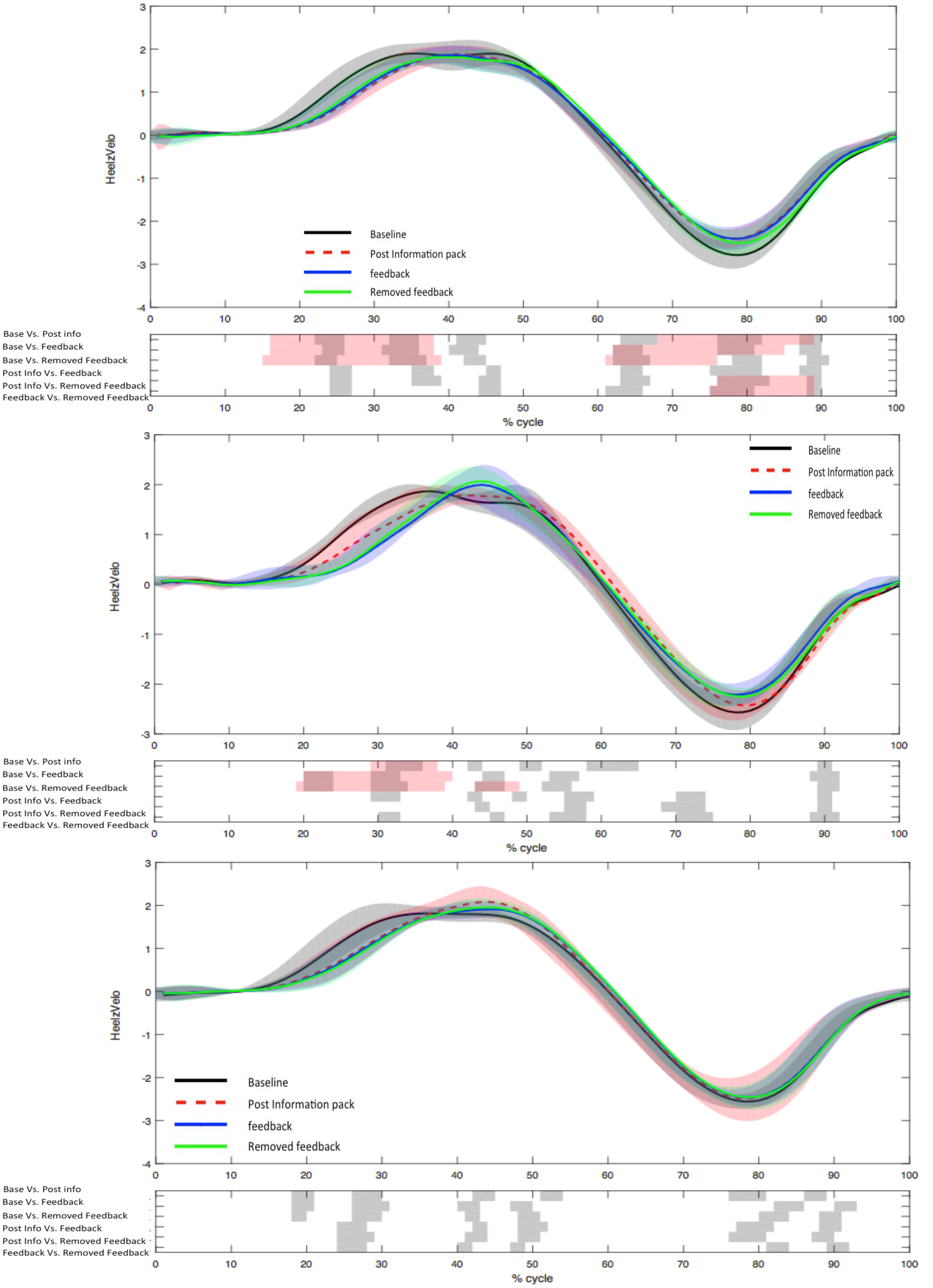

Figure 5.3. 8: The Change in heel vertical velocity (m/s) for each group across time points: top (biofeedback group), middle (verbal feedback group), bottom (no feedback group).

5.3.2.4 Group comparisons

Group comparisons revealed significant differences following feedback for ankle, knee, and hip angle but not for COM height and vertical velocity, or heel vertical velocity.

The verbal feedback group displayed a greater degree of dorsiflexion for 29-43% of the cycle ($T=2.71$, $p<0.016$, partial eta squared= 1.25) in comparison to the tibial biofeedback group.

The verbal feedback group displayed a larger degree of knee flexion than the tibial biofeedback group for 24-37% of the cycle ($T=3.58$,$p<0.016$, partial eta squared= 1.43), and the no-feedback group for 27-35% of the cycle ($T=2.23$,$p<0.016$, partial eta squared= 1.12).

Finally, the verbal feedback group displayed a significantly larger degree of hip flexion for 13-42% of the cycle ($T=2.42$, $p<0.016$, partial eta squared= 1.16) relative to the biofeedback group, and for 27-35% of the cycle ($T=2.23$, $p<0.016$, partial eta squared=1.12) when compared against the no-feedback group.

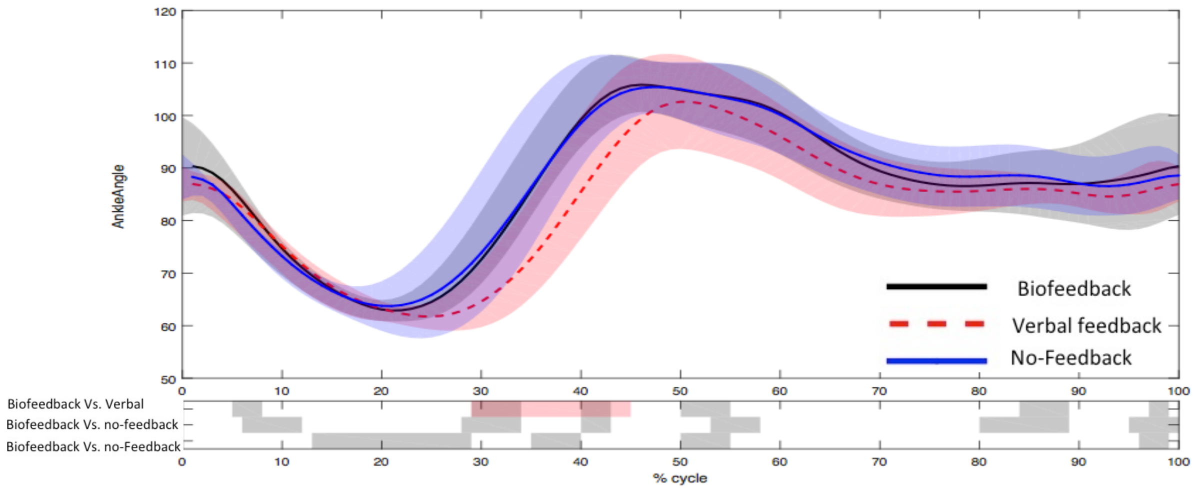

Figure 5.3. 9: Group difference at feedback measurements for ankle angle (degrees).

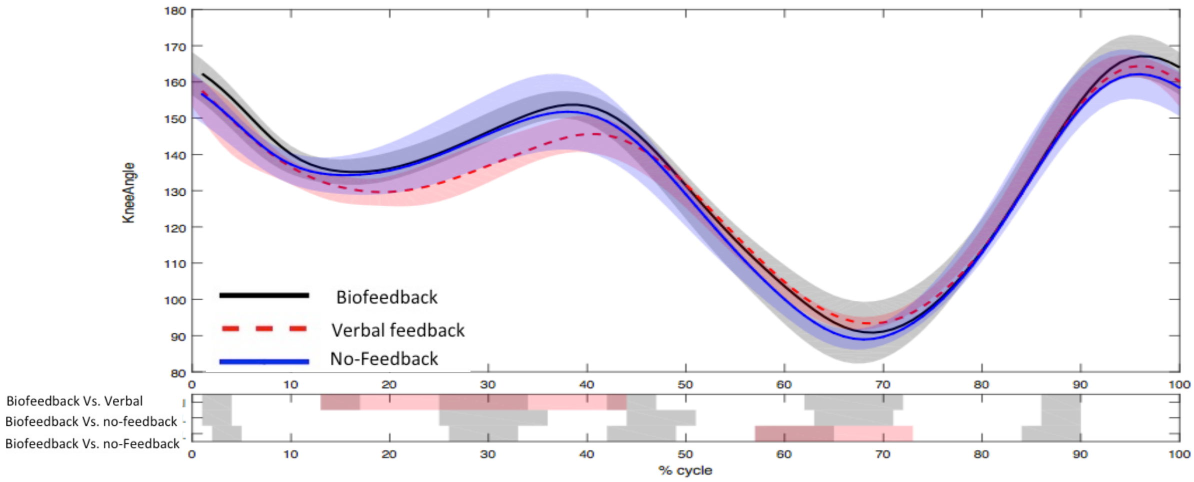

Figure 5.3. 10: Group difference at feedback measurements for knee angle (degrees).

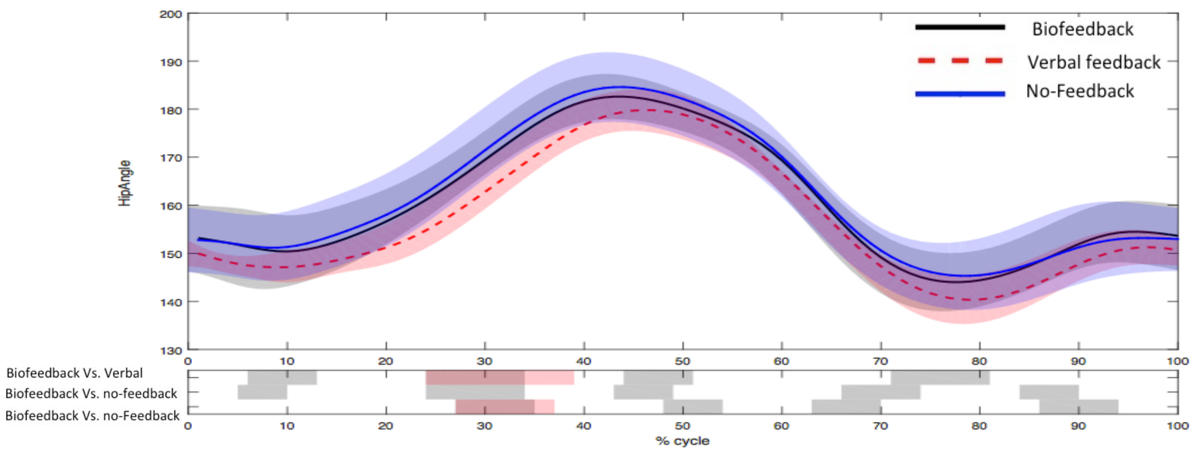

Figure 5.3. 11: Group difference at feedback measurements for hip angle (degrees).

5.4 Discussion

5.4.1 Peak accelerations

The purpose of this study was to examine the effect of verbal feedback, and biofeedback (tibial accelerometer feedback) in conjunction with simple technical instruction on their ability to alter running mechanics and reduce the magnitude of tibial and sacral peak impact accelerations.

Results indicate that provision of simple technical instruction, in the form of an information pack, significantly decreased the magnitude of peak tibial and sacral impact accelerations by 16% and 19% respectively, for all feedback types, when compared to baseline measures. Furthermore, results show that a subsequent 10-minute bout of tibial biofeedback was able to provide additional reductions in peak tibial accelerations, resulting in a decrease of 39% relative to baseline. A similar trend was observed after the removal of biofeedback as peak tibial accelerations were reduced by 46%, relative to baseline. With regard to sacral accelerations, the biofeedback and no-feedback groups failed to make further reductions after the information pack, whereas the verbal feedback group displayed a significant decrease of 22% (relative to baseline), following the feedback period.

While no previous studies appear to have examined the reduction in impact accelerations following provision of an information pack detailing compliant running strategies, the results are in agreement with previous related research. Landing studies have established that provision of simple technical instruction, guiding participants to more compliant lower limb kinematics during drop jump landing, can decrease the magnitude of the of vertical ground reaction force impact peak by 13.0%-26.7% (Prapavessis, McNair 1999, McNair, Prapavessis & Callender 2000, Prapavessis et al. 2003). Although the present study did not specifically measure vGRF, impact acceleration peaks have previously been shown to correlate highly with vGRF (Hennig, Milani & Lafortune 1993, Laughton, Davis & Hamill 2003). Similarly, it has been demonstrated that simple technical instruction can be used in interventions to successfully alter foot strike pattern, to facilitate a more compliant system(Arendse et al. 2004, Williams III, Green & Wurzinger 2012, Giandolini et al. 2013, Williams, McClay & Manal 2000). Therefore, it seems that provision of a simple information pack may be able to bring about acute changes to running style potentially decreasing the risk of injury development.

Combining an information pack with tibial biofeedback or verbal feedback has not previously been investigated, however it was envisaged that both might yield similar results. This conclusion was reached as the provision of verbal feedback was standardized and primarily based on the findings of previous research papers targeting more compliant running styles (McMahon, Valiant & Frederick 1987, Crowell et al. 2010, Crowell, Davis 2011, Cheung et al., 2011). Crowell et al (2010,2011) has shown on two occasions that significant reductions in tibial impact acceleration peaks of 26% and 48% (Crowell et al., 2010, Crowell, Davis 2011, Cheung et al., 2011) can be induced following the implementation of a tibial biofeedback system, due to decreased vertical stiffness values (28%) and heel vertical touch down velocities (5%)(Cheung et al., 2011). Also, previous work has shown that when participants are trained to run with a thigh angle of 50%, vertical stiffness was consequently reduced by 82% and vertical velocity at touch done was close to zero (McMahon et al., 1987). This information, combined with work in study 1 and 2 of this thesis formed the basis of the standardized verbal feedback provided to participants. However, verbal feedback in conjunction with an information pack yielded smaller sacral and tibial reductions than demonstrated by biofeedback alone (study 2), and smaller tibial reductions than displayed by the combination of tibial biofeedback and an information pack.

This may be due to the acute nature of the verbal feedback period in the current thesis. Studies demonstrating the successful use of verbal feedback in implementing both Chi and Pose running styles (Fletcher et al. 2008, Goss, Gross 2012, Dallam et al. 2005) utilised feedback periods of up to 12-weeks, suggesting longer intervention periods may be necessary when employing verbal feedback methods.

Another explanation for the apparent success of the biofeedback-based intervention over verbal feedback may lie in the characteristics of the type of feedback given to each group in relation to the specific attentional focuses. Research examining the adoption of a complex motor skill (overhead volleyball serve) following technical instruction (similar to this study) has revealed that participants who receive external focused feedback, which diverts the performers attention away from limb movements, produced significantly greater results than participants receiving internally focused feedback, which draws the performers attention directly towards the co-ordination of limbs (Wulf et al., 2002). As described in the methodology, the biofeedback group received visual biofeedback from an accelerometer attached to the tibia. The aim of this feedback was externally focused i.e. make the impact acceleration peaks on the screen as small as possible. On the other hand the verbal feedback group received instruction that was internally focused i.e. bend your knees, lower your hips etc; thus potentially offering an explanation for the observed differences in relation to the success of each intervention.

The successful reduction of peak tibial impact accelerations following tibial acceleration biofeedback is in agreement with previous research (Crowell et al. 2010). Crowell et al (2010) found that a 10-minute bout of tibial biofeedback followed by 10-minutes of running without any feedback reduced peak tibial impact accelerations by an average of 26%. The current study showed an overall decrease in peak tibial accelerations of 46%. Therefore, provision of the information pack prior to the biofeedback period appears to have induced larger reductions in tibial accelerations compared to those presented by Crowell et al (2010) and those presented in study 2 of this thesis (23% reduction). Further work by (Crowell et al., 2011) has shown that implementation of the same tibial biofeedback protocol across multiple practice sessions, over a two week period, can decrease peak tibial impact accelerations by an average of 48%. Numerous motor learning and feedback principles may explain the slightly larger reductions displayed by Crowell et al (2011) from multiple sessions and provide justification for examination of the current intervention across longer periods.

Firstly, the larger amount of practice completed by participants in Crowell and Davis's work (2011) has been established as an important influencing factor in motor skill development and learning(Winstein 1991, Salmoni, Schmidt & Walter 1984, Porte et al. 2007). Each participant completed 8 practice sessions, with practice time increasing incrementally from 15 minutes to 30 minutes across the two-week protocol (Crowell, Davis 2011). Secondly, a main focus of the work completed by Crowell et al (2011), was on implementing a feedback protocol that achieved motor learning of the new running style. "Motor learning" is defined as a relatively permanent change in the ability to perform a certain skill or task (Schmidt, 1988). In order to facilitate this a faded feedback design was utilized.

This meant the amount of biofeedback given to each participant was decreased as the intervention progressed. This type of feedback design has been shown to be more beneficial for motor learning and skill acquisition than constant feedback without breaks (as in the current study)(Winstein, Schmidt 1990). Following this feedback design motor learning was confirmed by a 1-month retention test. This therefore may have facilitated the greater decrease in peak tibial accelerations observed by Crowell and Davis (2011) and provides strong justification for examination of a longer intervention period with the addition of an information pack.

The purpose of comparing different forms of feedback was to determine what method could most effectively decrease peak impact acceleration loads. Reductions in peak tibial impact accelerations following provision of an information pack and tibial biofeedback (-46%) may potentially reduce local loading on the tibia and thus decrease the risk of injury. Numerous publications suggest that participants with a history of tibial stress fracture occurrence present peak tibial impact acceleration values between 13.95% and 32.53 % greater than an uninjured control population (Milner et al. 2006a, Pohl et al. 2008, Zifchock, Davis & Hamill 2006). This would indicate that following an acute bout of visual biofeedback (from a tibial accelerometer) all participants would be able to decrease tibial impact accelerations to within the limits experienced by populations with no history of tibial stress fractures. This is supported by prospective research indicating that participants who develop running related injuries and tibial stress fractures present tibial acceleration values 63% and 81% greater than uninjured controls (Davis et al., 2004, Davis et al., 2010).

Another measure that has been implicated in the development of bone injury is bone strain (Edwards et al. 2009). This occurs during running as a result of the force associated with impact. Edwards et al (2009), suggests that reductions in peak tibial impact acceleration of 23%-27% accounts for a subsequent decrease of 20%-24% in tibial bone strain. These decreases are smaller than those found in the present study thus providing further evidence to support the potential protective capacity of an acute bout of tibial biofeedback.

This suggests that this biofeedback system may be useful as a screening and injury prevention tool. Uninjured participants presenting with peak tibial acceleration values in the higher range may be prescribed biofeedback sessions that could decrease peak tibial acceleration values to within normal limits, thus decreasing the risk of a general running related injury. Furthermore, this system may be useful in the rehabilitation of running injuries.

Results indicate that tibial biofeedback and verbal biofeedback induced no additional reductions in sacral accelerations after provision of the information pack. Given that tibial impact acceleration peaks are highly correlated to forces acting on the centre of mass (vGRF and Rate of fore development) (Hennig, Milani & Lafortune 1993, Laughton, Davis & Hamill 2003), and there was a significant decrease in peak tibial impact accelerations following biofeedback, it was hypothesized that a similar decrease would be present at the sacrum. However this was not the case. This may indicate that biofeedback may only be effective at reducing acceleration values at the location it is provided from. Therefore, considering study 2, either treadmill or sacral biofeedback in conjunction with an information pack may be more appropriate for reducing sacral accelerations.

5.4.2 Kinematics

Consideration of the kinematic changes displayed for each group across time points provides an insight into the strategies employed in an attempt to reduce loading. It should be noted that kinematic data indicates that toe off occurs in the region of 30-36% of the cycle, as previously reported (Riley et al., 2008).

Examination of kinematic changes from baseline to post information pack for the tibial biofeedback group appears to indicate that the reduction in tibial (-25%) and sacral accelerations (-6%) were largely induced via alterations to mechanics in mid-to-late stance. Reduction of COM height during mid-to-late stance/early swing (which includes the maximum height of COM)(20-55% of cycle)(partial eta squared=0.76) may be explained by the increased knee and hip flexion during mid-to-late stance. This is evident from the reduce COM velocity for 12-32% (partial eta squared= 1.39) of the cycle and 58-79% (partial eta squared= 1.26) of the cycle; representing mid-to-late stance of the right and left foot contacts. This decreased velocity during propulsion results in a lower maximum COM height (as found in the present study), reducing fall height, and thus according to the laws that govern projectile motion, reduced vertical landing velocity. Increased vertical landing velocity has been previously associated with increased vertical ground reaction force impact peak (Gerritson et al., 1995, Liu and Nigg, 2000, Zadpoor et al., 2007) and thus may explain the reduced tibial and sacral acceleration values displayed in the tibial biofeedback group following the information pack. Although vertical landing velocity of the COM appears to trend towards lower values prior to contact (following information pack =0.6m/s, baseline =0.8m/s) it has not been highlighted as a key area of variance. However, a reduced heel vertical velocity prior to contact is evident (68-87% of cycle)(partial eta squared=1.15), thus reducing the magnitude of impact force via manipulation of the impulse-momentum relationship. This reduced vertical heel velocity may be somewhat explained by a more extended knee during mid-swing phase (53-70% of cycle) (partial eta squared=0.88), potentially indicating participants were attempting to keep the foot closer to the ground throughout swing, thus reducing vertical velocity of the heel prior to contact. Furthermore, participants appeared to increase hip flexion prior to initial contact (79-96% of cycle) (partial eta squared= 1.08). This may have been an attempt to maintain a lower COM, as well as increase system compliance at initial contact.

Following the information pack, the verbal feedback group appeared to make similar changes to knee and hip kinematics as the tibial biofeedback group (explained above), which resulted in a reduction in COM vertical velocity prior to contact of left foot (40-51% of cycle) and a reduced vertical heel velocity during late stance (29-36% of cycle)(partial eta squared= 1.12). Interestingly, although this resulted in an insignificant reduction in tibial acceleration values (-10%), this resulted in a larger decrease in sacral accelerations (verbal feedback=-18% Vs. visual biofeedback=-6%), relative to the biofeedback group at this time point. Examination of between group differences indicates that this may be due to increased hip flexion values for the verbal feedback group during mid-to-late stance/early swing, supporting the suggestion in study 2 that reductions in sacral accelerations may be largely driven by changes to kinematics during this period. Furthermore, the increased reduction in tibial accelerations observed for the tibial biofeedback group, between baseline and information-pack measures (relative to both the verbal feedback and no-feedback groups) may be due to the reduced vertical velocity of the heel prior to contact present in the tibial biofeedback group, but not the other two groups.

Although the no-feedback group demonstrated similar reductions in both tibial and sacral accelerations as the verbal feedback group, following the information pack, they appeared to make very few significant kinematic alterations. The only significant change appeared to be a less flexed knee during early-to-mid swing phase (44-54% of cycle)(partial eta squared= 0.70). This may be a strategy employed to keep the foot closer to the ground during swing (as explained above) in an attempt to reduce vertical velocity of the heel prior to contact, however this was not evident. This lack of significant kinematic change may be due to participants employing a variety of different strategies. However, this possibility was not explored further.

Given that each group was provided with the same information pack, it was thought that similar reductions in peak accelerations and similar kinematic strategies would be employed. However, as can be seen from above, this was not the case. This may be due to a variance in the ability of the participants within each group to alter kinematics towards those displayed in the information pack.

Examination of kinematics following the feedback period (for the biofeedback group) indicated that participants maintained similar kinematics that were implemented following the information pack, and explained above. However, following tibial biofeedback participants displayed a reduced COM velocity during the impact phase (1-6% of cycle) (partial eta squared= 0.89) compared to baseline, indicating that the COM was decelerating to a lesser extent than at baseline. This may be due to the lower COM velocity prior to initial contact (82-91% of the cycle)(partial eta squared=1.44). These further kinematics alterations, combined with a practice effect may explain the further increase in tibial acceleration reduction following the 10-minute bout of tibial biofeedback (from -25% following the information pack to -39% following tibial biofeedback). Peak tibial accelerations continued to reduce following removal of feedback (to -46% relative to baseline), despite an increase in vertical velocity of the heel during late swing relative to both post-information pack (76-87% of cycle)(partial eta squared= 0.56) and tibial biofeedback measures (75-87% of cycle)(partial eta squared=0.44). This indicates that the increase in vertical heel velocity may not have been large enough to bring about an increase in tibial accelerations. Reduction in peak sacral accelerations remained unchanged (relative to the post-information pack measurements), for the remaining time points in the tibial biofeedback group.

