

# Player load monitoring: Protecting the elite player from overload using miniature high frequency triaxial accelerometers

UEFA priority topic: Footballer's optimum load (medical sciences)

## **FINAL REPORT**

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## **1.0 Executive Summary**

Low cost GPS and accelerometer units are used on a daily basis by football clubs to evaluate the metabolic and mechanical loading experienced by players. These units are typically worn on the trunk but there is little understanding what meaningful information the accelerations might provide from a player overload or injury perspective.

This project firstly characterised the acceleration signals of a commercial accelerometer (sampling at 100 samples per second) compared to three research accelerometers at the trunk, pelvis and tibia (sampling at 1500 samples per second). Ground reaction forces (GRF) were also recorded as was the whole body acceleration (which predicts the GRF). Accelerometer location and the type of task affected the usefulness of the recorded accelerations. The commercial accelerometer underestimated the loading rate, which helped develop a suitable algorithm to pre-process field-based acceleration data.

Secondly, a computer model representing the body was produced to link trunk accelerations to ground reaction forces. The model considered multi-body accelerations and forces and it was validated against measured GRF so that the link to trunk acceleration could be determined. The model was able to successfully predict ground reaction forces from a trunk accelerometer. Whilst the accuracy of the prediction is based on knowing the initial state of the body at impact, it does not require subject specific parameters other than body mass.

The final study was a retrospective case-control study of hamstring injuries from elite Premier League football clubs. Trunk worn accelerometer data from injured players three weeks prior to injury was compared to uninjured controls. Analysis of the pre-processed acceleration data showed that the loading rate of the uninjured players was less than the controls in 7 out of 9 cases. This could be evidence of modified behaviour in the three weeks leading up to the injury but further analysis in a greater number of injuries is required for this to be tested comprehensively.

In summary this project provides a novel rigorous approach to gain meaningful information from trunk-worn accelerometers in the future. By establishing the biomechanical context in which trunk accelerations can predict player loading, there is now scope for monitoring internal loads to the musculoskeletal structures and eventually preventing injury using data collected every day in football clubs.

## 2.0 Background and rationale

An elite football player undertakes some 500 rapid accelerations and decelerations in a single match (Bloomfield, Polman, & O'Donoghue, 2009). The player does this as part of a team effort to win the match, pushing the body to the limits. Helping a player understand when the body is likely to exceed its limits can ultimately help prevent injury. Micro Global Position System (GPS) sensors including heart rate monitors, gyroscopes and accelerometers have been introduced into professional team sports to try and do this. These sensors, typically worn on the trunk (upper back), enable the measurement of movement patterns and have in the past decade started to help coaching staff and researchers to gain a better understanding of the physical demands of team sports (Figure 1). It allows the optimisation of training to enhance the performance whilst also preventing injury by monitoring and adjusting the cumulative load on the players.



*Figure 1. A typical sensor worn daily by football players to measure GPS and accelerations.*

In order to generate high whole-body centre of mass (CoM) accelerations and decelerations, the musculoskeletal system can incur damage that is mostly produced through the generation of *internal loads* between the different structures of the musculoskeletal system. This places bones, cartilage, muscles, tendons and ligaments under considerable stress. With sufficient recovery this leads to desirable adaptations of the entire musculoskeletal system (according to well-known supercompensation principles), but it can also lead to undesirable damage, either when there is insufficient recovery (overload injuries) or when any of the forces become too large for the system to deal with (acute injuries). Essentially, the generation of high forces against the environment is necessary to accelerate and decelerate the body as a whole, yet it imposes high internal loads on the various musculoskeletal structures that can cause injury.

The question is then if it is possible to measure either the external forces on the body, or the consequential whole body accelerations. Measuring the external forces is feasible in a lab where force platforms can be built into the ground, but not in the field. If one wishes to measure whole body accelerations and decelerations, it is important that one estimates the accelerations of the body as a whole. To do this, biomechanists are able to measure relative motions between individual body segments and evaluate the global body movement through calculating the centre of mass (CoM). Whilst all segments together represent the whole body CoM, the relationship between accelerations of one individual segment (e.g. trunk) and the whole body CoM accelerations is dictated by complex multi-body dynamics. This is by no means a simple linear relationship, as has been demonstrated by poor results in studies using linear approaches to define the relationship (e.g. Roisin et al., 2014; Tran, 2010; Wundersitz, Netto, Aisbett, & Gastin, 2013). In other words, one needs to estimate the behaviour of the other segments of the body in order to get to know the relationship between one segment's acceleration and the acceleration of the whole body CoM.