The verbal biofeedback group made a number of further kinematic changes following the 10-minute bout of verbal feedback that appeared to facilitate a slightly larger reduction in sacral acceleration (-22%) relative to baseline, but did not alter tibial accelerations. These changes manifested as an increase in hip flexion throughout stance (1-41% of cycle)(partial eta squared=1.47), and during late swing phase, prior to, and at initial contact (73-100% of cycle)(partial eta squared=1.46), a lower COM during mid-to-late stance/ early swing (including max COM height)(20-53% of cycle)(partial eta squared=0.89), and a reduced COM velocity during early stance/ impact phase (3-8%)(partial eta squared=1.12). Increased hip flexion throughout stance is likely responsible for the lower COM during mid-late stance. A reduced COM velocity during the propulsion phase may subsequently manipulate the impulse momentum-relationship causing a decrease in the active peak of the vertical ground reaction force, resulting in a lower maximum COM height, a reduced fall height, and thus reduced impact loading (as has been observed previously by McMahon et al., 1987). Furthermore, the increase in hip flexion during initial contact may increase system compliance, thus reducing effective mass, and reducing impact loading, reflected by reduced sacral accelerations, as has been reported previously with regard to alteration to knee angle (Derrick, 2004, Devita & Skelly, 1992).

As suspected the no-feedback group made no further kinematic changes after the post-information pack measures. This was reflected by no significant changes in peak tibial or sacral acceleration values.

Group comparisons at each time point revealed significant differences following the 10-minute feedback period but not at the other time points. Relative to the biofeedback group, the verbal feedback group had: a larger degree of ankle dorsiflexion during late stance/early swing (29-43% of cycle)(partial eta squared= 1.25), a larger degree of knee flexion during mid-to-late stance/early swing (24-37% of cycle)(partial eta squared= 1.43), and a larger degree of hip flexion during early-to-late stance/early swing (13-42% of cycle)(partial eta squared= 1.16). Thus, the kinematic strategies employed for each group appear to somewhat reflect the findings of study 2. The tibial biofeedback group appeared to avoid alteration to knee angle (as in study 2). This may be due to the association between increased knee flexion at contact and increased tibial accelerations (Lafortune et al., 1996, Potthast et al., 2010), thus initial experimentation with increased knee flexion may have resulted in increased tibial acceleration, which was visible to the tibial biofeedback group.

Finally, increased hip and knee flexion in mid-to-late stance appear be to associated with reduced sacral accelerations, whereas as reduced tibial accelerations appear to manifest as a result of reduced vertical heel velocity prior to contact; facilitated by maintaining a more extended knee during early-to-mid-swing.

5.5 Conclusion

Given the potential of this biofeedback system in conjunction with simple technical instruction to reduce impact accelerations and possibly the risk of running related injuries, this may be an extremely effective tool for sports or exercise facilities. Tibial impact acceleration peaks have been implicated in the development of running injuries both prospectively (Davis, Milner & Hamill 2004, Davis, Bowser & Mullineaux 2010) and retrospectively (Hreljac 2004, Milner et al. 2006a, Pohl et al. 2008, Zifchock, Davis & Hamill 2006) highlighting the potential benefit of this system as an injury preventative tool. Also, loading characteristics associated with foot-strike during running in injured populations have been shown to remain unchanged following recovery from injury (Seebeck et al. 2005), thus presenting a potential benefit of this system for rehabilitation purposes and prevention of re-injury. Finally, reduction of sacral accelerations appear to be driven by increased hip and knee flexion during mid-to-late stance (as observed in study 2), whereby reduction of tibial accelerations appear to be largely driven by kinematic strategies that facilitate a reduced heel vertical velocity prior to initial contact.

5.6 Limitations

- Increased baseline tibial acceleration values for the tibial biofeedback group may have influenced participant's ability to reduce peak tibial accelerations via kinematic alterations.
- Participants within this study were healthy and may therefore not demonstrate the same response to biofeedback as at risk or injured participants.
- Convenience based sampling was employed for recruitment of participants and therefore limits generalizability to wider populations.

5.7 Future recommendations

- A longer intervention period utilizing the presented biofeedback system should be examined, as acute kinetic or kinematic changes may not reflect changes as a result of chronic use of such a system.

- Motor learning should be assessed with a 1-month (or larger) retention to test.

Chapter 6: Study 4

The effect of a 4-week accelerometer-based biofeedback intervention on impact loading and energy expenditure

6.1 Introduction

In light of a current inactivity epidemic (as described in chapter 1.1) that is causing major health (World health organization, 2013) and financial problems (Chenoweth, 2005; Cadhillac et al, 2011), it is imperative that all barriers to physical activity are eliminated, or limited as much as possible. Running is one of the most popular forms of physical activity in Ireland and worldwide and has numerous health benefits (Warburton, Nicol & Bredin 2006); however large numbers (up to 70% (Hreljac, 2004)) of both competitive and recreational runners suffer running related injuries on a yearly basis, thus potentially ruling them out of physical activity for large periods of time.

Visual accelerometer based biofeedback has been shown to be successful for impact acceleration reduction acutely by up to 60% at the tibia (study 2 and 3, and (Crowell et al. 2010)) and following a 2 week gait retraining intervention using tibial accelerometer feedback (Crowell et al., 2012, Cheung et al., 2011). However the effect that a longer (4-weeks) gait-retraining program has on kinematics, kinetics or energy expenditure has yet to be examined. This study aims to examine the effect of a 4-week gait retraining intervention using treadmill accelerometer biofeedback (as this proved most effective in study 2) and a faded feedback design on tibial and sacral impact acceleration values, kinetics, kinematics, and energy expenditure.

Aim of study 4:

- To examine the effect of a 4-week visual biofeedback gait-retraining programme on
    - Impact acceleration values.
    - Kinetics and kinematics
    - Cost of locomotion and Cost of Transport

- To examine the effect of speed on impact accelerations and energy expenditure, in normal and compliant running.

6.2 Methodologies

This study implemented a pre-post repeated measures study design whereby the effect of a 4-week biofeedback based intervention on kinematics, kinetics, and energy expenditure variables was examined. Participants were recruited from a university population and were required to attend the biomechanics laboratory in the School of Health and Human Performance (Dublin City University) for all experimental measures. All participants completed a physical activity readiness questionnaire (PARQ) (ACSM, 1997) and informed consent form prior to participation. This study was granted ethical approval by DCU's ethics committee, as required.

6.2.1 Participants

Twelve male participants were recruited via email within the School of Health and Human Performance, and from the larger DCU and local area population. One participant subsequently dropped out of the study, leaving a total of 11 participants (aged 18-34, mass: 80.94 (±6.49) height: 180.09 (±4.55)). All participants were involved in running or running related activities for at least 30-mintues a week, for the six-month period prior to commencement of this study. Participants were excluded if any injuries (defined as any physical impairment affecting running distance, speed, duration or frequency), cardiovascular diseases, or any neurological disorder that may affect their gait were present or if they had any lower limb surgery (as in study 1,2,and 3)(Devita, Hunter & Skelly, 1992). Volunteers from study 1, 2, and 3 were excluded from participating in order to prevent the influence of learning and practice effects. Convenience based sampling was employed for recruitment of participants and therefore limits generalizability to wider populations.

Table 6.2 1:Participants height and mass

Participant	Mass (kg)	Height (cm)
1	66.6	172.0
2	79.9	178.0
3	82.0	178.0
4	84.5	182.5
5	98.0	188.0
6	80.5	184.0
7	73.9	174.0
8	78.3	181.5
9	82.6	179.0
10	78.0	183.0
11	86.0	181.0
Mean	80.9	180.1
Standard Deviation	±6.4	±4.5

6.2.2 Experimental procedure

The experimental procedure required each participant to visit the biomechanics lab in the School of Health and Human Performance (DCU) a total of 15 times. The first and final two visits were the main testing days and were 90-120 minutes in duration. The 12 sessions in between were the intervention sessions and increased from 20-40 minutes in duration across a 4-week period. There was a 1-week break between completion of the intervention and the final two testing days that brings the total duration of participation for each participant up to 7 weeks.

Baseline measures

Normal running measurements:

- Over-ground kinetic and kinematic measures
- Treadmill kinetic and kinematic measures
- Energy expenditure measurement
- Completed at self-selected paces

Intervention period

- 3 Biofeedback sessions weekly over 4 weeks
- Running time incrementally increased from 15-30 minutes
- Treadmill biofeedback provided using a faded feedback design (Table 6.2.1)

1-week break from practice

Post-Intervention Compliant running

New running style measurements:

- Over-ground kinetic and kinematic measures
- Treadmill kinetic and kinematic measures
- Energy expenditure measurement
- Completed at self-selected paces

Post-Intervention Normal running

Repeat of normal running measurements:

- Over-ground kinetic and kinematic measures
- Treadmill kinetic and kinematic measures
- Energy expenditure measurement
- Paces matched to post intervention compliant running

Figure 6.2 1: Experimental procedure

The protocol for all three experimental testing days was identical with running style the only varying factor. The baseline testing was used to gain baseline values. For the post-intervention complaint running tests participants were instructed to run using mechanics learnt as a result of the intervention period, and for the post intervention normal running tests participants were instructed to run "normally" again. The post intervention normal running tests were completed at matched paces to post intervention compliant running to provide direct kinetic and kinematic comparisons, as speed influences both (Mercer et al., 2002, Armpatzis et al., 1999). In addition this was used to determine if a training effect with regard to energy expenditure was present. During testing kinetic and kinematic variables were examined for over-ground and treadmill running, and energy expenditure variables were examined during treadmill running trials. All equipment used was calibrated as according to standard operating procedures.

6.2.3 Over-ground running

Over-ground running kinetics and kinematics were examined during baseline and post-intervention experimental measures using 12 200hz high speed cameras (Vicon Oxford Metrics, UK) by tracking the position of retroflective markers attached to 21 specific anatomical sites on each participant's body. The 21 markers were placed on each participant in accordance with the Vicon lower body and torso Plug-in-Gait model (Vicon Oxford Metrics, UK), and as in study 1. Ground reaction force data were collected with an AMTI forceplate (1200 X 600 mm, 2000 Hz, Advanced Mechanical Technology Inc., Watertown, MA) that was longitudinally orientated and inserted level with the ground (Kinsella and Moran, 2007). Vicons plug-in-gait model (Vicon Oxford Metrics, UK) in combination with anthropometric data were used to calculate kinetic and kinematic variables of interest. Data were collected as each participant completed 8 running trials and made a right foot contact with the forceplate. The five cleanest trials (trials with the least amount of gaps in marker data) were selected and subsequently averaged. Gaps in marker data were filled using Vicon Bodybuilder software(Vicon Oxford Metrics, UK). Speed was standardized to self-selected pace and controlled with the use of speed gates set over a ten meter distance.

6.2.4 Treadmill running trials

Upon completion of overground running trials each participant completed five treadmill runs. Each run was 6-minutes in duration in order to facilitate stabilization of kinetic (White, Gilchrist & Christina 2002) and kinematic (Lavcanska, Taylor & Schache 2005) running variables. Running speed for each 6-minute run was different and was determined based on each participant's self-selected running speed. Self-selected (SS) pace was described as the pace that each participant would select if going for a 30-minute run. The five running speeds were as follows: SS, SS-1km/hr, SS-2 km/hr, SS+1 km/hr, and SS+2 km/hr. These were completed in random order, with the two fastest speeds excluded from randomization for the first 6-minute run. SS pace for baseline testing was determined during a warm-up run whereby participants ran for 4-6 minutes. SS pace for the first of the post-intervention testing days was determined based on the average of the last three SS pace selections during the intervention period. These speeds were matched for the final testing day.

Prior to commencement of the treadmill running trials two uniaxial capacitive accelerometers were attached to the tibia and sacrum, and a third accelerometer was attached to the treadmill. Accelerometers were attached as previously explained in study 2 and study 3.

To measure kinematic variables retrofelective markers were attached at 12 predetermined anatomical positions (as in study 2 and 3) (as in figures 4.2.3, 4.2.4 and table 4.2.1).

Both accelerometer and kinematic data were collected for a 15-second window at the end of each 6-minute running trial. A custom MATLAB program (The MathsWorks Inc., United Kingdom) was used to calculate both lower extremity kinematics and to identify impact acceleration peaks. From each 15-second window of data collection, the custom MATLAB program (The MathsWorks Inc., United Kingdom) identified each consecutive foot strike. This data was then averaged.

6.2.5 Energy Expenditure

Breath-by-breath oxygen consumption was measured for the duration of each 6-minute treadmill running trial using the K4b2 portable pulmonary gas exchange system (Cosmed, The metabolic company, Italy). Steady-state O_2 data were averaged over the last 2-minutes of each 6-minute trial (Clansey et al, 2014). The standard Weir equation (1949) was used to calculate caloric expenditure from steady sate data (as seen below).

$$Calorie\ expenditure\ (kcal \cdot min^{-1}) = 3.94 * \dot{V}O_2\ (\mathrm{l} \cdot min^{-1}) - 1.1 * \dot{V}CO_2\ (\mathrm{l} \cdot min^{-1})(\text{Weir, 1949})$$

Each participant's running speed was subsequently used to calculate COT (ll.km^{-1}) by determining how long it would take to complete to complete 1km at each speed and then multiplying that value by kcal.min^{-1}.

COT (kcal.km^{-1})= Time taken to complete 1km *Kca.min^{-1}

Cost of locomotion and cost of transport were measured as they provide specific ecological meaning (Steudel-Numbers & Wall Scheffler, 2009). Time of day, footwear, and room temperature were all controlled for each energy expenditure measurement. A food and hydration self report diary was completed by each participant prior to baseline measurements. Participants were asked to follow the same recorded nutrition prior to each day of testing. Participants were asked to refrain from exercise the day prior to each testing day, however this was not objectively monitored.

6.2.6 Intervention period

During the intervention period each participant was required to visit the biomechanics lab three times a week for supervised training. Running time increased incrementally from 15-minutes up to a maximum of 30-minutes (as in Crowell et al, 2012, and Cheung et al., 2011), across the 4-week period. Prior to the first session of biofeedback, each participant received the same information pack detailed in study 3. Visual biofeedback was provided continuously for the first four sessions and gradually removed in a faded-feedback style. This has been shown to successfully impart reduced tibial loading, which is maintained at a 1-month follow-up, following similar biofeedback-based interventions (Crowell et al., 2011, Cheung et al., 2011). It is thought that a faded feedback design most effectively imparts motor learning of a new skill (Schmidt et al., 1989). Biofeedback was provided from the treadmill-mounted accelerometer, as described in study 2. The intervention and faded biofeedback schedule can be seen in table 6.2.2.

Table 6.2 2: Faded Biofeedback schedule

Session number	Running Time	Feedback schedule
1	15	• Continuous
2	17	• Continuous
3	19	• Continuous
4	21	• Continuous
5	23	• 5 minutes at start, • 5minutes in middle, • 5minutess at end
6	25	• 4 minutes at start, • 4mins in middle • 4mins at end
7	27	• 3 minutes at start • 3mins in middle • 3 minutes at end
8	29	• 2 minutes at start • 2minutes in middle • 2minutes at end
9	30	• 2 minutes at start • 1minute in middle • 2minutes at end
10	30	• 1.5minutes at start • 1min at 14min30 • 1.5 minutes at 28.5 min
11	30	• 1minute at start, • 1minute at14min30secs • 1minute at 29min mark
12	30	• 45seconds at start • 30 seconds at 14mins 45secs • 45seconds at 29mins15secs

6.2.7 Dependent variables

For each stride the following variables were measured.

Over-ground running

The following variables are reported for stance period:

- Vertical ground reaction force
- Ankle, knee, and hip moments
- Ankle, knee, and hip angles
- COM vertical height
- Contact time

Treadmill running

The following variables are reported over a full stride cycle:

- Peak tibial and sacral accelerations
- Ankle, knee, and hip angles
- COM vertical height

Energy expenditure

- Cost of Locomotion (COL)
- Cost of Transport (COT)
- Running Economy (RE)

With reference to treadmill running, kinetic and kinematic data were collected for 15-second windows and averaged. Over-ground running variables were determined by an average of 5 trials.

6.2.8 Data analysis

In order to determine the effect of an accelerometer biofeedback-based intervention on peak accelerations at the tibia and sacrum, and on COT, COL, and RE multiple repeated measures ANOVA's were completed.

In order to determine the effect of running speed on peak accelerations at the tibia and sacrum and RE at each time point (baseline normal, post intervention compliant, and post intervention normal) multiple repeated measures ANOVA were also completed. For all tests, statistical significance was set at $p<0.05$.

In order to do complete full curve comparisons for all kinematic and kinetic data an Analysis of Characterizing Phases was performed (Richter et al., 2014) as described in study 2 (4.2.7). Multiple paired sample T-tests (with a Bonferonni adjustment) were subsequently completed to determine significantly different key phases. Please note that each curve represents a full stance period for over-ground kinetics and kinematics and a full cycle (initial contact to initial contact) for treadmill kinematics.

Normality of data was determined using the Shapiro-Wilk test for normality. Mauchleys test was used to determine sphericity. In cases where the assumption of sphericity was violated a Greenhouse-Geisser correction was employed.

6.3 Results

6.3.1Peak impact accelerations

Multiple repeated measures ANOVA's were conducted to assess the impact of an accelerometer based biofeedback intervention on peak tibial and sacral acceleration values across three time points (baseline normal, post intervention compliant, and post intervention normal) at various paces (self-selected, normal self-selected, and compliant self-selected).

At self-selected paces (note: post intervention normal is always matched to post intervention compliant for pace), both the tibia ($F(2,18)=12.415$, $p<0.05$, partial eta squared= 0.580) and sacrum ($F(2,18)=6.469$, $p<0.05$, partial eta squared= 0.418) displayed a significant main effect of time. Pairwise comparisons for tibial accelerations indicated that post intervention compliant running displayed smaller values than both baseline normal (40%, $p<0.05$) and post intervention normal running (20%, $p<0.05$). A similar pattern was observed at the sacrum, with post intervention compliant running displaying smaller accelerations than baseline normal running (42%, $p<0.05$) and post intervention normal running (36%, $p<0.05$).

Subsequent power analysis indicates an observed power of 0.977 with regard to tibial accelerations and 0.849 with regard to sacral accelerations, at self selected paces. Thus indicating that the study was adequately powered to determine an effect of the biofeedback intervention on impact accelerations.

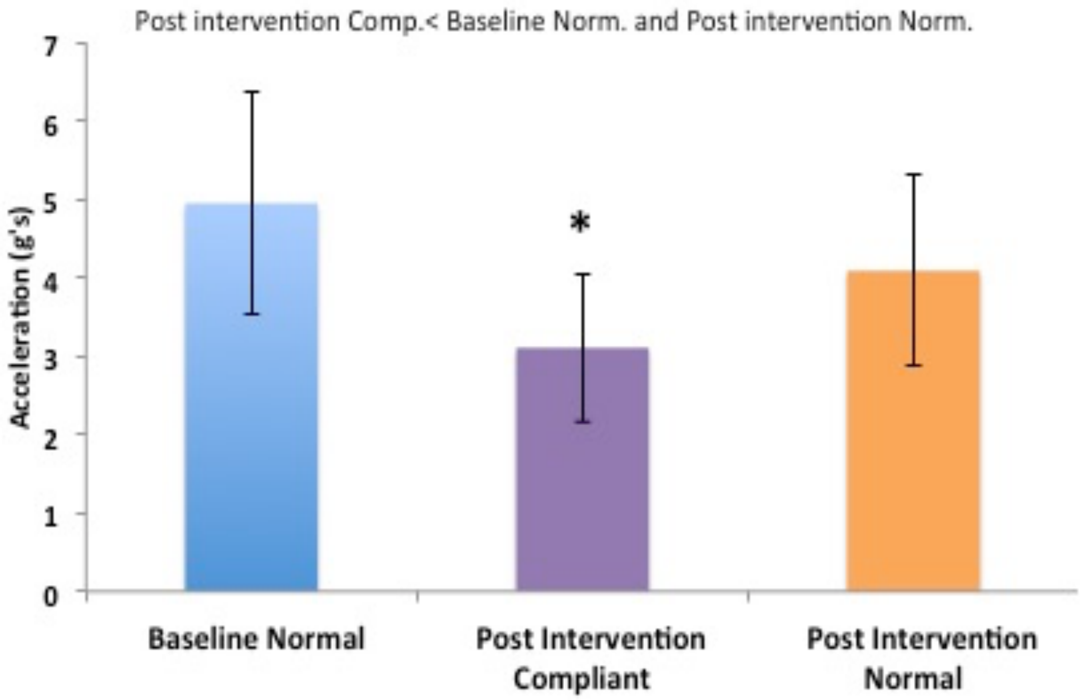

Figure 6.3 1: Peak tibial acceleration for each condition at self selected paces (mean+SD)(*p<0.05). Note: post intervention normal always matched to post intervention compliant.

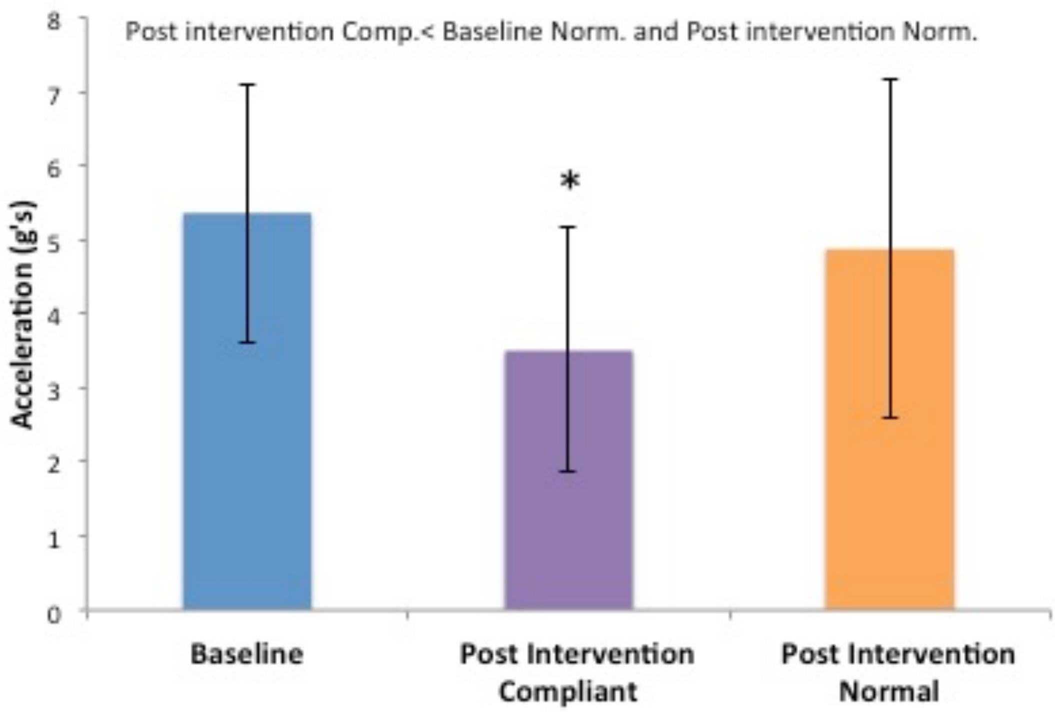

Figure 6.3 2: Peak sacral acceleration for each condition at self selected paces(mean+SD)(*p<0.05).

When pace was matched to post intervention compliant self-selected pace similar results were displayed, with both tibial (F (2,18)=11.141, p<0.05, partial eta squared= 0.533) and sacral (F (2,18)=4.457, p<0.05, partial eta squared= 0.311) accelerations presenting a significant main effect for time.

Pairwise comparison revealed that post intervention compliant running displayed significantly smaller tibial accelerations than both baseline normal (44%, p<0.05) and post intervention normal (27%, p<0.05). At the sacrum post intervention compliant running displayed smaller accelerations than baseline normal (38%, p<0.05) and post intervention normal (33%, p<0.05)

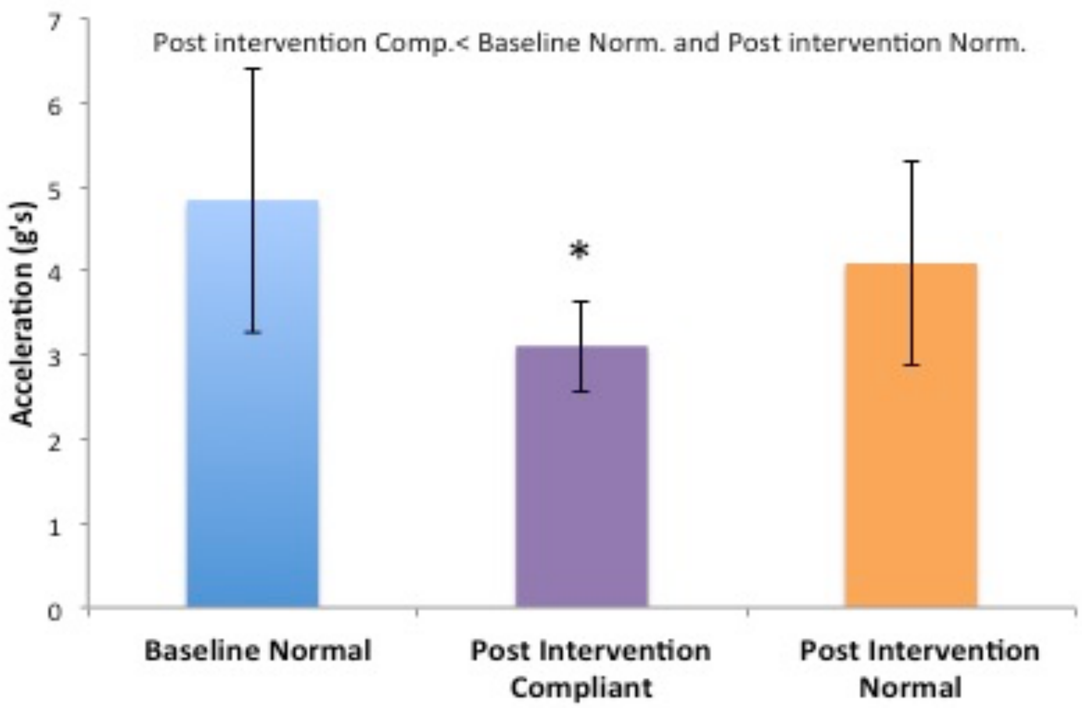

Figure 6.3 3:Peak tibial acceleration for each condition at Compliant SS pace(mean+SD) (*p<0.05).

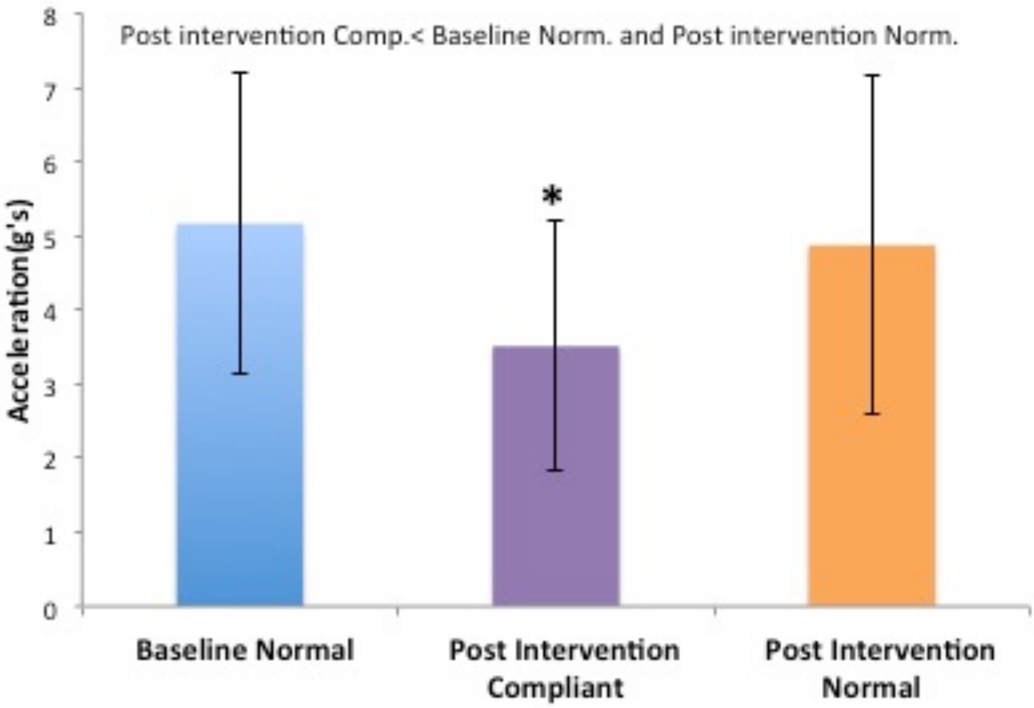

Figure 6.3 4: Peak sacral acceleration for each condition at Compliant SS pace (mean+SD)(*p<0.05).

When pace was matched to baseline normal self-selected pace tibial accelerations (F (2,18)= 9.122, p<0.05, partial eta squared= 0.503) and sacral accelerations (F (2,18)= 5.701, p<0.05, partial eta squared= 0.388) both displayed a significant main effect of time. Pairwise comparisons revealed that post intervention compliant tibial accelerations were smaller than both baseline normal (40%, p<0.05) and post intervention normal (26%,p<0.05). At the sacrum post intervention compliant running produced smaller values than both baseline normal (36%, p<0.05) and post intervention normal running (30%, p<0.05).

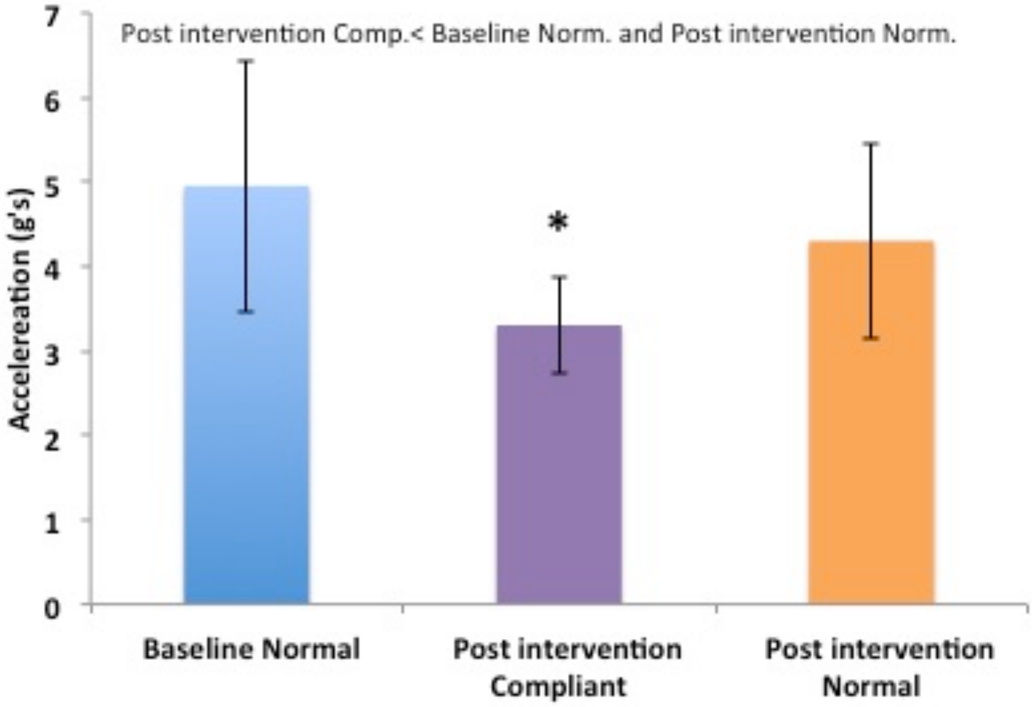

Figure 6.3 5: Peak tibial acceleration for each condition at Normal SS pace (Mean+SD)(*p<0.05).

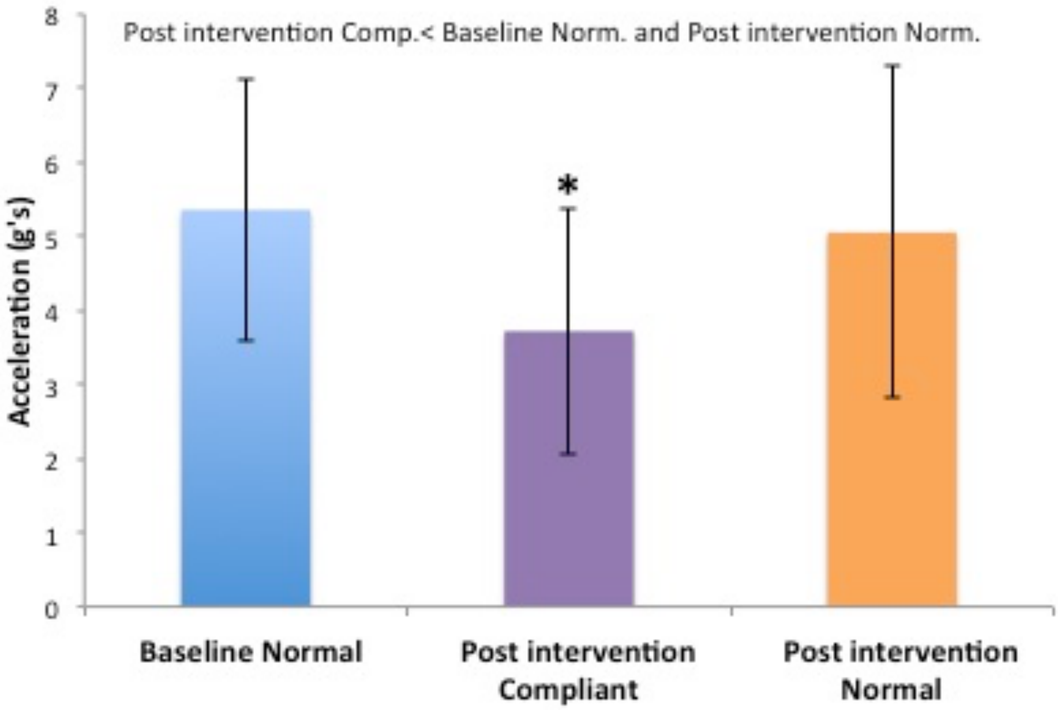

Figure 6.3 6:Peak sacral acceleration for each condition at Normal SS pace(mean+SD) (* p<0.05).

Table 6.3 1: Peak tibial acceleration for each running style and pace.

Measure	Running Style	Pace	Mean (g's)	Confidence Interval (95%)	
				Lower	Upper
Peak tibial acceleration	Normal	SS	4.952	3.884	6.021
		Compliant SS	4.842	3.717	5.968
	Compliant	SS	3.108	2.720	3.496
		Normal SS	3.304	2.892	3.716
	Post intervention Normal	Compliant SS	4.091	3.220	4.960
		Normal SS	4.302	3.476	5.126

Table 6.3 2: Peak sacral acceleration for each running style and pace.

Measure	Running Style	Pace	Mean (g's)	Confidence Interval (95%)	
				Lower	Upper
Peak sacral acceleration	Normal	SS	5.360	4.096	6.626
		Compliant SS	5.166	3.708	6.623
	Compliant	SS	3.511	2.305	4.717
		Normal SS	3.722	2.529	4.915
	Post intervention Normal	Compliant SS	4.875	3.243	6.507
		Normal SS	5.056	3.458	6.654

Table 6.3 3: Percentage differences between conditions at each pace for peak tibial acceleration values (* indicates a significant difference, p<0.05)

Pace	Location	Baseline Normal Vs. Post Intervention Compliant (% Diff.)	Baseline Normal Vs. Post Intervention Normal (% Diff.)	Post Intervention Compliant Vs. Post Intervention Normal (% Diff.)
Self Selected paces	Tibia	40%*	20%	-20%*
	Sacrum	42%*	7%	-36%*
Compliant Self Selected pace	Tibia	44%*	17%	-27%*
	Sacrum	38%*	6%	-33%*
Normal Self Selected pace	Tibia	40%*	14%	-26%*
	Sacrum	36%*	6%	-30%*

6.3.2 Cost of Locomotion and Cost of Transport

Multiple repeated measures ANOVA's were conducted to assess the impact of an accelerometer based biofeedback intervention on cost of transport (kcal.kg^{-}1.km^{-1}) and cost of locomotion (kcal.kg^{-}1.km^{-1}) across three time points (baseline normal, post intervention compliant, and post intervention normal) at various paces (self-selected, normal self-selected, and compliant self-selected).

No significant main effect of time was displayed for cost of transport at self-selected paces ($F_{(2,16)} = 1.495$, $p=0.254$, partial eta squared = 0.157), when paces were matched to the compliant self-selected pace ($F_{(2,16)}=2.5$, $p=0.114$, partial eta squared=0.238), or when paces were matched to the normal self-selected pace ($F_{(2,16)}=3.215$, $p=0.071$, partial eta squared= 0.315).

Similarly, no significant main effect of time was displayed for cost of locomotion when paces were matched to the compliant self-selected pace ($F_{(2,16)}=2.387$, $p=0.124$, partial eta squared=0.230), or when paces were matched to the normal self-selected pace ($F_{(2,16)}=3.157$, $p=0.074$, partial eta squared= 0.311). However, a significant main effect of time was displayed for self-selected paces ($F_{(2,16)}= 7.772$, $p<0.05$, partial eta squared= 0.493). Pairwise comparison revealed that baseline normal displayed significantly larger COL than post intervention normal (8%, $p<0.05$)

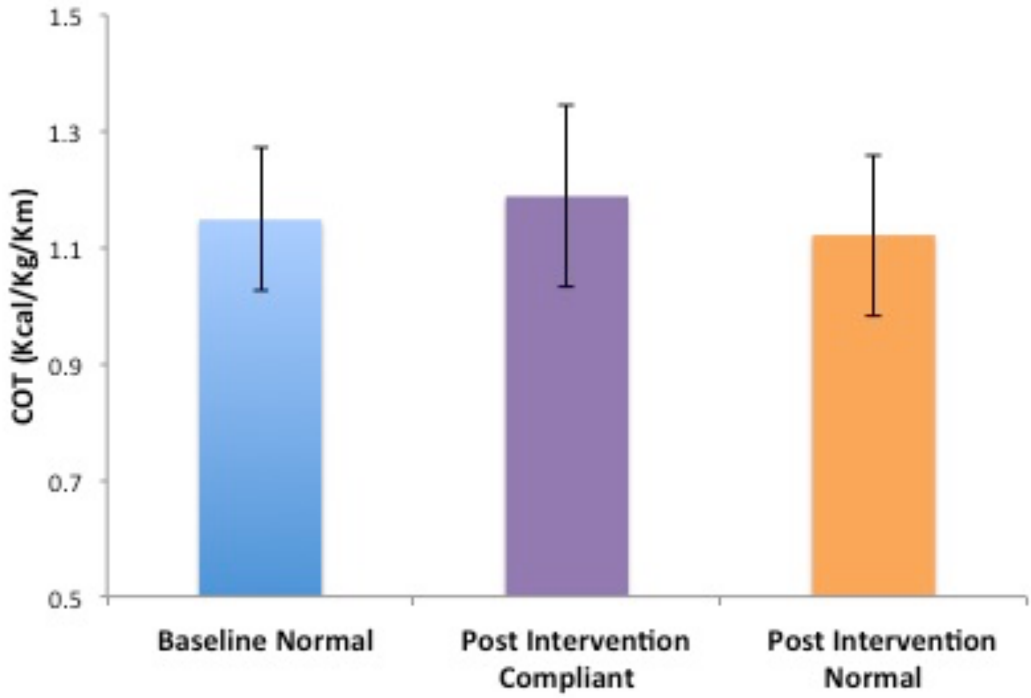

Figure 6.3 7: Cost of Transport at self selected paces. Note: Post intervention Normal always matched to post intervention Compliant (Mean + Standard deviation).

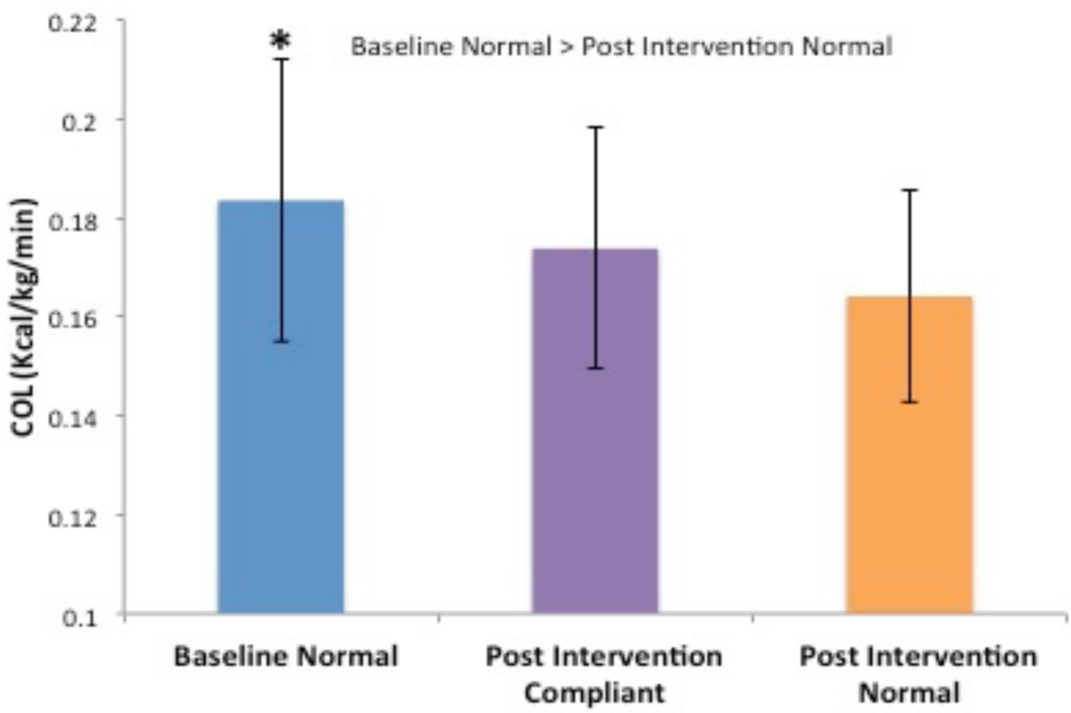

Figure 6.3 8: Cost of Locomotion at self selected paces (*p<0.05) (Mean + Standard deviation).

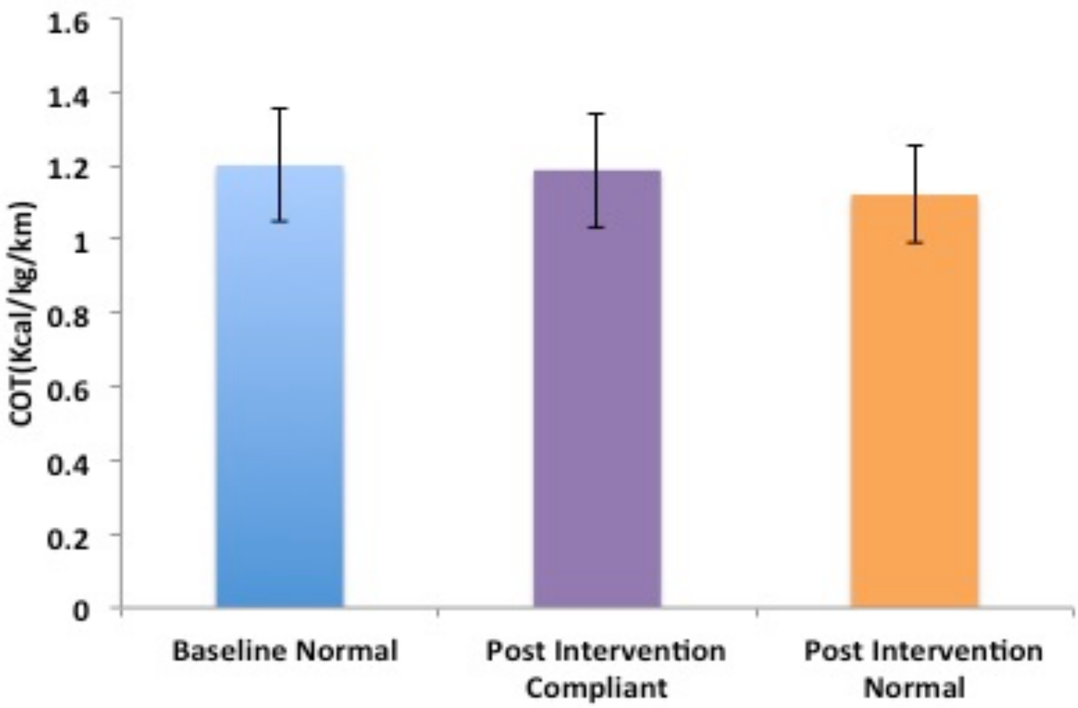

Figure 6.3 9:Cost of Transport at Compliant SS pace($p<0.05$) (Mean + Standard deviation).

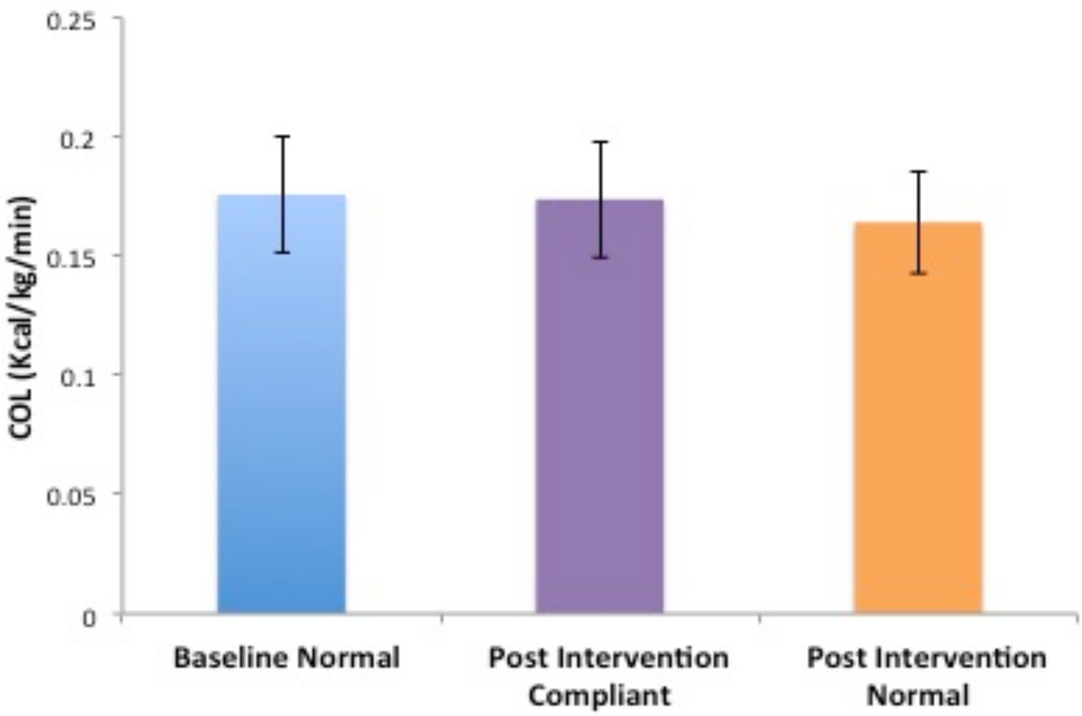

Figure 6.3 10:Cost of Locomotion at Compliant SS pace($p<0.05$) (Mean + Standard deviation).

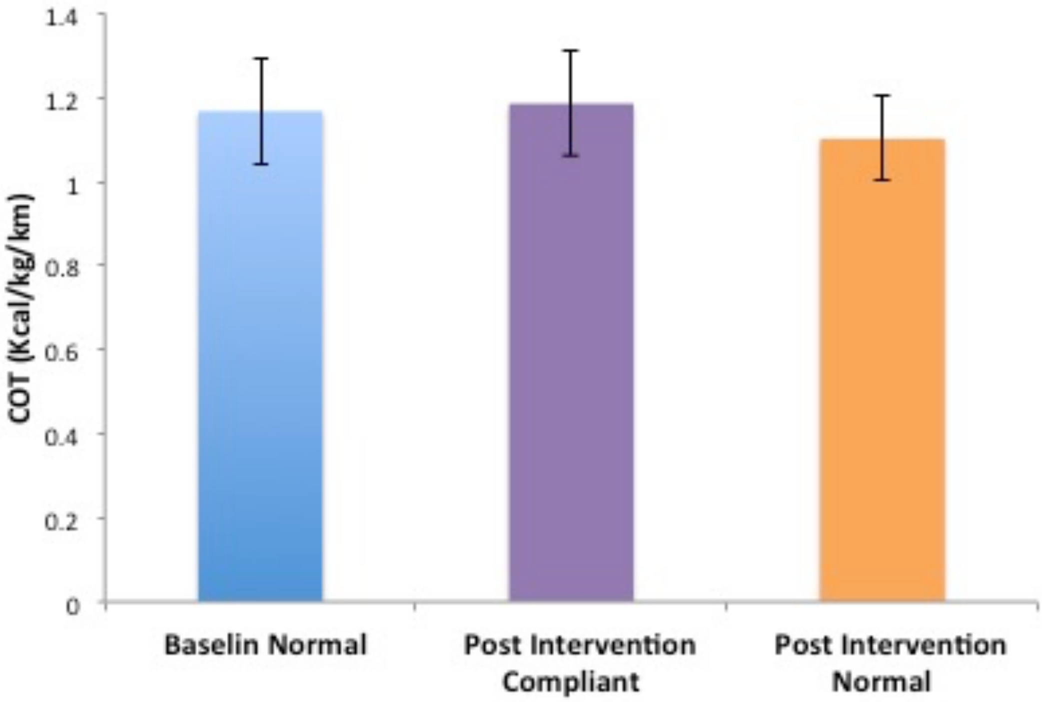

Figure 6.3 11:Cost of Transport at Normal SS pace(P<0.05) (Mean + Standard deviation).