NOTE: With technological progress one has to be prudent not to embark on a reverse engineering mission. It is essential not to define problems for the new technological solution, but to start from the problem at hand and identify measurement needs. In this case, the problem is the monitoring of mechanical loads on the player, and the measurement need is whole body CoM acceleration as a representation of these loads. The technical question is therefore whether trunk-mounted accelerometry allows us to estimate whole body CoM accelerations.

To establish how trunk accelerations relate to whole body CoM accelerations, one needs to deploy a more complex approach that incorporates knowledge of how the various segments interact with each other. Building a computer model of how the body interacts with the environment allows the true relationship between trunk acceleration and the acceleration of the whole body CoM to be established. This relationship is expected to be different between various tasks, but is expected to be fairly robust across tasks with similar acceleration/deceleration profiles. The challenge is then to make the computer model sufficiently complex so that it can establish the relationship well, whilst keeping it sufficiently simplistic so that it remains computationally feasible for usage in real time. Such computer models can be validated in a lab environment, to see whether and how well it

reproduces the relative behaviours of the body segments, and ultimately how well it can mimic the external forces as a result of those behaviours.

Once a computer model is tested in a controlled environment, its usage requires the preparation of field-based data to reveal relevant parts of the continuous data streams that come from the accelerometer. Particularly the removal of irrelevant information is important, as that would otherwise lead to unnecessary background noise blurring the relevant information. More than two thirds of the total distance covered during soccer matches involves motions that are of no immediate interest, as only about one third is covered using high intensity utility movements (Thatcher & Batterham, 2004). To use an analogy on how this affects the observation, Figure 2 shows how an image becomes useless to interpret when it is blurred by irrelevant noise. The same holds true for the interpretation of acceleration signals. Attempts should be made to remove irrelevant information, e.g. low level whole body accelerations that involve negligible internal loads.



*Figure 2. An example of a stress fracture blurred and not blurred. This is likened to the extraction of meaningful acceleration data.*

### **3.0 Literature Review**

#### **3.1 GPS-Accelerometry monitoring**

Since the development of micro Global Position System (GPS) sensors including heart rate monitors, magnetometers, gyroscopes and accelerometers for professional team sports in 2003 the use of these devices has increased within recent years (Dellaserra et al. 2014). They are primarily used during training in professional team sports (e.g. football, Australian football, hockey and rugby) and they are even allowed during games in Australian Football and rugby, whilst in football there is still the reliance on the semi-automatic tracking systems (e.g. Prozone) during games. Due to the increased application in traditional outdoor team sports and the development of local indoor GPS systems as well as the information provided by the inertial sensors (magnetometer, gyroscope and accelerometer) the systems are now applied in indoor team sports as well (e.g. basketball and netball) (Montgomery et al. 2010, Cormack et al. 2013a, Cormack et al. 2013b). These sensors are typically located on the upper trunk between the scapulae within a small pocket of a tight fitted elastic vest.

Information from GPS enables objective measurements of player position, distance travelled, running velocity, time spent in specific running velocity zones etc. This information is expected to help coaching staff and researchers to gain a better understanding of the activity profiles and physiological/metabolic demands in team sports. This allows the coaching staff to adjust and optimize training to reach the desirable physiological training adaptations or improve the players' tactical performance (Dellaserra et al. 2014, Cummins et al. 2013).

As GPS data is only recorded at 10Hz (10 times per second) its' ability to accurately detect high intensity accelerations/decelerations and changes in direction is questionable (Varley et al. 2012, Jennings et al. 2010, Akenhead et al. 2013). Researchers and practitioners have therefore turned their attention to 100 Hz acceleration data from three-dimensional accelerometers within the GPS device to estimate loads to which the musculoskeletal system is exposed to from the high external forces during every foot contact (Cunniffe et al. 2009, Gabbett et al. 2011). The accelerometers measure the frequency and magnitude of linear accelerations along its individual axes and have commonly been used for daily life activity monitoring, gait analysis and balance assessment (Ward et al. 2005, Yang and Hsu 2010, Dellaserra et al. 2014). The higher sampling frequency of the accelerometer (100 Hz) compared to the GPS (typically between 10-15Hz) improves the ability to monitor high



intensity movements so that no valuable information about the magnitude or change in acceleration is lost.