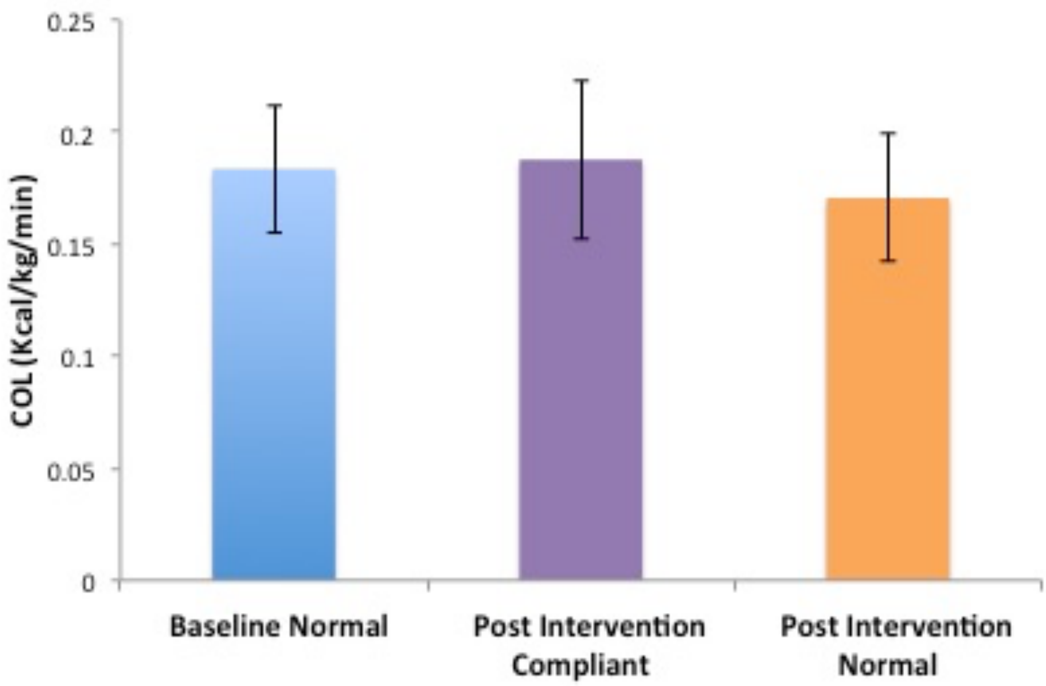

Figure 6.3 12:Cost of Locomotion at Normal SS pace(P<0.05) (Mean + Standard deviation).

Table 6.3 4: Cost of locomotion for each style at each pace.

Measure	Running Style	Pace	Mean (kcal.kg-1.min-1)	Confidence Interval (95%)		10mins running (Kcal.kg-1)	20mins running (Kcal.kg-1)	30mins running (Kcal.kg-1)	40mins running (Kcal.kg-1)
				Lower	Upper				
Cost of Locomotion (kcal.kg-1.min-1)	**Normal**	**SS**	0.184	0.162	0.205	1.84	3.68	5.52	7.36
		Compliant SS	0.176	0.157	0.194	1.76	3.52	5.28	7.04
	Compliant	**SS**	0.174	0.155	0.193	1.74	3.48	5.22	6.96
		Normal SS	0.186	0.154	0.211	1.86	3.72	5.58	7.44
	Post intervention Normal	**Compliant SS**	0.164	0.148	0.181	1.64	3.28	4.92	6.56
		Normal SS	0.170	0.147	0.194	1.70	3.40	5.10	6.8

Table 6.3 5: Cost of transport for each condition at each pace.

Measure	Running Style	Pace	Mean (kcal.kg-1.km-1)	Confidence Interval (95%)		10Km running (Kcal.kg-1)	20Km running (Kcal.kg-1)	30Km running (Kcal.kg-1)	40Km running (Kcal.kg-1)
				Lower	Upper				
Cost of Transport (kcal.kg-1.km-1)	Normal	SS	1.213	1.054	1.242	12.13	24.26	36.39	48.52
		Compliant SS	1.201	1.084	1.319	12.01	24.02	36.03	48.04
	Compliant	SS	1.126	1.068	1.308	11.26	22.52	33.78	45.04
		Normal SS	1.186	1.069	1.286	11.86	23.72	35.58	47.44
	Post intervention Normal	Compliant SS	1.124	1.022	1.222	11.24	22.48	33.72	44.96
		Normal SS	1.103	1.018	1.187	11.03	22.06	33.09	44.12

Table 6.3 6: Running economy, Cost of locomotion, and Cost of transport for each speed.

Measure	Running Style	Units	SS-2km.hr^{-1}	SS-1km.hr^{-1}	SS	SS+1km.hr^{-1}	SS+2km.hr^{-1}
Energy Expenditure measures	Normal	$\dot{V}$ O$_2$.ml^{-1}.kg^{-1}.min^{-1}	32.52 (±5.79)	34.59 (±5.78)	37.13 (±5.72)	38.44 (±5.67)	41.39 (±5.47)
		Kcal. kg^{-}1.min^{-1}	0.16 0.03	0.17 0.03	0.18 0.03	0.19 0.04	0.22 0.04
		Kcal.kg^{-1}.km^{-1}	1.31 (±0.17)	1.25 (±0.18)	1.21 (±0.19)	1.08 (±0.20)	1.13 (±0.19)
	Compliant	$\dot{V}$ O$_2$.ml^{-1}.kg^{-1}.min^{-1}	29.80 (±4.27)	33.49 (±4.81)	35.55 (±4.96)	37.49 (±6.77)	41.00 (±5.45)
		Kcal. kg^{-}1.min^{-1}	0.14 (±0.02)	0.16 (±0.02)	0.17 (±0.04)	0.18 (±0.03)	0.20 (±0.03)
		Kcal.kg^{-1}.km^{-1}	1.24 (±0.19)	1.25 (±0.16)	1.13 (±0.25)	1.12 (±0.19)	1.11 (±0.14)
	Post intervention Normal	$\dot{V}$ O$_2$.ml^{-1}.kg^{-1}.min^{-1}	27.11 (±6.01)	30.63 (±4.07)	33.60 (±4.85)	35.51 (±5.33)	38.25 (±4.58)
		Kcal. kg^{-}1.min^{-1}	0.13 (±0.03)	0.15 (±0.02)	0.16 (±0.02)	0.17 (±0.02)	0.19 (±0.02)
		Kcal.kg^{-1}.km^{-1}	1.19 (±0.25)	1.16 (±0.15)	1.12 (±0.13)	1.07 (±0.14)	1.06 (±0.11)

6.3.3 Running Economy

Multiple repeated measures ANOVA's were conducted to assess the impact of an accelerometer based biofeedback intervention on running economy across three time points (baseline normal, post intervention compliant, and post intervention normal) at various paces (self-selected, normal self-selected, and compliant self-selected).

No significant main effect of time was displayed when paces were matched to the post intervention compliant pace ($F(2,16)= 2.012$, $p=0.166$, partial eta squared=0.201) or the baseline normal pace ($F(2,16)=2.350$, $p=0.132$, partial eta squared=0.251). However a significant main effect of time was displayed when compared at self-selected paces ($F(2,16)=6.205$, $p<0.05$, partial eta squared= 0.437). Pairwise comparisons indicated that baseline normal displayed significantly larger values than post intervention normal (6%, $p<0.05$).

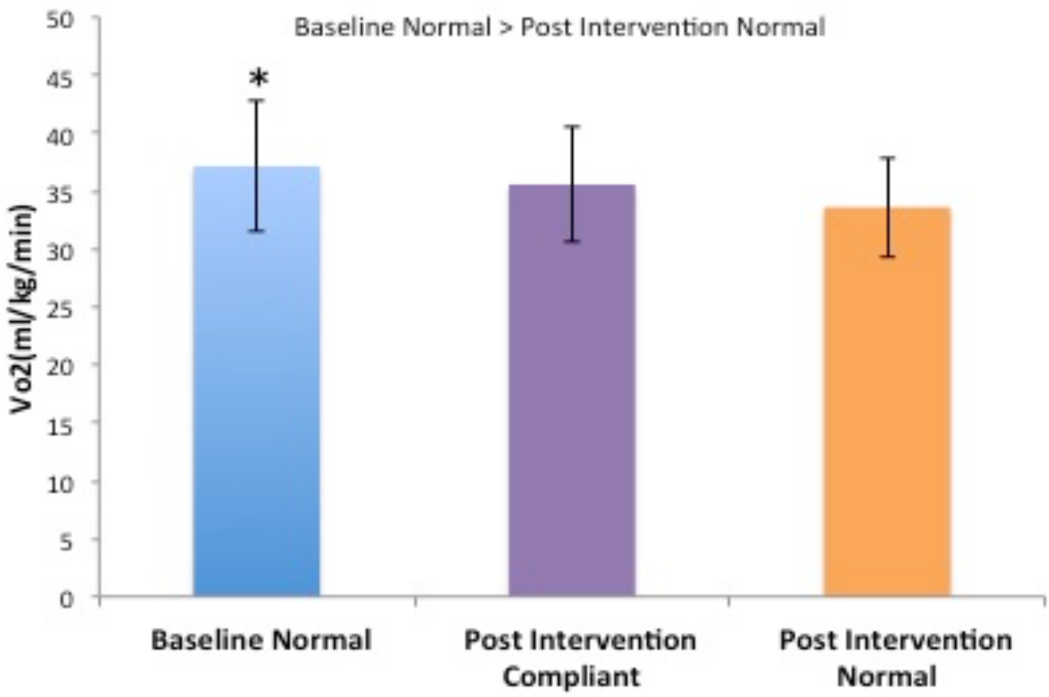

Figure 6.3 13:Running Economy at SS paces (Mean + SD) (* indicates a significant difference to post intervention normal, p<0.05).

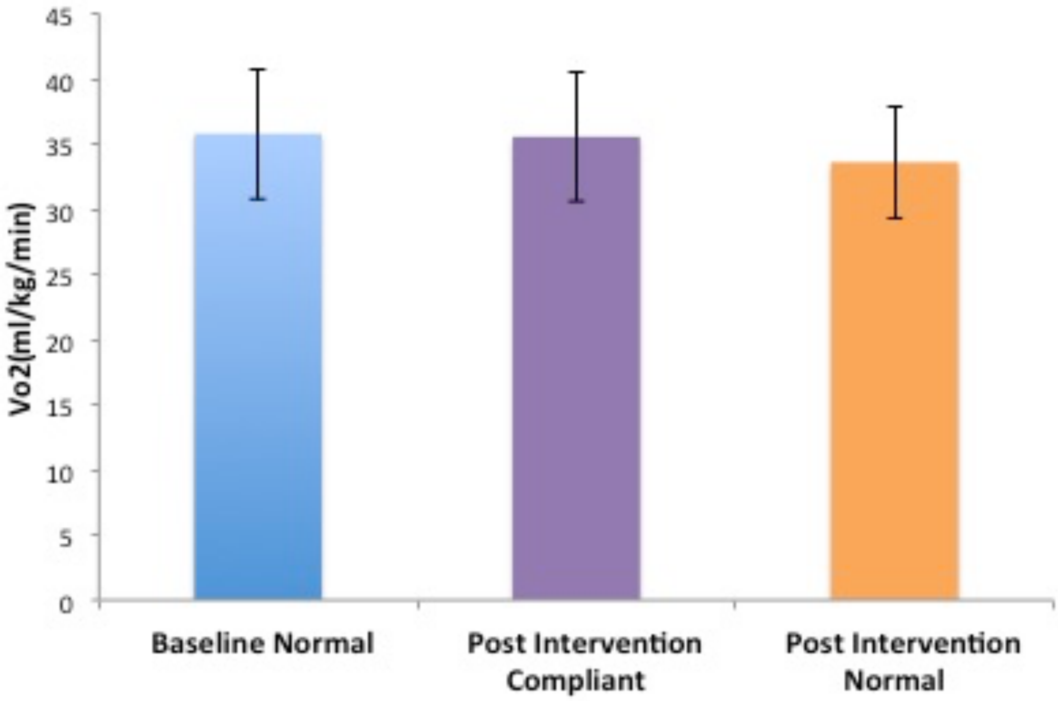

Figure 6.3 14:Running Economy at Compliant SS(Mean + SD).

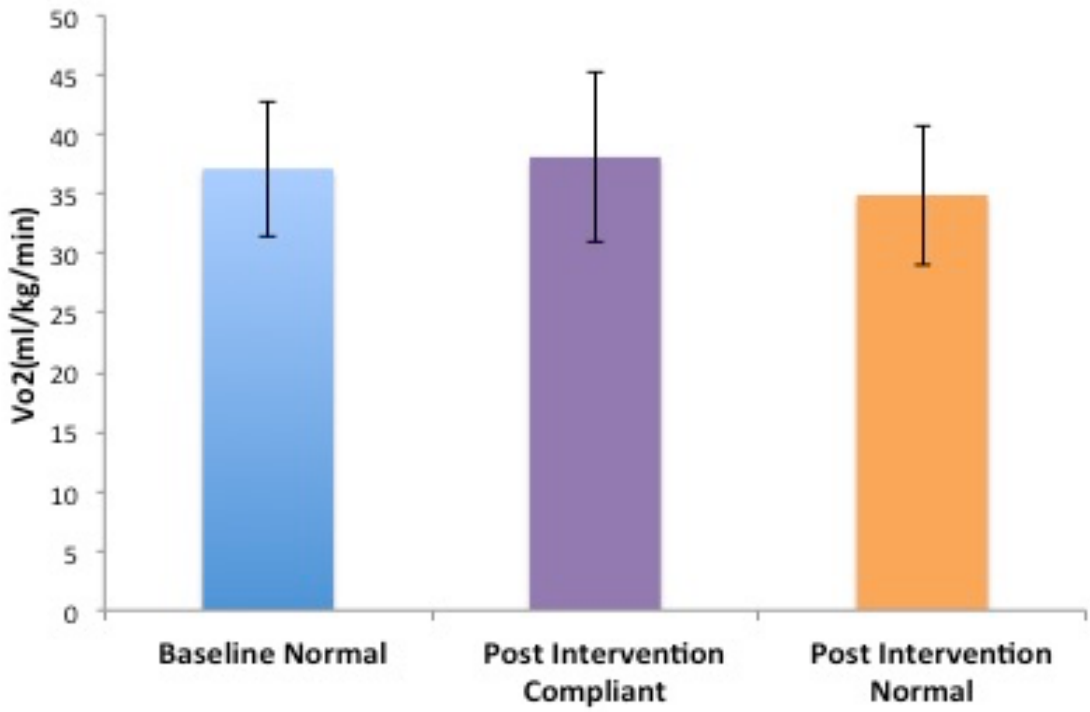

Figure 6.3 15: Running Economy at Normal SS pace(Mean + SD).

Table 6.3 7: Running Economy for each condition at each pace.

Measure	Running Style	Pace	Mean	Confidence Interval (95%)	
				Lower	Upper
Running Economy ($\dot{V}O_2$.kg^{-1}.min^{-1})	**Normal**	**SS**	37.13	32.74	41.53
		Compliant SS	35.78	31.92	39.64
	Compliant	**SS**	35.55	31.74	39.36
		Normal SS	38.11	31.32	42.73
	Post intervention Normal	**Compliant SS**	33.60	30.29	36.90
		Normal SS	34.92	30.01	39.83

6.3.4 The affect of speed on peak acceleration values

Multiple repeated measures ANOVA's were completed to examine the effect of speed (SS-2km/hr, SS-1km/hr, SS, SS+1km/hr, and SS+2km/hr) on peak tibial and sacral accelerations for each condition (baseline normal, post intervention compliant, and post intervention normal).

A significant main effect of speed was displayed at both the tibia and sacrum for baseline normal running (tibia: $F_{(4,36)}=10.55$, $p<0.05$, partial eta squared= 0.540)(sacrum: $F_{(4,36)}=6.243$, $p<0.05$, partial eta squared= 0.410), post intervention compliant running (tibia: $F_{(4,36)}=9.068$, $p<0.05$, partial eta squared= 0.502)(sacrum: $F_{(1.949, 17.537)}=9.891$, $p<0.05$, partial eta squared=0.524) and post intervention normal running (tibia: $F_{(2.196, 19.765)}=9.730$, $p<0.05$, partial eta squared= 0.519)(sacrum: $F_{(4,36)}= 12.863$, $p<0.05$, partial eta squared=0.588).

The locations of these differences were revealed by pairwise comparisons and can be most easily observed in figures 6.3.16-6.3.21.

Table 6.3.7 displays the average paces for baseline normal running and compliant running (post intervention normal running pace was patched to post intervention compliant pace). Paces were determined based on each participant's self-selected pace (as explained in section 6.2.4).

Table 6.3 8: Speed at each pace, for each participant, for normal and compliant running.

Running style	Participants	SS-2km/hr	SS-1km/hr	SS	SS+1km/hr	SS+2km/hr
Baseline Normal running	1	7	8	9	10	11
	2	9	10	11	12	13
	3	8	9	10	11	12
	4	7	8	9	10	11
	5	6	7	8	9	10
	6	7	8	9	10	11
	7	8	9	10	11	12
	8	9	10	11	12	13
	9	8	9	10	11	12
	10	6	7	8	9	10
	11	7	8	9	10	11
	Mean	7.44	8.44	9.44	10.44	11.44
Post intervention compliant running	1	7	8	9	10	11
	2	7	8	9	10	11
	3	7	8	9	10	11
	4	7	8	9	10	11
	5	7	8	9	10	11
	6	6	7	8	9	10
	7	7	8	9	10	11
	8	7	8	9	10	11
	9	8	9	10	11	12
	10	6	7	8	9	10
	11	7	8	9	10	11
	Mean	6.91	7.91	8.91	9.91	10.91

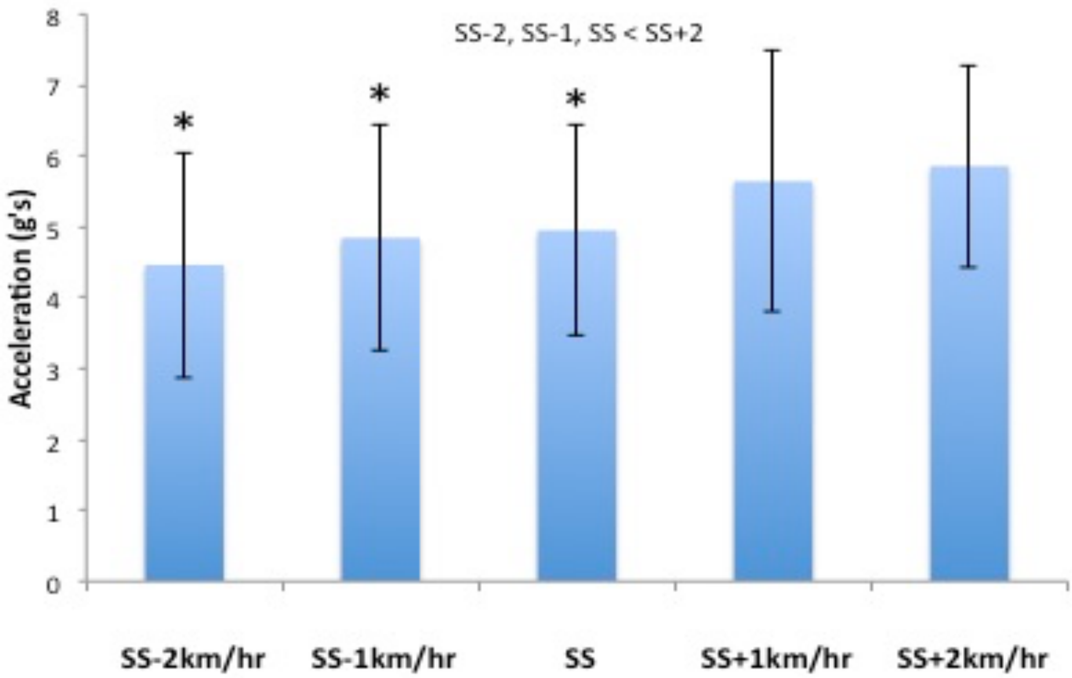

Figure 6.3 16:The effect of running speed on peak tibial accelerations for normal running (Mean + SD) (*p<0.05).

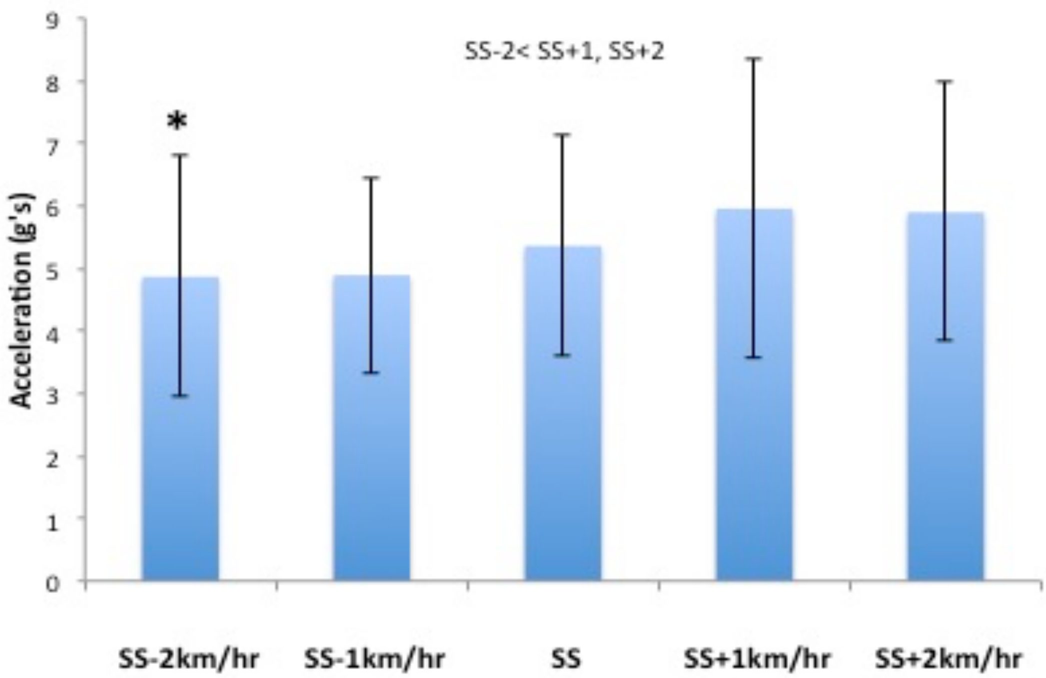

Figure 6.3 17:The effect of running speed on peak sacral accelerations for normal running(Mean + SD) (*p<0.05)

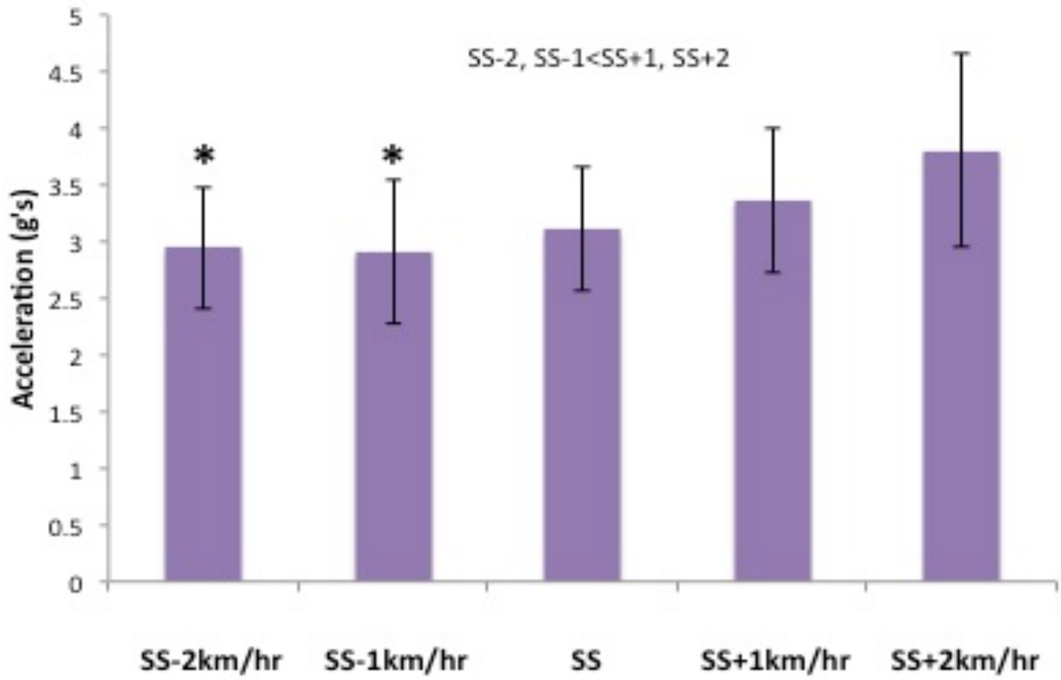

Figure 6.3 18:The effect of running speed on peak tibial accelerations for post intervention Compliant running(Mean + SD) (*p<0.05)

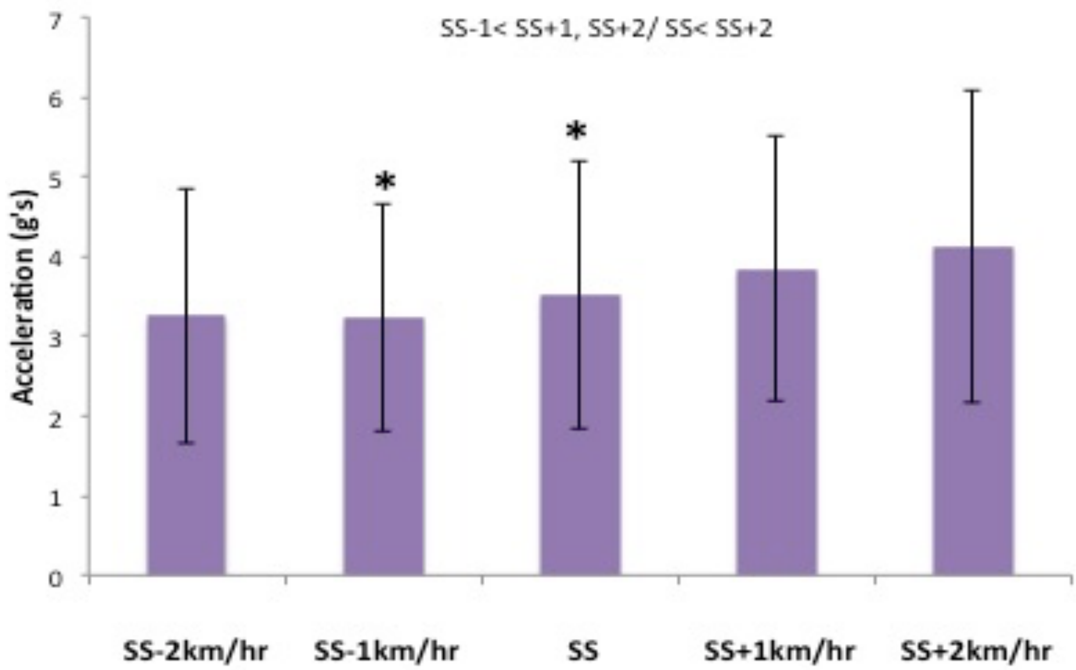

Figure 6.3 19: The effect of running speed on peak sacral accelerations for post intervention Compliant running(Mean + SD) (*p<0.05)

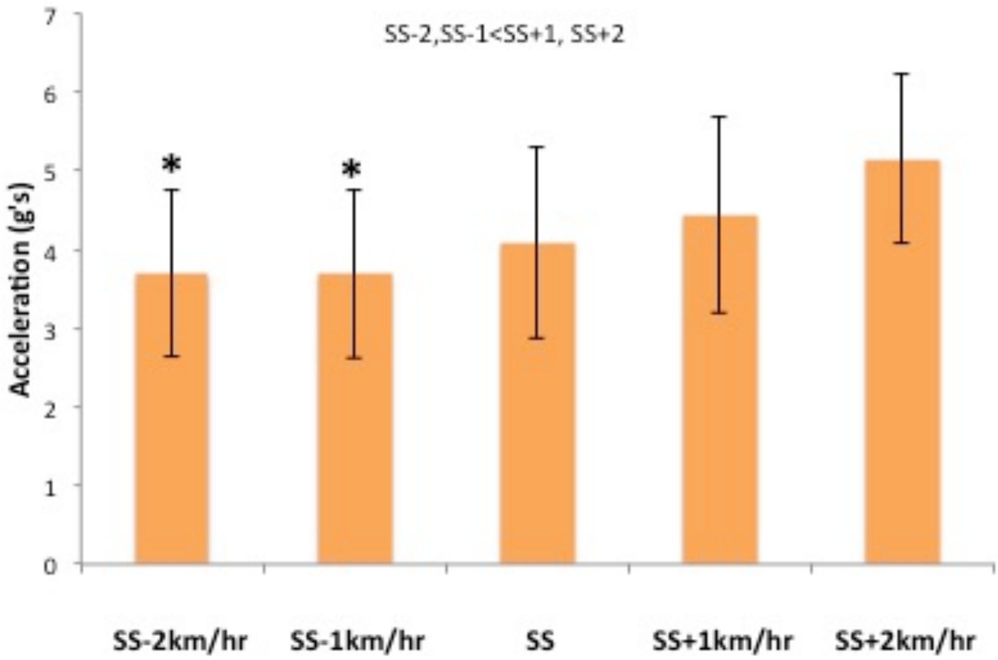

Figure 6.3 20: The effect of running speed on peak tibial accelerations for post intervention normal running(Mean + SD) (* p<0.05)

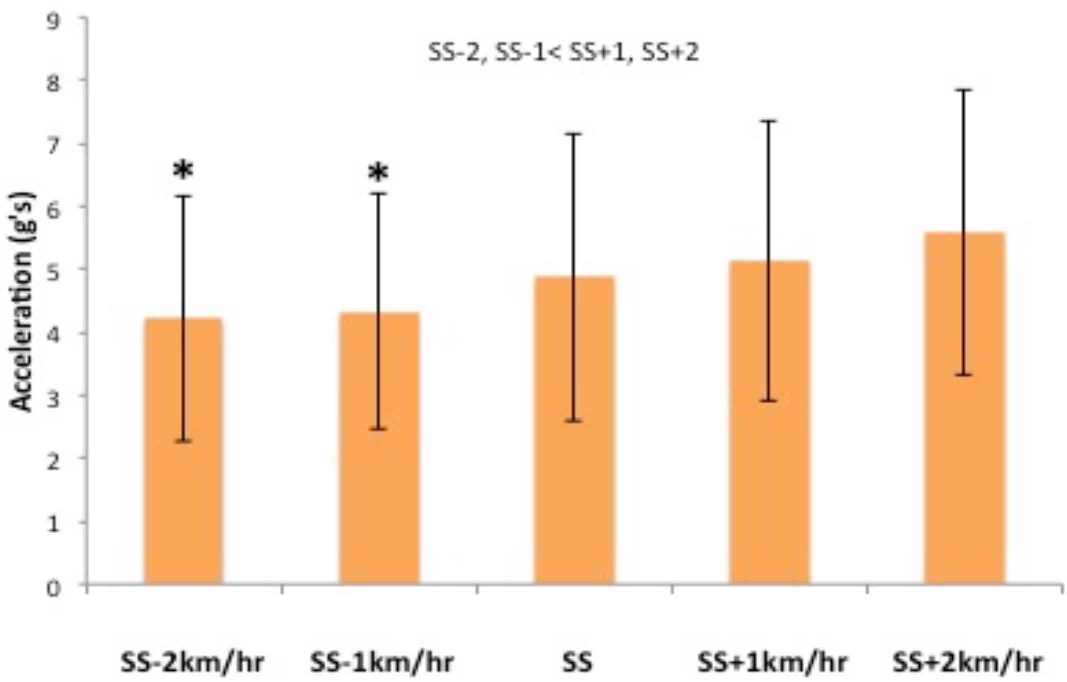

Figure 6.3 21:The effect of running speed on peak sacral accelerations for post intervention normal running(Mean + SD) (*p<0.05).

Table 6.3 9: The effect of speed on peak acceleration values at the tibia and sacrum (*p<0.05)

Running Style	Location	SS-2 Vs. SS-1	SS-2 Vs. SS	SS-2 Vs. SS+1	SS-2 Vs. SS+2	SS-1 Vs. SS	SS-1 Vs. SS+1	SS-1 Vs. SS+2	SS Vs. SS+1	SS Vs. SS+2	SS+1 Vs. SS+2
Normal	Tibia	-8%	-10%	-23%	-27%*	-2%	-15%	-18%*	-13%	-16%*	-3%
	Sacrum	-1%	-10%	-20%*	-19%*	-9%	-20%	-19%	-11%	-10%	1%
Compliant	Tibia	1%	-5%	-13%*	-25%*	-7%	-14%*	-27%*	-8%	-20%	-12%
	Sacrum	1%	-8%	-16%	-24%	-8%	-17%*	-24%*	-9%	-16%*	-7%
Post interventio n Normal	Tibia	0%	-10%	-13%*	-33%*	-10%	-18%*	-33%*	-8%	-23%	-15%
	Sacrum	-3%	-14%	-20%*	-28%*	-12%	-17%*	-25%*	-5%	-14%	-8%

6.3.5 The effect of speed on O_2 cost of running

Multiple repeated measures ANOVA's were completed to examine the effect of speed (SS-2km/hr, SS-1km/hr, SS, SS+1km/hr, and SS+2km/hr) on O_2 consumption for each condition (baseline normal, post intervention compliant, and post intervention normal).

A significant main effect of speed was displayed for baseline normal running (F (1.279,10.235)=15.533, p<0.05, partial eta squared= 0.660), post intervention compliant running (F (1.1964, 15.712= 49.231, p<0.05, partial eta squared= 0.860)), and post intervention normal running (F (1.443, 8/659)= 27.422, p<0.05, partial eta squared=0.820). Pairwise comparison revealed the location of these differences and can be most easily observed in table 6.3.22- 6.3.24.

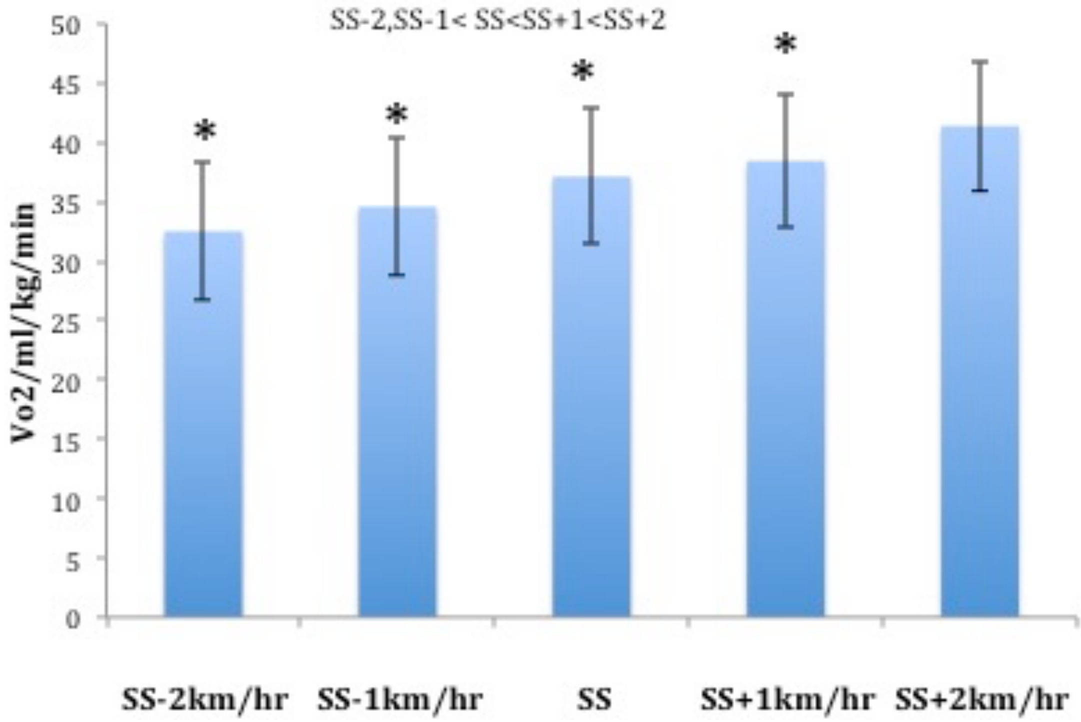

Figure 6.3 22:The effect of running speed on O2 consumption for normal running(Mean + SD) (* p<0.05).

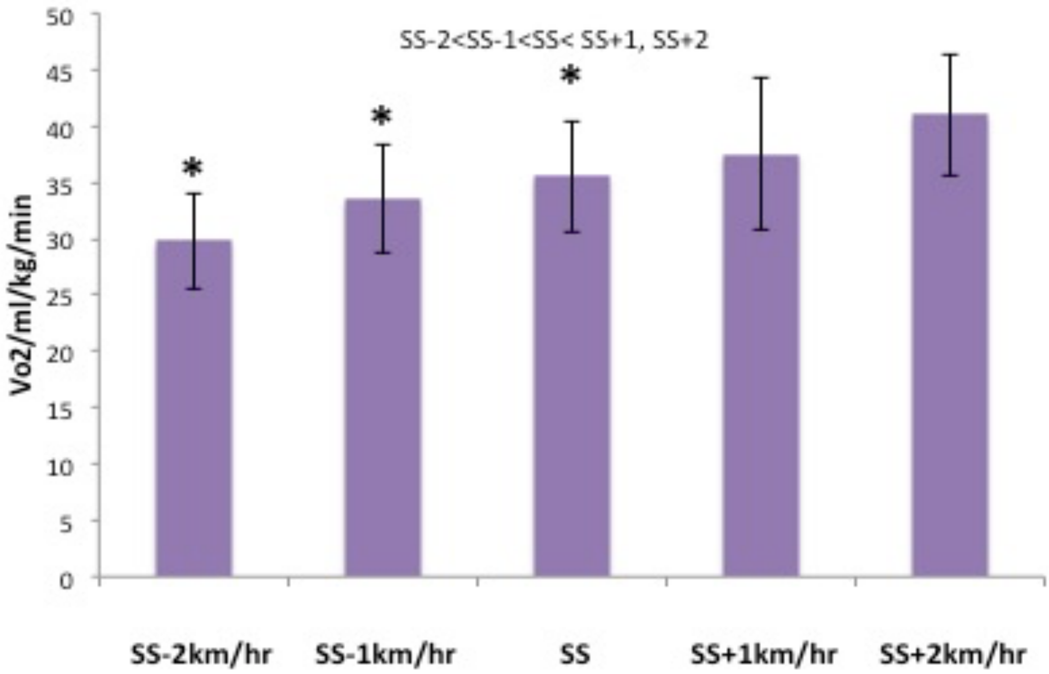

Figure 6.3 23: The effect of running speed on O2 consumption for Compliant running(Mean + SD) (*p<0.05)

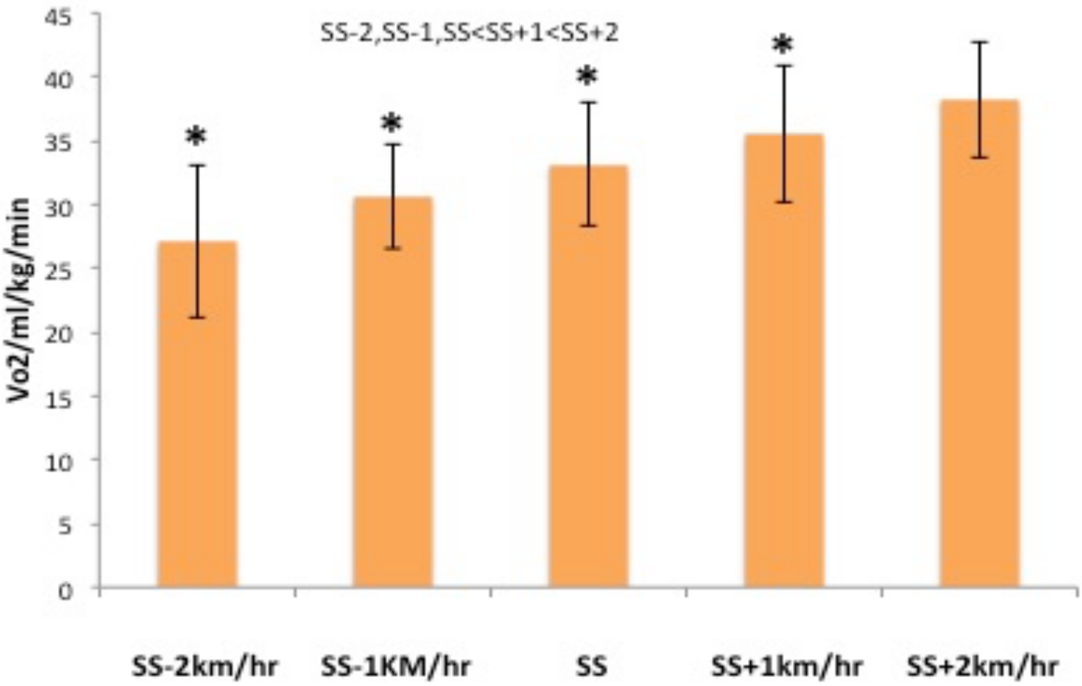

Figure 6.3 24:The effect of running speed on O2 consumption for post intervention normal running(Mean + SD) (* p<0.05)

Table 6.3 10: The effect of speed on $\dot{V}O_2$ for each style (* indicates a significant % difference)

Running Style	SS-2 Vs. SS-1	SS-2 Vs. SS	SS-2 Vs. SS+1	SS-2 Vs. SS+2	SS-1 Vs. SS	SS-1 Vs. SS+1	SS-1 Vs. SS+2	SS Vs. SS+1	SS Vs. SS+2	SS+1 Vs. SS+2
Normal	-6	-13*	-17*	-24*	-7*	-11*	-18*	-3*	-11*	-7*
Compliant	-12%*	-18%*	-23%*	-32%*	-6%*	-11%*	-20%*	-5%*	-14%*	-9%
Post intervention Normal	-12%	-20%	-27%*	-34%*	-8%	-15%*	-22%*	-7%*	-14%*	-7%*

6.3.10 Over-ground Kinetics

Post intervention compliant running demonstrated a significantly lower vertical ground reaction force than both baseline for 4-82% of stance ($T=4.16, p<0.016$, partial eta squared=1.43), and post intervention normal for 6-67% of stance ($T=4.11$, $P<0.016$, partial eta squared= 1.46). Furthermore, baseline normal running displayed significantly larger vertical ground reaction force values than post intervention normal running for 32-61% of stance ($T= 4.15$, $P<0.016$, partial eta squared= 0.59).

Post intervention compliant running demonstrated significantly lower ankle flexor moments than both baseline normal running for 21-85% of stance ($T=2.91, p<0.016$, partial eta squared= 1.24), and post intervention normal for 10-84% of stance ($T=3.14$, $p<0.016$, partial eta squared=1.28).

Post intervention compliant running displayed significantly lower knee flexor moments than both baseline normal running for 2-4% of stance ($T=3.02$, $p<0.016$, partial eta squared=1.28), 18-46% of stance ($T=4.55, p<0.016$, partial eta squared=1.05), and post intervention normal running for 17-43% of stance ($T=2.62$, $p<0.016$, partial eta squared= 0.78). Post intervention compliant running demonstrated significantly larger knee flexor moments than baseline normal running for 75-100% of stance ($T=3.73$, $p<0.016$, partial eta squared= 1.3), and post intervention normal running for 61-100% of stance ($T=3.21$, $p<0.016$, partial eta squared=1.05).

Post intervention compliant running demonstrated significantly lower knee abductor moments than baseline normal running for 1-10% of stance ($T=2.93$, $p<0.016$, partial eta squared=1.21) and post intervention normal running for 16-50% of stance ($T=3.20, p<0.016$, partial eta squared=0.91).

Post intervention compliant running demonstrated significantly larger hip flexor moments than baseline normal running for 2-4% of stance ($T=3.05, p<0.016$, partial eta squared= 1.23) an 20-27% of stance ($T=5.22$, $p<0.016$, partial eta squared=0.84) and significantly lower hip flexor for 10-14% of stance ($T=3.77$, $p<0.016$, partial eta squared=0.95). Post intervention compliant running also displayed significantly larger hip flexor moments than post intervention normal running for 48-57% of stance ($T=3.10$, $p<0.016$, partial eta squared=0.86) and 18-25% of stance ($T=3.09$, $P<0.016$, partial eta squared =0.89). Furthermore, post intervention normal running displayed significantly larger hip flexor moments than baseline normal for 2-4% of stance ($T=3.37$, $p<0.016$,partial eta squared= 0.87).

Post intervention compliant running demonstrated significantly lower hip abductor moments than baseline normal running for 18-85% of stance ($T=4.86$,$p<0.016$, partial eta squared=1.42) and 4-12% of stance ($T=4.12$,$p<0.016$, partial eta squared=0.99), and post intervention normal for 18-74% of stance ($T=3.46$,$p<0.016$, partial eta squared=1.36).

Note: for all kinematic and kinetic variables, significantly different key phases are highlighted in red below each curve.

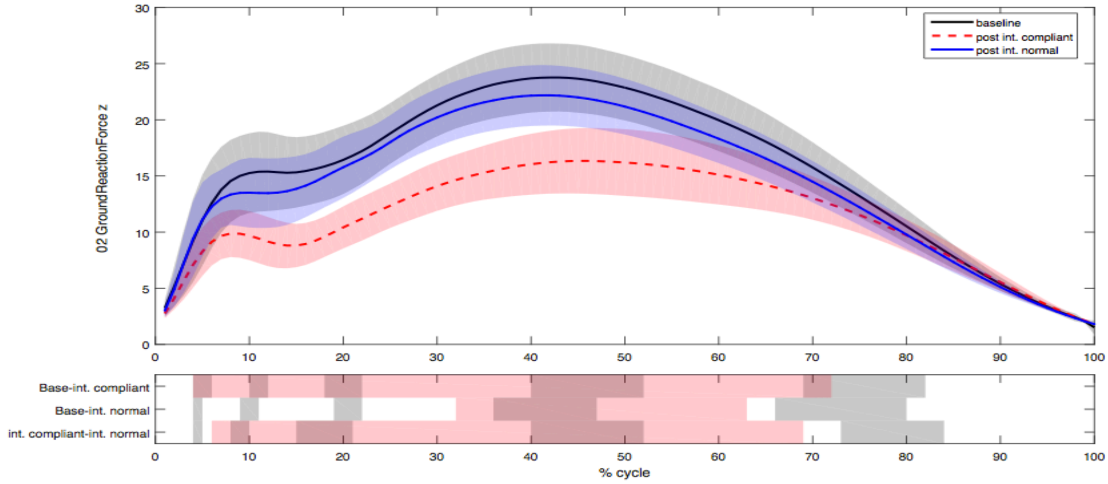

Figure 6.3 25: The effect of a 4-week treadmill accelerometer based biofeedback intervention on vertical ground reaction force (N/kg)

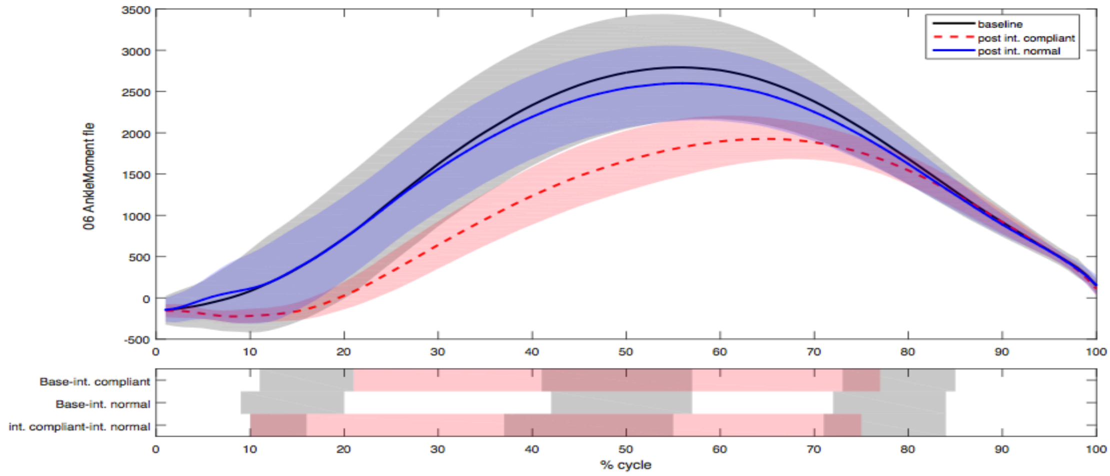

Figure 6.3 26: The effect of a 4-week treadmill accelerometer based biofeedback intervention on ankle flexor moments (N/mm/kg).

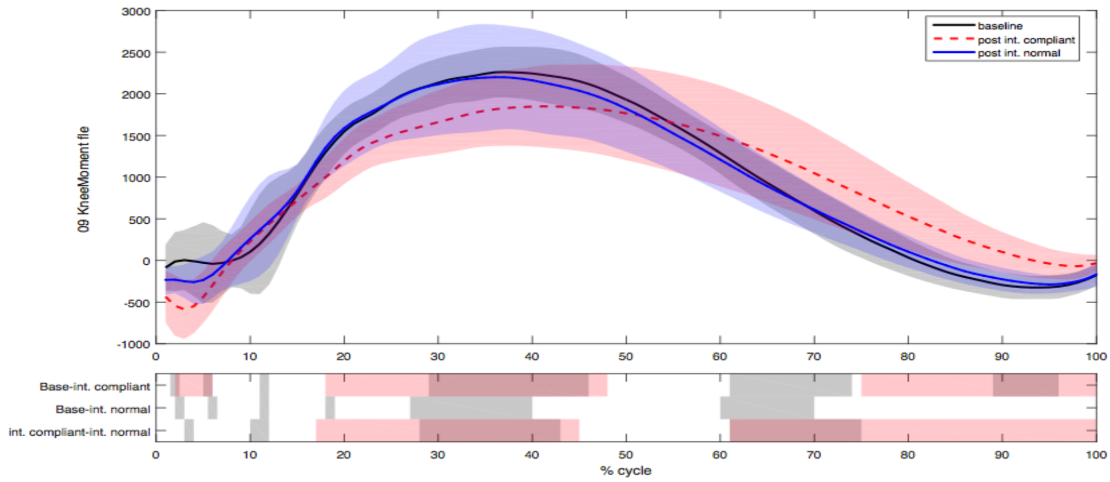

Figure 6.3 27: The effect of a 4-week treadmill accelerometer based biofeedback intervention on knee flexor moments (N/mm/kg).

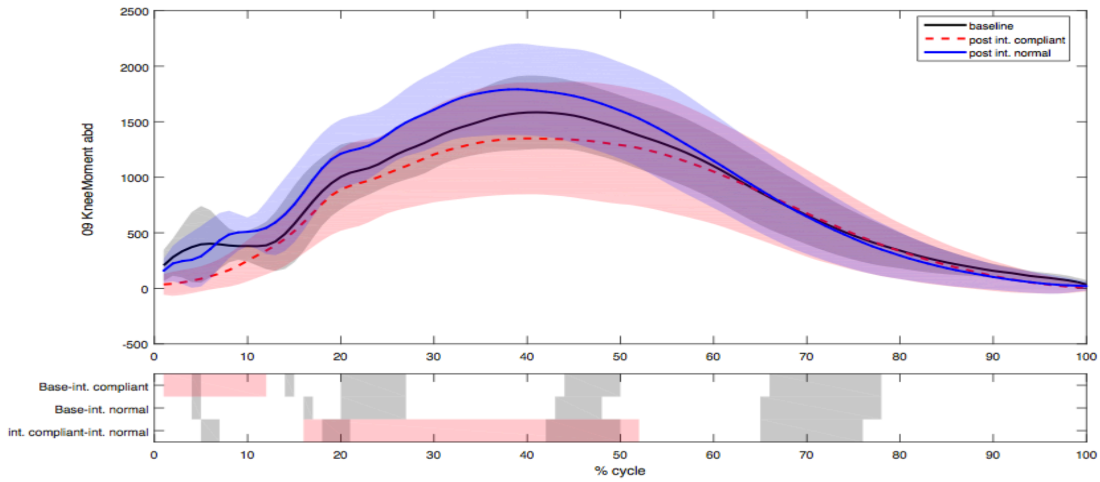

Figure 6.3 28: The effect of a 4-week treadmill accelerometer based biofeedback intervention on knee abductor moments (N/mm/kg).

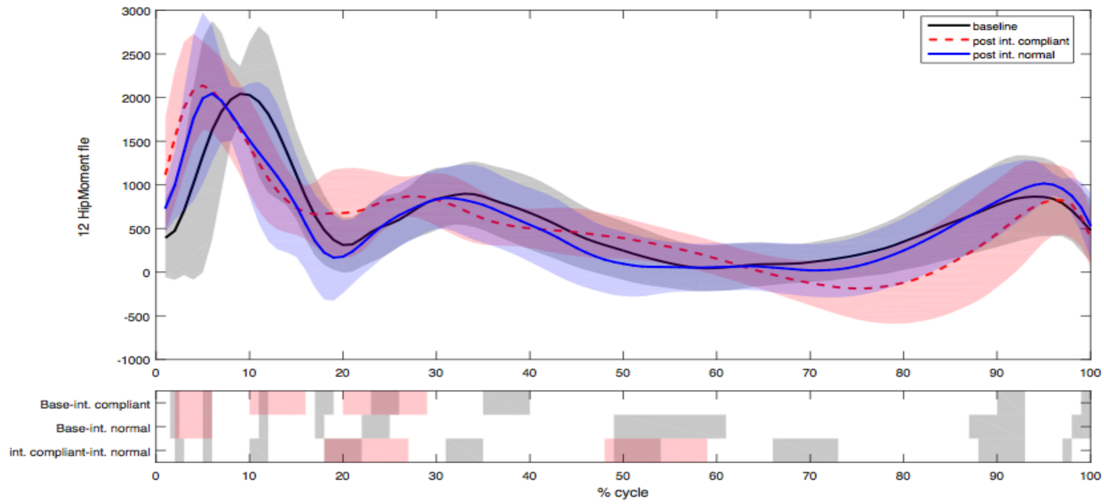

Figure 6.3 29: The effect of a 4-week treadmill accelerometer based biofeedback intervention on hip flexor moments (N/mm/kg).

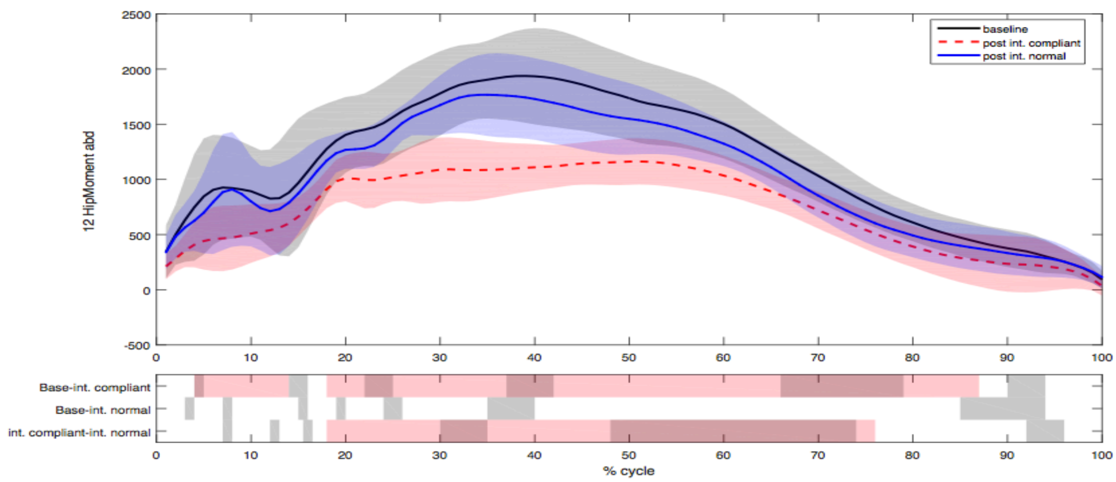

Figure 6.3 30: The effect of a 4-week treadmill accelerometer based biofeedback intervention on hip abductor moments (N/mm/kg).

6.3.11 Over-ground Kinematics

Participants displayed a more dorsiflexed ankle for post intervention compliant running relative to baseline normal running for 57-100% of stance ($T=3.39, p<0.016$, partial eta squared= 0.77) and relative to post intervention normal running for 72-100% of stance ($T= 2.84$, $p<0.016$, partial eta squared=0.99).

Participants displayed a significantly more flexed knee during stance for post intervention compliant running, when compared to baseline normal running for 1-100% of stance ($T=3.1, p<0.016$, partial eta squared= 1.17), and when compared to post intervention normal running for 83-100% of stance ($T=2.99, p<0.016$, partial eta squared=1.05).

Post intervention compliant running demonstrated a significantly larger amount of hip flexion than baseline normal running for 1-35% of stance ($T=3.6, p<0.016$, partial eta squared = 1.08), and than post intervention normal running for 1-27% of stance ($T=2.79, p<0.016$, partial eta squared= 0.89).

Post intervention compliant running displayed a significantly lower COM than baseline for 100% of stance ($T=3.22$, $p<0.016$, partial eta squared=0.77), and lower than post intervention normal running for 1-84% of stance ($T=3.25, p<0.016$, partial eta squared= 1.46).

Finally, post intervention compliant running demonstrated a significantly longer contact time than both baseline normal running (11-100% of stance)($T=3.88, p<0.016$, partial eta squared=1.36), and post intervention normal running for 11-100% of stance ($T=3.46, p<0.016$, partial eta squared= 1.24). Furthermore, post intervention normal running displayed a significantly longer contact time than baseline normal running for 11-100% of stance ($T=3.35, p<0.016$, partial eta squared=0.519).

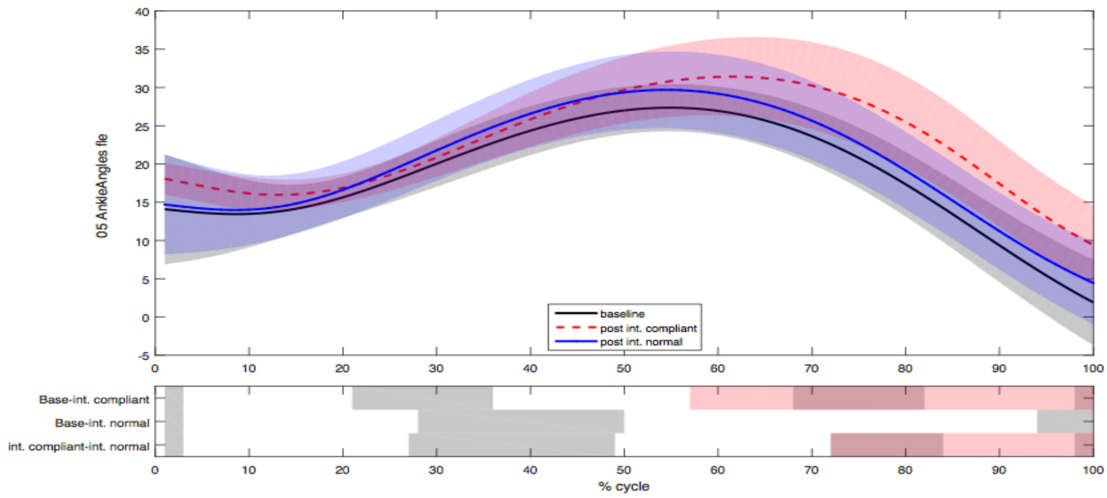

Figure 6.3 31: The effect of a 4-week treadmill accelerometer based biofeedback intervention on ankle angle during stance (degrees).

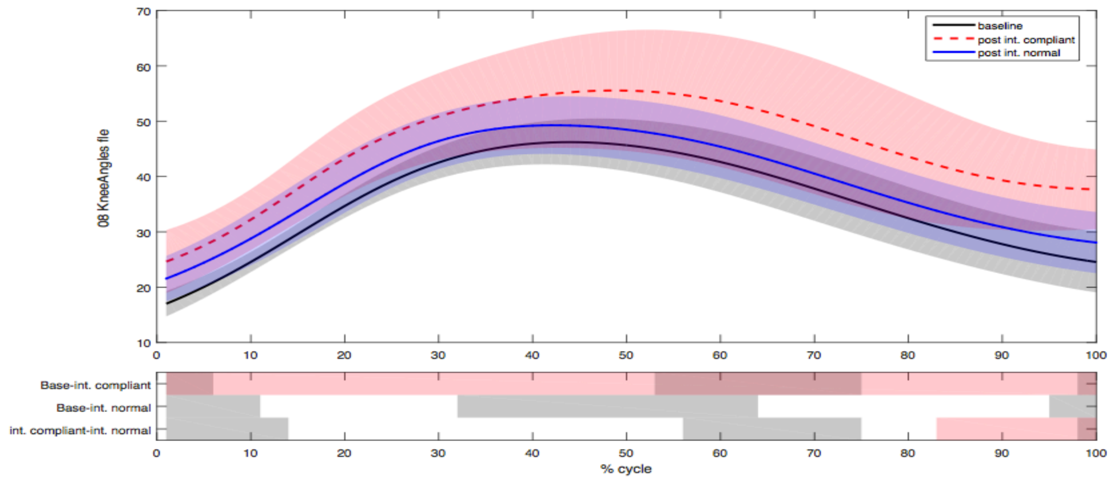

Figure 6.3 32: The effect of a 4-week treadmill accelerometer based biofeedback intervention on knee angle during stance (degrees).

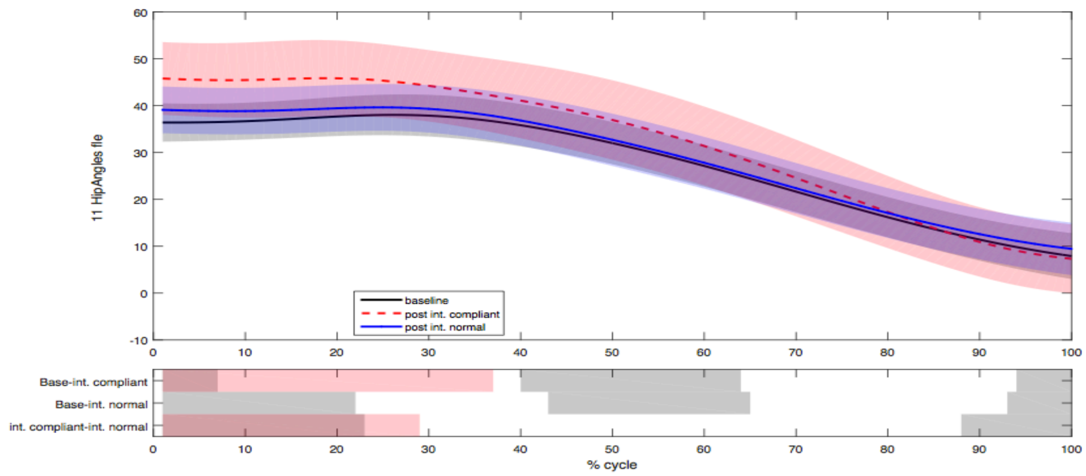

Figure 6.3 33: The effect of a 4-week treadmill accelerometer based biofeedback intervention on hip angle during stance (degrees).

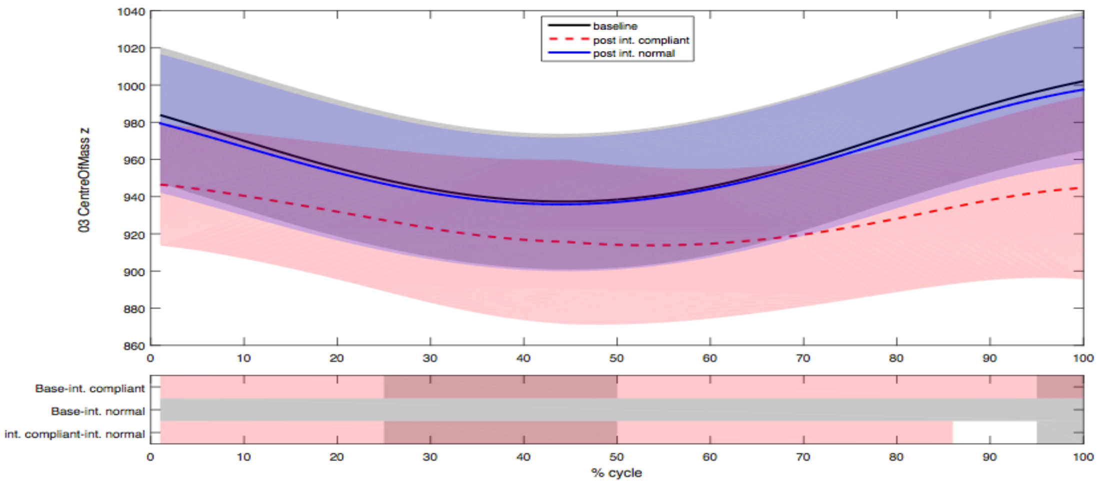

Figure 6.3 34: The effect of a 4-week treadmill accelerometer based biofeedback intervention on COM height during stance (mm).

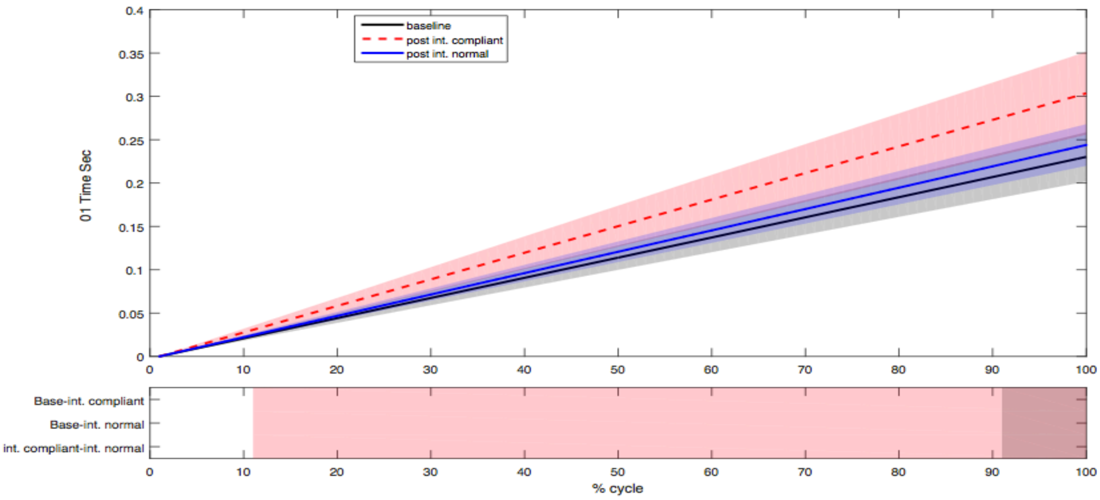

Figure 6.3 35: The effect of a 4-week treadmill accelerometer based biofeedback intervention contact time (seconds).

6.3.11 Treadmill kinematics

Analysis of treadmill data indicates that post intervention compliant running displayed a greater degree of knee flexion than baseline normal running for 59-68% of cycle ($T=2.86$, $p<0.016$, partial eta squared=1.28), and larger hip flexion for both 16-24% of cycle ($T=2.43$, $p<0.016$, partial eta squared=0.96) and 49-90% of cycle ($T=3.10$, $p<0.016$, partial eta squared=1.21).

When compared to post intervention normal running, post intervention compliant running displayed: (i) less dorsiflexed ankle for 6-25% of cycle ($T=4.2$, $p<0.016$, partial eta squared=1.39), (ii) larger knee flexion for 1-7% of cycle ($T=3.46$, $p<0.016$, partial eta squared=1.41) and 42-97% of cycle ($T=5.12$, $p<0.016$, partial eta squared=1.47), and (iii) larger hip flexion for 1-15% of cycle ($T=2.65$, $p<0.016$, partial eta squared=1.09), and 35-100% of cycle ($T=6.78$, $p<0.001$, partial eta squared=1.64). Comparison of baseline normal running and post intervention normal running indicates that post intervention normal running displayed a larger knee flexion angle for 59-76% of cycle ($T=2.64$, $p<0.016$, partial eta squared= 1.14), and larger hip flexion for both 15-23% of cycle ($T=2.47$, $p<0.016$, partial eta squared=0.96) and 70-90% of cycle ($T=3.11$, $p<0.016$, partial eta squared= 1.21).

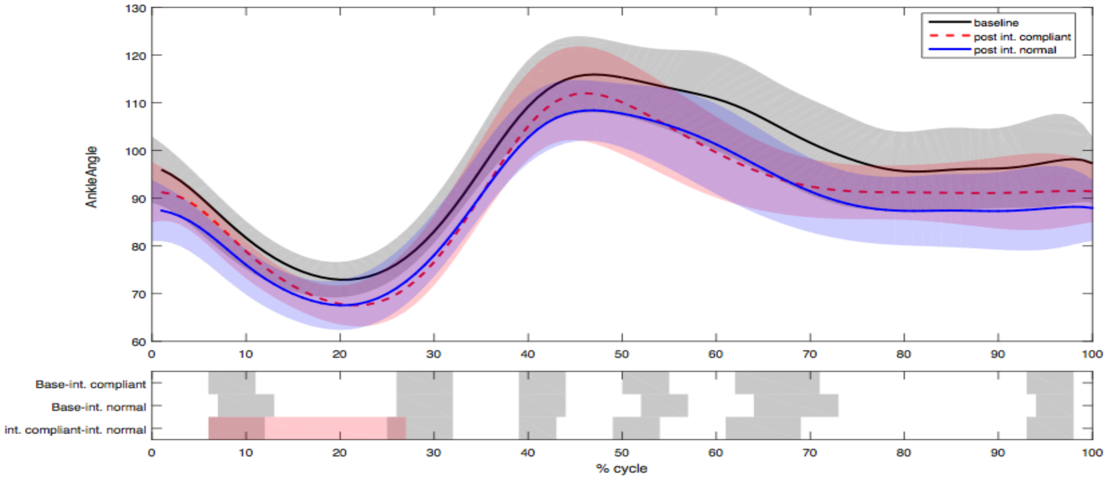

Figure 6.3 36: The effect of a 4-week treadmill accelerometer based biofeedback intervention on ankle angle (treadmill) (degrees)

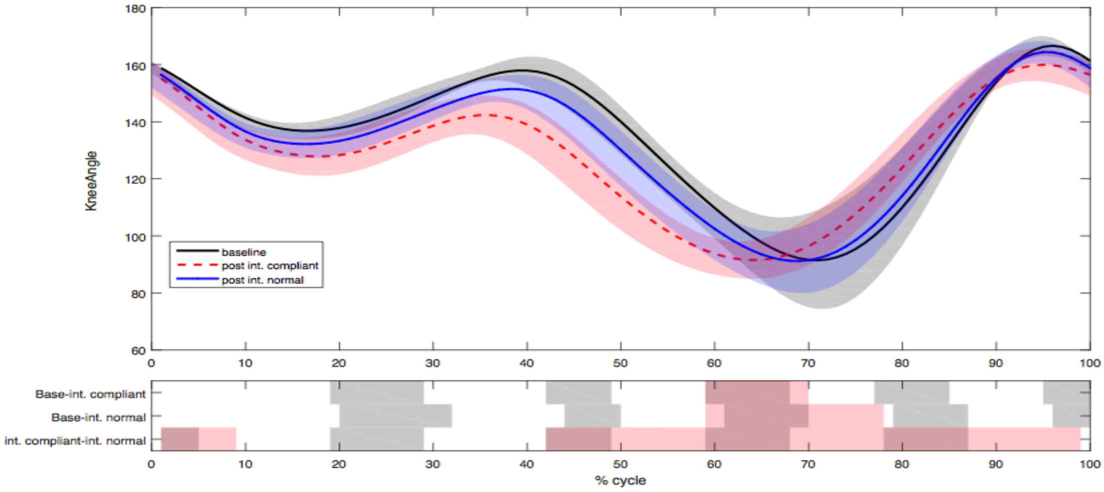

Figure 6.3 37: Figure 6.3 38: The effect of a 4-week treadmill accelerometer based biofeedback intervention on knee angle (treadmill) (degrees)

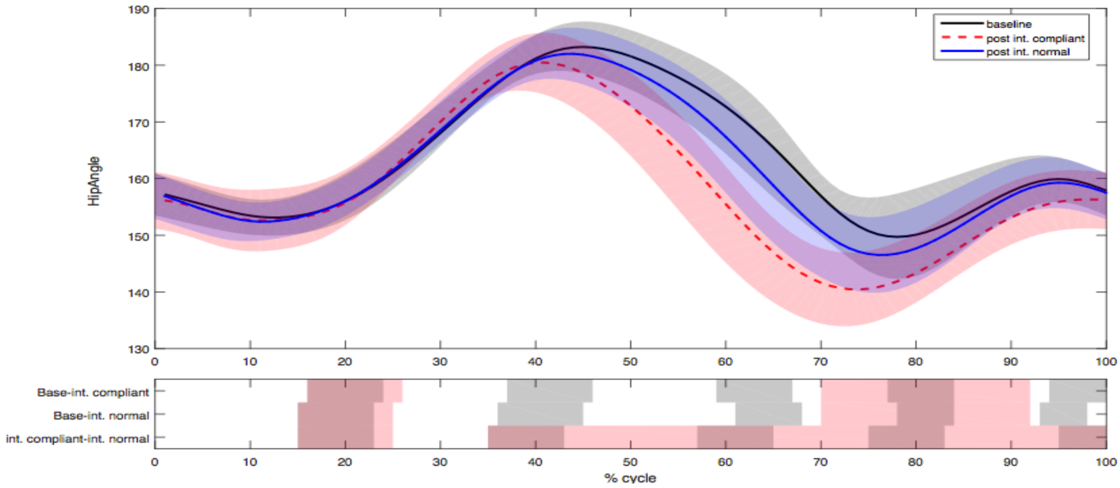

Figure 6.3 38: The effect of a 4-week treadmill accelerometer based biofeedback intervention on hip angle (treadmill) (degrees)

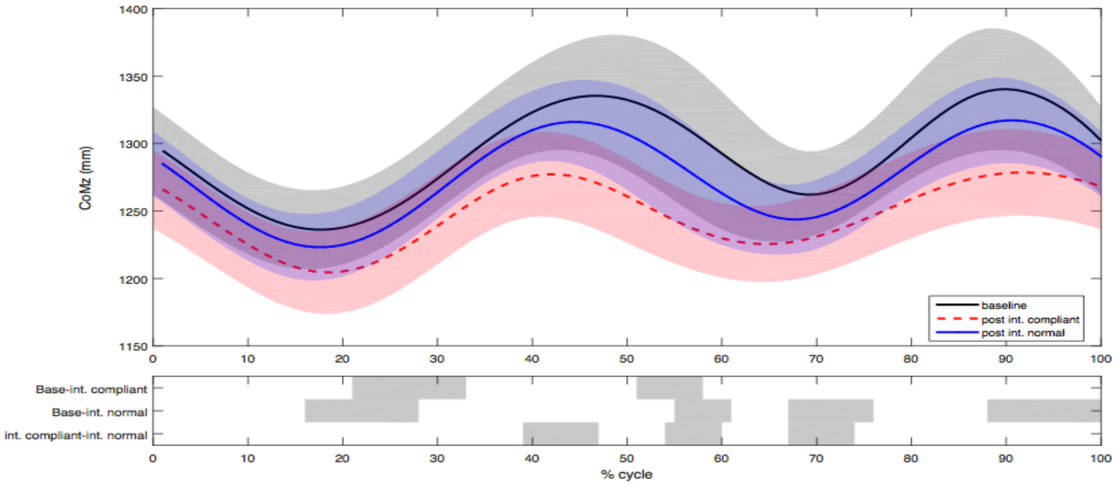

Figure 6.3 39 The effect of a 4-week treadmill accelerometer based biofeedback intervention on COM height (treadmill)(mm)

6.4 Discussion

6.4.1 Peak accelerations

One of the primary aims of this study was to examine the effect of a 4-week biofeedback-based intervention on peak tibial and sacral impact accelerations. With regard to this it is clear that the provision of treadmill-based biofeedback provided over a 4-week period has the ability to significantly reduce both peak tibial and sacral accelerations. A comparison at self-selected paces for each style (perhaps the most ecologically valid comparison) indicates that compliant running presents peak tibial and sacral acceleration values 40% and 42% smaller, relative to baseline values. At matched paces, similar results are present with reductions at the tibia of 44% and 40%, and sacrum of 38% and 36%, when matched to compliant and normal self-selected paces, respectively. Given that peak tibial acceleration values have been associated with the development of running related injuries and stress fractures prospectively (Davis et al., 2004, Davis et al., 2010) and tibial stress fractures retrospectively (Milner et al. 2006b, Pohl et al. 2008, Zifchock, Davis & Hamill 2006b, Milner, Hamill & Davis 2007), it appears the provision of treadmill biofeedback over a 4 –week period may be able to reduce the risk of injury development. Furthermore, prospective research has shown that in a 13 week period 14% of male and 15% of female recreational runners may sustain an overuse injury to the lower back/pelvic region (Taunton et al. 2003). Given the nature of injury and impact loading it is logical to assume that reductions in peak acceleration values experienced at the sacrum (for compliant running) may reduce the likelihood of developing such injuries. Sacral impact acceleration values have been suggested as a measure of impact experienced by the centre of mass (Henriksen et al. 2008), and thus potentially given an indication of whole body loading, and risk of injury development. Therefore, it is clear that compliant running induced by a 4-week intervention period of treadmill-based biofeedback may alter running mechanics to facilitate reduced loading, and consequently a reduced risk of impact related injury development.

Crowell et al (2011) completed a similar intervention over a 2-week period utilising a tibial accelerometer to provide real-time visual biofeedback and displayed a decrease in tibial impact acceleration values of 48% and 44% from pre-intervention measures to post-intervention, and pre-intervention to 1-month post-intervention measures respectively; thus presenting similar reductions to the current study. This is further supported by Cheung et al (2011) who completed the same biofeedback protocol as described above by Crowell et al (2011) and demonstrated a reduction in peak tibial acceleration values of 45%. However, Clansey et al (2014) employed a different tibial biofeedback protocol and demonstrated a reduction in tibial acceleration of 31% post-intervention and 22% at a 1-month follow up; thus the treadmill biofeedback protocol in the current study may be more successful with regard to reducing peak tibial accelerations. However, there may be a number possible explanations for this. Firstly, the larger amount of practice in the present study (3 sessions a week, running time increasing from 15-30 minutes, for 4 weeks) in comparison to that employed by Clansey et al (2 sessions a week, 20 minutes each, for 3-weeks) may partly explain the larger reductions. Volume of practice has been established as an important influencing factor in motor skill development and learning and thus the participants in the current thesis may have retained the compliant kinematics, induced by the intervention period, to a greater extent (Winstein 1991, Salmoni, Schmidt & Walter 1984, Porte et al. 2007). Furthermore, feedback in the current study was faded so that the total amount of biofeedback (after the 4th session) was gradually reduced every day. This type of feedback design is thought to be most effect at inducing motor learning of a new skill (Schmidt et al., 1989). The amount of feedback provided to each participant remained constant throughout the 3-week intervention period completed by Clansey et al (2014), and thus may explain the greater reductions in tibial acceleration present in the current study. This is somewhat confirmed by examination of the work completed by Crowell et al (2011) (as described above) who implemented a similar faded feedback design and found that participants were able to maintain reductions in peak tibial acceleration values to a greater extent (- 48% post intervention and – 44% at 1 month follow-up), than in Clansey et al's work (- 31% post intervention and -22% at 1 month follow up). Finally, it is possible that the continuous nature of biofeedback in the first 4 sessions of the intervention period in the current study may have facilitated a greater ability to experiment with various kinematic strategies, as participants were immediately able to determine if strategies were successful. In comparison, participants only received biofeedback every fifth stride in Clansey et al's study (2104). It therefore may have been more difficult to immediately determine the success of a change in kinematic strategy.

However, something that has not been considered in previous studies is the effect that the current intervention (this thesis) has on peak sacral accelerations. Examination of data presented in study 2 of this thesis indicates that tibial biofeedback was only able to reduce peak sacral accelerations by 6% following a 10-minute acute bout of biofeedback, in comparison to a 17% reduction displayed by treadmill biofeedback. Considering the similarity between the current intervention (this study) and that completed by Crowell et al (2010,2011), it is likely that the treadmill biofeedback employed in this study presents a unique advantage of reducing both tibial and sacral accelerations, in comparison to previously completed research, and thus may provide an additional protective capacity.

A secondary aim of the present study was to examine the effect of running speed on peak tibial and sacral accelerations. This is an important consideration as the probability of developing a tibial stress fracture from 3 miles of daily running has been shown to increase from 17%, to 19%, and to 33% as running speed increases from 2.5m/s, to 3.5m/s, to 4.5m/s (Edwards et al., 2009). Furthermore, increasing running speed from 3m/s to 5m/s has been shown to increase the passive vertical ground reaction force impact peak by 48% (Munro et al., 1987). Therefore, it appears that impact-loading increases as a result of increased running speed, and subsequently may increase the risk of injury development. Although this has never been reported for sacral accelerations, it was predicted a similar relationship would be present.

Results indicate that for each condition, both peak tibial and sacral impact accelerations increased with speed (as seen in table 6.3.8), and are thus in agreement with previous research (Mercer et al., 2002, Munro et al., 1987). Furthermore, as can be seen more clearly in table 6.3.8, each condition responded similarly to increasing speed at both the tibia and sacrum. However, compliant running demonstrates a larger number of significant changes to peak accelerations, as a result of increasing speed. This possibly indicates that it is more difficult to maintain compliant strategies at faster speeds, resulting in an increased sensitivity to changes in speed with regard to tibial and sacral accelerations. However, examination of the effect of speed on post intervention normal running indicates similar significant differences for tibial and sacral accelerations as post intervention compliant running (as can be seen in table 6.3.8). Given that the average speeds were larger for baseline normal running than at both post-intervention conditions (where the speeds were matched), it is possible that the observed differences with regards to the pattern of change in peak acceleration values are due to the varying paces and not running styles. Despite this, the most ecologically valid comparison remains to be of that between baseline normal and post intervention compliant (as they were determined by self-selected paces). Considering this, even with the apparent increased sensitivity of compliant running to speed (with regard to peak acceleration values), absolute peak acceleration values remain smaller for compliant running; therefore this increased sensitivity does not pose an increased risk of injury development.

6.4.2 Kinetics and kinematics

Consideration of kinetic variables further supports the idea of compliant running demonstrating a potential protective capacity. Vertical ground reaction force values were significantly lower for compliant running when compared to baseline normal running from 4-82% of stance (partial eta squared=1.43) and from 6-67% of stance when compared to post intervention normal running (partial eta squared= 1.46). With regard to both comparisons compliant running displays significantly lower values for the vertical (passive) impact peak and the active peak. The vertical impact peak has been associated with the development of running related injuries, tibial stress fractures and plantar fasciitis both in prospective research (Davis et al., 2010, Bowser et al., 2010, Davis, Milner & Hamill, 2004) and retrospectively (Zifchock, Davis & Hamill 2006a, Creaby, Dixon 2008, Pohl et al. 2008, Milner et al. 2006a); thus it appears that compliant running may offer a reduced risk of running injury development. Furthermore, reduction of the active vertical ground reaction force peak may subsequently reduce other variables that have been suggested to place the distal tibia under high loads and increased risk of stress fracture (as discussed in section 2.2.1.3)(Phuah et al. 2010, Sasimontonkul et al., 2007). A reduced vertical ground reactive force active peak may result in a subsequent decrease in joint reaction forces, causing a decrease in compression, shear forces, and tensile forces created by sagittal bending moments; all of which have been implicated in predisposing the distal tibia to high loads during running (Phuah et al. 2010, Sasimontonkul et al., 2007). This is supported by Sasimontonkul et al (2007) who suggests that the shear force experienced at the distal tibia during mid-stance is primarily due to the vertical ground reaction force active peak and by Leissring et al (2010) who demonstrated that 99% of knee joint reaction forces can be explained by the vertical ground reaction force active peak. Therefore, a reduction of the active peak appears to reduce numerous other factors that predispose the tibia to increased loading during mid-stance. Furthermore, vertical ground reaction force values are significantly smaller for the majority of stance phase (4-82%). Given that injury is due to relative excessive load, it seems logical to assume that this reduced loading throughout stance may reduce the risk of a load becoming relatively excessive and causing damage, at any point.

This reduction in vertical ground reaction force was reflected in reduced ankle flexor moments, knee flexor moments, knee abductor moments, and hip abductor moments for compliant running. Research indicates that prior to the development of patella femoral pain participants display knee abductor moments 65% larger than healthy counterparts and that participants with a history of patella femoral pain display knee abductor moments 31% greater than uninjured controls (Stefanyshyn et al. 2006, Stefanyshyn et al. 1999). Therefore, the significant reduction of knee abductor moments when running with a compliant technique, may be indicative of a protective capacity with regard to the development of patella femoral pain. Adoption of such a technique using a biofeedback system may also be an effective rehabilitative tool for participants with a history of patella femoral pain. Furthermore, increased knee flexor moments of 33% have been displayed in participants with a history of tibial stress fracture (Farley and Gonzalez, 1996). Therefore, the reduction in knee flexor moments, in the current study, during both the impact phase and mid-stance may reduce the risk of stress fracture development.

It has also been suggested that increased abductor moments at the knee and hip during mid-stance, and increased knee flexor moments, may contribute to an increase in bone bending moments at the distal femur, predisposing this location to increased risk of stress fracture while running (Edwards et al., 2008). Therefore, the reduced magnitude of both hip and knee abductor moments as well as knee flexor moments with compliant running may reduce the risk of developing femoral fractures.

Although very little evidence exists to directly implicate joint moments in the development of running related injuries, it is logical to assume that any reduction in loading will reduce the risk of injury development at that given point, and thus compliant running appears to display a reduced risk of injury at the ankle, knee, and hip.

It should be noted that compliant running displayed significantly larger knee flexor moments during late stance. However, given that these values are much smaller than peak knee flexor moments, occurring around 30-40% of stance, it is unlikely that these loads will become relatively excessive, and should therefore not pose an increased risk of injury development.

Compliant running displayed larger hip flexor moments compared to baseline normal running for 2-4% and 20-27% of stance. However, the former difference appears to be due to the fact that peak hip flexor moments are occurring earlier for compliant running than baseline normal running (10-15% of stance)(figure 6.3.39). Furthermore, hip flexor moments displayed for compliant running during 20-27% of stance are extremely small relative to peak values (600 N.mm versus 2000 N.mm). Therefore these differences may not demonstrate an increased risk of injury development.

As described in section 2.2.2, manipulation of running technique to reduce loads experienced by the body during running can be largely explained by the impulse momentum relationship. Subsequently, the kinematic strategies employed by participants following the treadmill biofeedback intervention in the present study clearly demonstrate this. Compliant running displayed a significantly larger amount of knee flexion for 100% of stance when compared to baseline measurements. Therefore, this increased knee flexion is likely to cause a reduced effective mass (reducing the mass element of the impulse momentum relationship) and subsequent reduction in force (Derrick 2004, Daoud, 2009, Devita and Skelly, 1992, Denoth 1986, Gerritson et al., 1995), thus potentially explaining the found decreases in both vertical ground reaction force and peak tibial and sacral accelerations. Similar results have been demonstrated by Lafortune et al (1996), whereby utilising a swinging pendulum device to mimic foot strike, increasing knee flexion from 0-40° was associated with a reduction in vertical ground reaction force impact peak of 30% (Lafortune et al., 1996) and by Potthast et al (2010) who demonstrated a 158N increase in vertical ground reaction force impact peak when knee flexion was increased from 0-40° at 30% of maximal voluntary contraction. However, similar research appears to indicate that an increase in knee flexion would subsequently result in an increase in tibial acceleration values (Lafortune et al., 1996, Potthast et al 2010, Derrick, 2004). In fact, Derrick (2004) suggests that for every 1° increase in knee flexion there would be an associated 0.27g increase in tibial acceleration due to a decreased effective mass (F=ma➔ a=F/m, therefore decreased mass would increase acceleration). This is further supported by Potthast et al (2010) who showed that tibial acceleration increased by 39%, 48%, and 46% as knee flexion increased from 0-40° at MVC values of 0%, 30%, and 60% respectively. However, the current study demonstrated a decrease in tibial acceleration values of 40-44% with an increase in knee flexion. This may indicate that in compliant running (versus a pendulum device or modelling), where the increase in flexion is less severe (12° increase at peak values), the associated reduction in force (57% decrease in vertical ground reaction force impact peak for compliant running) may be a more dominant factor than reduced effective mass, thus tibial accelerations decrease. This may also explain the larger reductions in tibial accelerations present in the current study, relative to study 1 (-10%) of this thesis, as study 1 demonstrated larger knee flexion angles at initial contact. Therefore, increased knee flexion in compliant running appears to reduce loading variables associated with injury development (peak accelerations, vertical ground reaction force impact peak etc), as described earlier in the discussion. Further support for a decreased risk of injury development as a result of increased knee flexion comes from research indicating that participants suffering from pre-osteoarthritic knee pain display 300% less knee flexion (more extension) during the impact phase and 85% less knee flexion during stance, than uninjured controls (Radin et al., 1991).

Hip flexion was also significantly larger for compliant running when compared to both baseline and post intervention normal running during early stance. Although, there appears to be no current research linking increased hip flexion to reduced loading in running, it is likely that that the increased flexion of the hip acts to further divide the body into separate segments, thus reducing impact loading via a reduction of effective mass, as described above in relation to knee flexion (Derrick, 2004).

Increasing the time over which a force is applied and decreasing the change in velocity will also act to decrease the magnitude of a force, according to the impulse momentum relationship. Compliant running displayed a significantly larger contact time relative to both baseline normal running, and post intervention normal running for the majority of stance, resulting in an average total contact time for compliant running of 0.28 seconds versus 0.20 seconds for baseline normal running. McMahon et al (1989) demonstrated that increasing contact time via increased knee flexion (a running style referred to as Groucho running) was associated with a subsequent reduction in vertical touchdown velocity (0.5-0.8m/s in normal running versus almost 0m/s in Groucho running) due to a reduced vertical oscillation of COM. Therefore the magnitude of force experienced when running with a more compliant style may also be reduced via reduction of velocity prior to initial contact. This is supported by research indicating that vertical ground reaction force impact peaks increase by 36% as landing velocity increases from 0.56m/s to 1.36m/s (Zadpoor et al., 2007). It is therefore clear that treadmill biofeedback alters kinematics to increase cushioning during impact. However, participants also appear to demonstrate increased knee flexion and a lower COM during mid-to-late stance. This, in conjunction with the reduced vertical ground reaction force active peak, is indicative of a reduced vertical propulsive force and subsequently a reduced maximum COM height. A reduced fall height may subsequently lead to a reduced vertical velocity prior to contact, which has previously been associated with decreased impact loading (Liu and Nigg, 2000,Zadpoort et al., 2007, McMahon et al.,1987).

Examination of treadmill kinematic results support the findings for over-ground compliant running, indicating that participants displayed increased knee and hip flexion during the both the impact phase and swing phase. This may indicate that participants maintained increased knee and hip flexion during the swing phase in an effort to maintain a lower COM, thus reducing fall height, which has been associated with reduced vertical landing velocity (McMahon et al, 1989) and subsequent reduced impact force (McMahon et al, 1989, Liu and Nigg, 2000, Zadpoor et al, 2007). COM height did not demonstrate significant change however did trend towards lower values (as observed in figure 6.3.40). This may be due to high variation displayed for kinematics in treadmill running in comparison to over-ground running (Nigg et al., 1995).

In summary, it appears that reduction of impact loading results from 2 mechanisms. Firstly, an increase in system compliance at initial contact (as evident by increased knee and hip flexion), resulting in a decrease in effective mass and a subsequent reduction in force, and secondly, a reduction in vertical propulsion (evident by reduced vGRF active peak), decreasing maximum COM height, consequently reducing vertical landing velocity, and impact loading.

6.4.3 Energy expenditure

Examination of energy expenditure data indicates that COL, COT, and RE of compliant running do not differ significantly to either baseline normal running or post intervention normal running. In fact, the only significant difference occurred between baseline normal running and post intervention normal running at self-selected paces and this is likely due to the increased running speed at baseline measures relative to post intervention normal running (9.44km/hr Vs. 8.78 km/hr).

These findings are somewhat surprising given the association between more compliant kinematics and increased energy expenditure (as observed in study 1), largely due to a diminished capacity to maximise the enhancements associated with effective use of the SSC (McMahon et al., 1987, Derrick et al., 2000, Bonacci et al. 2009, Divert et al. 2005, Spurrs et al., 2003)(described in section 2.3.1.6). Therefore, it would seem logical that the increased contact time, knee flexion, and hip flexion displayed for compliant running would significantly increase energy expenditure. This was observed by McMahon et al (1989) who demonstrated that as knee angle decreased from 70° to 60° (thus becoming more flexed) energy expenditure increased by 50%, which was also associated with an increased contact time (values not displayed). However, research indicates that the magnitude of the vertical ground reaction force active peak gives an indication of the overall muscular contribution during ground contact, and thus plays a large role in determining the energetic cost of running (Taylor et al. 1980, Farley, McMahon 1992, Kram, Taylor 1990b). In fact, vertical ground reaction force impulse (area under the ground reaction force curve) has been shown to explain 32% of the variance in energy expenditure between participants, with the more economical runners displaying smaller values (Heise and Martin, 2001). Farley and

McMahon (1992) demonstrated that runners reduced energy expenditure by 25% as a result of a 25% decrease in vertical ground reaction force (active peak). Therefore, considering that compliant running in the present study significantly decreased the magnitude of vertical ground reaction force for the majority of stance (represented by a 56% decrease in the active peak relative to baseline, and a 47% decrease relative to post intervention normal running) there may be a balance whereby the more compliant kinematics (increased knee and hip flexion and increased contact time) decrease efficiency, via a diminished capacity to maximise SSC mechanisms, but the decreased vertical ground reaction force active peak increases efficiency. This is supported by Clansey et al (2014) who demonstrated no change in RE following a 3-week tibial-accelerometer biofeedback intervention that resulted in reduced peak tibial accelerations of 31% (similar to the current study).

Consideration of study 1 where an increase in energy expenditure (21%) was associated with similar compliant kinematics as the present study may have numerous explanations. Firstly, the larger knee and hip flexion angles present in study 1 at initial contact (knee: +5°; hip: +12°) and toe-off (knee: +20°; hip +33°) in comparison to the present study, may have required increased muscular recruitment to maintain the more crouched position, and thus increased energy expenditure (McMahon et al., 1987). In addition, the longer intervention period and larger amount of practice (4 weeks, 3 sessions a week, running time increasing from 15-30mins) in the present study, in comparison to study 1 (3 weeks, 1 session a week, 10-15mins), may have facilitated improved adaptation to the more compliant kinematics. However, the use of convenience based sampling in study 1 and 4, as well as the high variability of impact acceleration data in study 1 (which may be indicative of a varied response to the verbal feedback), may indicate that the subsequent conclusions with regard to energy expenditure may not be reflective of wider populations.

Finally, RE increased with increased running speed at each time point (baseline normal, post intervention compliant, and post intervention normal) as expected based on previous literature (Kyrolainen et al., 2001).

6.5 Conclusion

It is clear that a 4-week treadmill-based biofeedback intervention directs participants to adopt a kinematic strategy that may offer increased protection from running related injuries. This is evident by the reduction of peak tibial and sacral accelerations, vertical ground reactions forces, and joint moments, all of which have been associated with running injury (Davis et al, 2005, Zifchock et al., 2006, Creaby and Dixon 2008, Pohl et al. 2008, Milner et al. 2006a, Stefanyshyn et al., 2006, Stefanyshyn et al., 1999, Phuah et al., 2011, Sasimontonkul et al., 2007). Given the relatively low financial cost of such a biofeedback system this may have major implications in both clinical and general fitness/health related industries for rehabilitative and prehabilitative purposes. Kinematic and kinetic data indicates that reductions in impact loading stem from increased cushioning at impact, and a reduction in the vertical ground reaction force active peak, reducing vertical displacement of the COM. Finally, strategies adopted by participants to minimise loading did not affect energy expenditure.

6.6 limitations

- Equipment failure resulted in not being able to measure the magnitude of treadmill acceleration reduction.
- Participants within this study were healthy and may therefore not demonstrate the same response to the treadmill biofeedback intervention as at risk or injured participants.
- Convenience based sampling was employed for recruitment of participants and therefore limits generalizability to wider populations.
- Post intervention testing days were not randomized and therefore may limit ecological validity.

6.7 Future recommendations

- A longer retention test should be employed to determine motor learning (1-month post-intervention or larger).
- Future research employing such a biofeedback system within clinical populations (e.g. obese, osteoarthritic etc) may yield interest information with regard to the potential clinical application of this technology.

Measurement of energy expenditure variables and impact accelerations in out-door running (following biofeedback) may yield more ecological findings.

Chapter 7:Summary and conclusion

The primary aim of this thesis was to examine if a compliant running technique reduces peak impact accelerations, and what the associated kinematics and kinetics of this style are. Furthermore, this thesis sought to determine what method should be employed in the teaching of runners to adopt such a technique. With regard to both of these aims, the four studies undertaken in this thesis provide a clear consensus.

Gait re-training can clearly alter running kinematics to reduce impact loading, however the magnitude of reduction appears to be heavily influenced by the mechanism through which kinematic alterations are directed. In this regard, visual accelerometer based biofeedback appears to demonstrate a greater ability to facilitate adoption of more compliant kinematics than verbal feedback. More specifically, visual biofeedback, utilising treadmill acceleration as the input, appears to display a unique ability to reduce both tibial (-26%) and sacral (-17%) accelerations acutely (following 10 minutes of biofeedback), with reductions increasing further when the intervention period is extended to 4-weeks (tibia:-40%; sacrum:-42%). In comparison, an acute and a 3 week bout of verbal feedback produced lower reductions in impact accelerations (tibia: -8%, sacrum: -22%; tibia: -10%, sacrum: -41%;); while targeted segment based biofeedback produced large reductions but more localised to their source of feedback: sacral biofeedback (tibia: -1%; sacrum: -27%), and tibial biofeedback (tibia: -39%; sacrum: -4%). The above reductions in tibial and sacral accelerations for treadmill-based biofeedback were accompanied by significant reductions in vertical ground reaction force, and joint moments, therefore suggesting a potential protective capacity given the association with each kinetic measure to injury (Davis et al., 2010, Bowser et al., 2010, Davis et al., 2004, Zifchock, Davis & Hamill 2006, Creaby, Dixon 2008, Pohl et al. 2008, Milner et al. 2006, Stefanyshyn et al. 2006, Stefanyshyn et al. 1999, Farley and Gonzalez, 1996, Edwards et al., 2008).

In order to facilitate the reductions in impact loading participants' adopted a technique that appears to reduce impact loading via 2 main mechanisms. Firstly, an increase in cushioning at initial contact (as evident by increased knee and hip flexion), results in a decrease in effective mass and a subsequent reduction in force. Secondly, a reduction in vertical propulsion (evident by reduced vGRF active peak), decreasing maximum COM height, consequently reducing vertical landing velocity, and impact loading.

A secondary aim of this thesis was to examine the affect of compliant running on energy expenditure. Study 4 showed that the kinematic strategy associated with the compliant running style adopted following treadmill-based biofeedback appeared not to affect running economy, cost of transport, or cost of locomotion. In comparison, study 1 demonstrated that verbally directed compliant running displayed larger energy expenditure (+21%) relative to normal running. This may be due to the relatively larger degree of knee and hip flexion at initial contact (knee: +5°; hip: +12°) and toe-off (knee: +20°; hip +33°) displayed by participants in study 1 in comparison to study 4, resulting in increased muscular recruitment to maintain the more crouched position in study 1 (McMahon et al, 1989).

In conclusion, verbal and technology-based gait retraining methods demonstrate an ability to reduce impact loading, and thus potentially reduce the risk of injury development. However, treadmill accelerometer based biofeedback appears to display the unique benefit of decreasing segmental accelerations both at the tibia and sacrum, as well as whole body loads; all of which have been implicated in injury development (Bowser, Davis, 2010, Davis et al, 2004,2010, Milner et al, 2006 Zifchock et al, 2006). Given the relatively low cost of such a system, this may have huge potential for use in both clinical and health/fitness settings as a prehabilitative and rehabilitative tool. This reduced loading is facilitated both by increased cushioning at impact, and by reducing vertical oscillation of the COM. In addition, these kinematic changes appear not to affect energy expenditure.

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Appendices

Appendices 1:General health questionnaire used in each study

School Of Health and Human Performance
Dublin City University

General Health Questionnaire

Name:……………………………… Occupation:……………………………

Address:……………………………………………………………………………

Telephone: (Home)………………… (Work):………………………………

Do you have, or have you ever suffered from: -Diabetes? Yes / No

-Asthma? Yes / No

-Epilepsy? Yes / No

Have you ever had pains in your chest or heart? Yes / No

Do you ever feel faint or have spells of dizziness? Yes / No

Do you have or have you ever had high blood pressure? Yes / No

Do you have a muscle, back or joint problem that could be aggravated by physical activity or made worse with exercise? Yes / No

Do you have any current injuries? Yes / No

In the past week, have you suffered from any illness, which required you to be in bed
or off work for one day or more? Yes / No

Do you smoke? If yes, how many per day?
Yes / No

Do you drink? If yes, how many units per week? Yes / No

Is there a good physical reason not mentioned here why you should not carry out laboratory testing? Yes / No

Please provide any further information concerning any condition/complaints which you suffer from and any medication which you may be taking by prescription or otherwise:

……………………………………………………………………………………………………

……………………………………………………………………………………………………

……………………………………………………………………………………………………

……………………………………………………………………………………………………

……………………………………………………………………………………………………

Date: Signature

Appendices 2: PAR-Q used for each study

Physical Activity Readiness Questionnaire (PAR-Q)

(representative of the American College of Sports Medicine standards)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly.

No	Yes	
☐	☐	1. Has your doctor ever said that you have a heart condition <u>and</u> that you should only do physical activity recommended by a doctor?
☐	☐	2. Do you feel pain in your chest when you do physical activity?
☐	☐	3. In the past month, have you had chest pain when you were not doing physical activity?
☐	☐	4. Do you lose your balance because of dizziness or do you ever lose consciousness?
☐	☐	5. Do you have a bone or joint problem that could be made worse by a change in your physical activity?
☐	☐	6. Is your doctor currently prescribing drugs (for example water pills) for your blood pressure or heart condition?
☐	☐	7. Do you know of <u>any other reason</u> why you should not do physical activity?

Please note: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

If you answered YES to one or more questions

Talk to your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want as long as you start slowly and build up gradually. Or you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

If you answered NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- Start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- Take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively.

Delay becoming much more active if:

- You are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better, or
- If you are or may be pregnant – talk to your doctor before you start becoming more active.

Print and sign form. Then submit with Patron Application to the Foy Fitness and Recreation Center service desk.

Print Form

Reset Form

I understand my signature signifies that I have read and understand all the information on the questionnaire, that I have truthfully answered all the questions, and that any questions/concerns I may have had have been addressed to my complete satisfaction.

Name (please print) ____________________

Signature ____________________ Date ____________

Appendices 3: Study 1 informed consent

DUBLIN CITY UNIVERSITY

Informed Consent Form

Research Study Title

The effect of Groucho running on impact accelerations and energy consumption

Working Title: The effect of Groucho running on the forces that travel throughout the body as a result of the foot striking the ground when running and the amount of energy used when using this style.

Clarification of the Purpose of the Research

When running forces called impact accelerations travel in a wave throughout the body as a result of the foot striking the ground. It is thought that these impact accelerations cause injury when excessive. Groucho running has been suggested as a running technique that may increase the body's ability to reduce these forces, thus reducing likelihood of injury. However the Groucho style may increase the amount of energy consumed when running. It has been shown that when fatigued the size of these forces travelling throughout the body increases when running. This study therefore aims to investigate the effect of Groucho Running on these impact accelerations in both fatigued and unfatigued conditions as well as the effect it may have on energy consumption with the overall goal of understanding the benefits that this running style may have as a injury preventive/rehabilitative tool.

III. Confirmation of particular requirements as highlighted in the Plain Language Statement

Participant – please complete the following (Circle Yes or No for each question)
Have you read or had read to you the Plain Language Statement
Yes/No
Do you understand the information provided?
Yes/No
Have you had an opportunity to ask questions and discuss this study?
Yes/No
Have you received satisfactory answers to all your questions?
Yes/No

IV. Confirmation that involvement in the Research Study is voluntary

Involvement within this research project is purely voluntary. Participants wishing to withdraw from the study at any stage throughout are entitled to do so. There will be no penalty enforced on any subjects wishing to quit the Research Study prior to all stages being completed.

V. Advice as to Arrangements to be made to Protect Confidentiality of Data, Including that Confidentiality of Information Provided is Subject to Legal Limitations.

Confidentiality is an important issue during data collection. Participant's identity, or other personal information will not be revealed or published. Subjects will be assigned an ID number under which all personal information will be stored in a secure file and saved in a password protected file in a computer at DCU. The investigators alone will have access to the data. However, confidentiality of information provided can only be protected within the limitations of the law. It is possible for data to be subject to subpoena, freedom of information claim or mandated reporting by some profession.

VI. Any other Relevant Information

Any participants who share a relationship of any kind with any of the researchers, relevant to this study, will not be affected in any way with regard to their ongoing studies throughout the entire duration of this study.

VII. Signature:

I have read and understood the information in this form. The researchers have answered my questions and concerns, and I have a copy of this consent form. Therefore, I consent to take part in this research project

Participants Signature: ______________________________

Name in Block Capitals: ______________________________

Witness: ______________________________

Date: ______________________________

Appendices 4: Study 1 plain language statement

DUBLIN CITY UNIVERSITY

Plain Language Statement

Introduction to the research study

Working Title: This study will examine the effect of the Groucho running style on impact forces that travel throughout the body as a result of the foot striking the ground when running and the amount of energy expended when using this style in comparison to that of normal runnning. It will also investigate the effect of fatigue in relation to these impact forces when running.

The study will be undertaken at the School of Health and Human Performance in DCU. The principal investigator is Dr. Kieran Moran who may be contacted at 017008011 (phone) or Kieran.moran@dcu.ie .

The second investigator is Mr. Ciarán Ó Catháin who may be contacted at 0872745184 (phone) or ciaran.ocathain2@mail.dcu.ie .

Details of what involvement in the Research Study will require

Participants will be required to attend the School of Health and Human Performance a total of five times. This involves three training sessions where the participants will be instructed on how to run using the Groucho technique on a treadmill, two sub-maximal running economy tests where physiological responses to exercise will be measured, and 2 more treadmill based tests where the size of the forces that travel throughout the body as a result of making foot contact with the ground during running will be measured.

The Groucho instruction sessions will take place over a two week period where each participant will attend the School of Health and Human Performance laboratories for no longer than 30 minutes at a time. The participants must agree to practice the Groucho running style at least three times outside these monitored practice sessions.

Running economy, run to fatigue and motion analysis using both Groucho and normal running will be carried out over two days; all tests completed both days with one day assigned to Groucho running and the other to normal running. During the running economy test physiological responses to exercise will be measured such as heart rate and rate of perceived exertion.

For the fatigue test participants will be asked to run at a self selected pace deemed ‘normal pace’ and the participants will continue to run until fatigue criteria is met. During this test the size of the waves of force that travel throughout the body as a result of foot contact with the ground during running will be measured using a device called an accelerometer. Accelerometers will be attached at the tibia, lower back, and head.

Motion analysis will involve placement of 23 sensors on different landmarks around the body and 5-6 15 meter runs in the biomechanics lab.

Testing should take no longer than 2 hours each day.

Potential Risks to Participants from Involvement in the Research Study (if greater than that encountered in everyday life)

There are no added potential risks to participants from involvement in the research study than would be encountered in everyday life. In order to take part in the study participants must regularly take part in running activities (at least 3 times a week) and will therefore be used to the amount of running required for completion of the testing procedure. As a result the study should have no negative risk effects.

Benefits (Direct or Indirect) to Participants from Involvement in the Research Study

Following completion of the study participants will have been taught to run using the Groucho Method. They will also know the effect this method has on reducing likelihood of injury and thus may choose to use the running style as a future injury preventive or rehabilitative tool.

Advice as to Arrangements to be made to Protect Confidentiality of Data, Including that Confidentiality of Information Provided is subject to Legal Limitations

Confidentiality is an important issue during data collection. Participant’s identity, or other personal information, will not be revealed or published. Subjects will be assigned an ID number under which all personal information will be stored in a secure file and saved in password protected file in a computer at DCU. The investigators alone will have access to the data.

Advice as to whether or not data is to be destroyed after a minimum period

Data will be stored for twelve months following the completion of the project, in line with University regulations for examinations. The principal investigator will destroy this data.

Statement that Involvement in the Research Study is Voluntary

Involvement within this research project is purely voluntary. Participants wishing to withdraw from the study at any stage throughout are entitled to do so. There will be no penalty enforced on any subjects wishing to quit the Research Study prior to all stages being completed.

If participants have any concerns about this study and wish to contact an independent person, please contact:

The Secretary, Dublin City University Research Ethics Committee, c/o Office of the Vice-President for Research, Dublin City University, Dublin 9. Tel: 017008000

Appendices 5: Study 2 informed consent

Informed Consent

Title: The use of biofeedback from an accelerometer to alter impact loading during treadmill Running**.**

Institution: Dublin City University the School of Health and Human Performance
Principal Investigator: Dr.Kieran Moran **Other Investigators:** Ciaran O' Cathain, Karen Brady

I. Clarification of the purpose of the research

The study will attempt to investigate any changes that occur in impact loads during treadmill running as a result of the biofeedback being utilized. Biofeedback in relation to the impacts at the tibia, sacrum, treadmill and wrist will all be investigated. The aim of the study is examine the most effective form of biofeedback at reducing impact loads during treadmill running.

III. Confirmation of particular requirements as highlighted in the Plain Language Statement

I will be asked to visit Dublin City University on one occasion whereby I will be required to attach accelerometers (small portable devices that measure accelerations) to my tibia, sacrum and wrist. I will be required to run continuously for a total of 20 minutes at a pace of 10km/hour. This will be broken up into a 5 minute baseline measurement, 10 minute biofeedback period which will then be followed by a 5 minute period of no biofeedback. During the biofeedback period I will be asked to look at a monitor that is located directly in front of the treadmill. This monitor is displaying my impact loads in relation to the sacrum, tibia, treadmill or wrist. I will be required during this biofeedback period to adapt my running technique in order to get my impact peaks which are displayed in blue on the monitor to stay below the threshold that has been set which is displayed as a yellow/green line.

.

iv) Confirmation of particular requirements as highlighted in the Plain Language Statement

Participant – please complete the following (Circle Yes or No for each question)
I have read the Plain Language Statement (or had it read to me)
Yes/No
I understand the information provided
Yes/No
I have had an opportunity to ask questions and discuss this study
Yes/No
I have received satisfactory answers to all my questions
Yes/No

v) Confirmation that involvement in the study is voluntary

-Involvement within this research project is purely voluntary. Participants wishing to withdraw from the study at any stage throughout are entitled to do so. There will be no penalty enforced on any subjects wishing to quit the Research Study prior to all stages being completed.

vi) Advice as to arrangements to be made to protect confidentiality of data, including that confidentiality of information provided is subject to legal limitations

Confidentiality is an important issue during data collection. Participant's identity or other personal information will not be revealed or published. Subjects will be assigned an ID number under which all personal information will be stored in a secure file and saved in a password protected file in a computer at DCU. The investigators alone will have access to the data. However, confidentiality of information provided can only be protected within the limitations of the law. It is possible for data to be subject to subpoena, freedom of information claim or mandated reporting by some profession.
Data will be stored for twelve months following the completion of the project, in line with University regulations for examinations. The principal investigator will destroy this data.

vii) Any other relevant information

If participants have any concerns about this study and wish to contact an independent person, please contact:
The Secretary, Dublin City University Research Ethics Committee, c/o Office of the Vice-President for Research, Dublin City University, Dublin 9. Tel: 017008000

viii) Signature:

I have read and understood the information in this form. The researchers have answered my questions and concerns, and I have a copy of this consent form. Therefore, I consent to take part in this research project

Participants Signature: ______________________________

Name in Block Capitals: ______________________________

Witness: ______________________________

Date: ______________________________

Appendices 6: Study 3 debrief form

Debrief Form

- Thank you for participating in this study.

- Please return 48 hours later from now for your second testing session. Please consume similar food and fluids within the same time frame before testing that you did today before your next testing session.
- Please bring the same footwear you wore today for testing session 2.

- Over the days break before testing session 2 you will be emailed a concise information pack which will contain a set of instructions on a new running style and a video clip of this running style being performed. Please familiarise yourself with this information pack so that you can replicate this running style in testing session 2.

- Please refrain from engaging in any running (including practicing instructions in the information pack) and heavy exercise during this days break as this could affect our results.

Appendices 7: Pre-participation form (study 3)

Pre-participation Testing Session Two Questionnaire

Subject I.D:	Circle Yes/No if applicable
1) Have you engaged in any running since testing session 1?	Yes/No
2) Did you receive the information pack email?	Yes/No
3) Did you understand the contents of this email?	Yes/No
4) Did you familiarise yourself with the contents of this email?	Yes/No
5) Are you wearing the same footwear that you were wearing at testing session 1?	Yes/No

If yes was answered to any of the questions above please specify with the item number in question beside the response:

Appendices 8: information pack for study 3 and 4

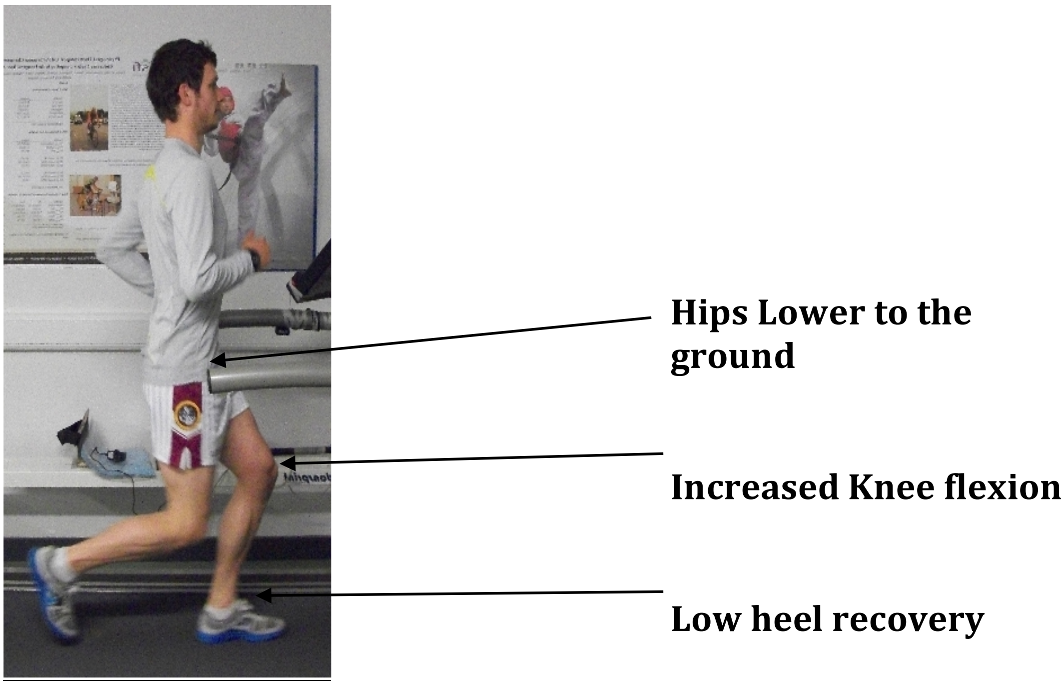
*Note: was slightly adapted to fulfil the needs of each study i.e. verbal feedback information was removed for the biofeedback group in study 3 and all for all participants in study 4 and the biofeedback information was removed for the verbal feedback group in study 3.

Compliant Running Information It has suggested that the below running technique can reduce impact loading experienced by the body.

- You will be asked to perform the below running technique at the next testing session
- You will then receive verbal feedback or tibial biofeedback on your performance of this technique by an expert in this running style.

Running technique guidelines

- **Lower hip position** while running
- **Hips should travel in a straight line** with no up and down movement
- Keep **feet close to the ground** throughout each stride
- See Example of running posture below.



- **See attached video (please mute)**

Compliant Running Information and Biofeedback Leaflet

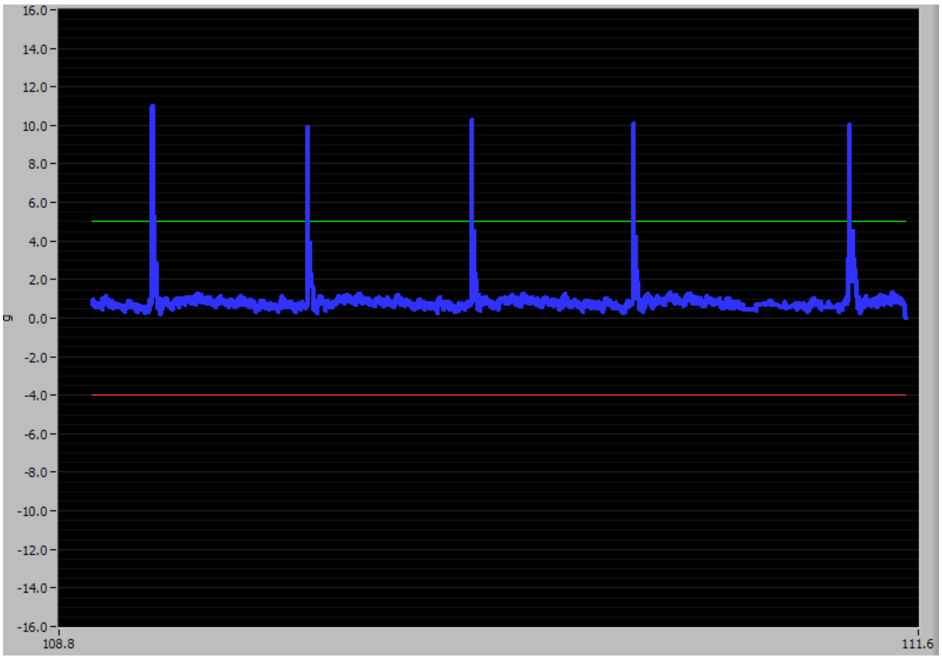
- The idea of the provided biofeedback system is to present the user with information that may aid in the alteration of their gait, with the goal of reducing load on the body, that can result in the development of injury.

- The system uses visual Biofeedback that allows the user to track the change in magnitude of impact peaks as they run, with the goal of reducing the size of the peaks as they progress through each session. A green line is set at a certain level on the graph that tracks changes in peak impact while running. This line represents 50% of the users peak acceleration. See an example of this graph below.

- The user should attempt to alter their running style to reduce their impact peaks below the green line as much as possible. The visual biofeedback will give the user information on what technique changes they make may be effective.

The user should attempt to reduce their peak impacts as much as possible while maintaining a comfortable movement pattern.

- Techniques that have been suggested to work can be seen below.



Visual Biofeedback screen

Appendices 9: informed consent (study 4)

Informed Consent Form

Introduction to the Research Study

i) **Research Study Title:** This research will examine the effects of 5 weeks of accelerometer-based biofeedback on impact loading, running mechanics, and energy expenditure.

-This study will be undertaken in the School of Health and Human Performance at DCU. The principal investigator is Dr. Kieran Moran who may be contacted at 017008011 (phone) or Kieran.moran@dcu.ie.
-Thep investigator in the study is Mr. Ciaran O Cathain who is currently studying for his PhD.

ii) **Clarification of the purpose of this study?**

-The purpose iof this study is to examine if a 5 week accelerometer based biofeedback running intervention can alter mechanics to reduce loading and subsequently risk of injury development. A secondary goal is to examine the effect this intervention has on running expenditure.

iii) **What will the study involve?**

The study will involve participants attending a gait-retraining session 3 times a week for 5 weeks. These sessions will incrementally increase from 15minutes of running to 30 minutes across the 5 weeks. Prior to the commencement of this there will be one day of baseline testing. At the end of the five-week period participants will return for two final testing sessions.

ix) **Confirmation of particular requirements as highlighted in the Plain Language Statement**

Participant – please complete the following (Circle Yes or No for each question)
I have read the Plain Language Statement (or had it read to me)
Yes/No
I understand the information provided
Yes/No
I have had an opportunity to ask questions and discuss this study
Yes/No
I have received satisfactory answers to all my questions
Yes/No

x) **Confirmation that involvement in the study is voluntary**

-Involvement within this research project is purely voluntary. Participants wishing to withdraw from the study at any stage throughout are entitled to do so. There will be no penalty enforced on any participants wishing to quit the Research Study prior to all stages being completed.

xi) Advice as to arrangements to be made to protect confidentiality of data, including that confidentiality of information provided is subject to legal limitations

Confidentiality is an important issue during data collection. Participant's identity or other personal information will not be revealed or published. Participants will be assigned an ID number under which all personal information will be stored in a secure file and saved in a password protected file in a computer at DCU. The investigators alone will have access to the data. However, confidentiality of information provided can only be protected within the limitations of the law. It is possible for data to be subject to subpoena, freedom of information claim or mandated reporting by some profession.
Data will be stored for twelve months following the completion of the project, in line with University regulations for examinations. The principal investigator will destroy this data.

xii) Any other relevant information

If participants have any concerns about this study and wish to contact an independent person, please contact:
The Secretary, Dublin City University Research Ethics Committee, c/o Office of the Vice-President for Research, Dublin City University, Dublin 9. Tel: 017008000

xiii) Signature:

I have read and understood the information in this form. The researchers have answered my questions and concerns, and I have a copy of this consent form. Therefore, I consent to take part in this research project

Participants Signature: ______________________________

Name in Block Capitals: ______________________________

Witness: ______________________________

Date: ______________________________

Appendices 10: Data collection sheet (study 4)

Study 4 data collection sheet:

Name:
Email:
Phone:

Shoes:
Time:

Anthropometric data baseline

Height	
Mass	
Leg Length	
Ankle width	
Knee Width	

Vicon baseline **(zero force plate after every trial)**

Self select pace	
Run 1	
Run 2	
Run 3	
Run 4	
Run 5	
Run 6	
Run 7	

Energy expenditure baseline **(zero force plate)**

Speed	**Heart Rate**	**RPE**	**ACC (last 15 secs)**

Appendices 11: Intervention period data collection (study 4)

Study 4 data collection: Week 1

Week 1	RPE	Naturalness/ Comfort	Comments (Pain/soreness)
Day 1 15mins Continuous Biofeedback			
Day 2 17mins Continuous Biofeedback			
Day 3 19mins Continuous Biofeedback			

Study 4 data collection: Week 2

Week 2	RPE	Naturalness/ Comfort	Comments (Pain/soreness)
Day 4 21mins Continuous Biofeedback			
Day 5 23mins Biofeedback: 5mins at start, 5mins at 9 min mark, 5mins at 18 min mark			
Day 6 25mins Biofeedback: 4mins at start, 4mins at 10mins 30secs, 4mins at 21mins			

Study 4 data collection: Week 3

Week 3	RPE	Naturalness/ Comfort	Comments (Pain/soreness)
Day 7 27mins Biofeedback: 3mins at start, 3mins at 12min mark, 3mins at 24min			
Day 8 29mins Biofeedback: 2mins at start, 2mins at 13min30secs mark, 2mins at 27min mark			
Day 9 30mins Biofeedback: 2mins at start, 1mins at 14mins 30secs, 2mins at 28mins			

Study 4 data collection: Week 4

Week 4	RPE	Naturalness/ Comfort	Comments (Pain/soreness)
<u>Day 10</u> 30mins Biofeedback: 1.5mins at start, 1min at 14min30 mark, 1.5 mins at 28.5 min			
<u>Day 11</u> 30mins Biofeedback: 1mins at start, 1mins at 14min30secs mark, 1mins at 29min mark			
<u>Day 12</u> 30mins Biofeedback: 45secs at start, 30 sec 14mins 45secs, 45secs at 29mins15secs			