### 3.2 Reliability and validity

The commercial GPS integrated accelerometers have demonstrated good reliability during laboratory and field tests (Boyd et al. 2011, Kelly et al. 2014). The accelerometers' ability to measure peak accelerations have been tested in recent studies with contradicting results. When commercial accelerometers were tested against a reference accelerometer underestimated peak accelerations were found, particularly during high frequency movements (Lake et al. 2014, Kelly et al. 2014), whereas the commercial accelerometer overestimated peak acceleration when tested against acceleration data obtained from three-dimensional motion analysis systems (Wundersitz et al. 2013a, Wundersitz et al. 2014, Wundersitz et al. 2015). The overestimation could be caused by the poor quality "fixing" of the accelerometer to the trunk resulting in unwanted high-frequency artefacts in the accelerometry data and the application of a low-pass filter has shown to improve the validity though the optimal filtering is yet to be established (Wundersitz et al. 2013a, Wundersitz et al. 2014, Wundersitz et al. 2015).

### 3.3 Monitoring of training loads

Despite the questionable validity, trunk mounted accelerometers are frequently used in the field to try and monitor training load (Boyd et al. 2013, Walker et al. 2015, Cummins et al. 2013) and detect tackle/collision impacts in rugby and Australian football (Kelly et al. 2012, Gastin et al. 2013, Gastin et al. 2014). Many of the commercial units generate an overall indicator of the training load (e.g. PlayerLoad<sup>TM</sup>), a scalar-based summative index measuring the changes in acceleration (Boyd et al. 2011). In this way, the overall training load is reduced to a single variable, making it easier for practitioners to use. A large number of studies have used PlayerLoad<sup>TM</sup> to determine players' physical activity profiles in a variety of team sports (Barron et al. 2014, Scott et al. 2013, Boyd et al. 2013, Polglaze et al. 2014, Cormack et al. 2013b, Sullivan et al. 2014, Kempton et al. 2014, Montgomery et al. 2010, McNamara et al. 2013, Chandler et al. 2014). These studies looked at the difference in physical activity profiles between match and competition, between different playing positions, and between teams at different levels. Other studies have also investigated the relationship between PlayerLoad<sup>TM</sup> and physiological markers (metabolic load), e.g. heart rate, VO<sub>2</sub>max and rating of perceived exertion (RPE) (Scott et al. 2013, Casamichana et al. 2013, Scanlan et

al. 2014, Gallo et al. 2014, Barrett et al. 2014), energy expenditure (Walker et al. 2015). Large variations in PlayerLoad<sup>TM</sup> have been observed between subjects, suggesting that inter-subject comparisons ought to be done with care (Barrett et al. 2014).

### 3.4 Mechanical Loading

The ability of trunk mounted accelerometry to estimate mechanical load in dynamic team sports is still largely unexplored. In biomechanics, force measuring platforms are typically mounted into the floor to measure the external forces acting on the body, but this is restricted to laboratory testing. Due to the linear relationship between force and acceleration demonstrated by Newton's second law of motion, measuring accelerations of the body or its individual segments is the closest one can get to estimate the mechanical load on the body during dynamic movements in the field. Recent development of inertial measurements units (IMU) have made it possible to measure motion and even estimate internal joint forces in the field. However as these systems require one sensor on each segment of the body it makes it difficult to apply in a practical football setting. A simpler system with four tri-axial accelerometers mounted on the trunk have been used to measure hip joint loading during walking and running (van den Bogert et al. 1999). In addition higher resultant peak acceleration measured from trunk mounted accelerometry during the deceleration phase of a turning movement does appear to correspond very well with indirect indicators of lower limb loading (ankle and knee joint angular velocity) (Nedergaard et al. 2014). The ability of commercial trunk mounted accelerometers to estimate mechanical loading has been investigated by comparing trunk acceleration data with ground reaction forces during running, change in direction, jumping and landing tasks (Wundersitz et al. 2013b, Tran 2010). Both studies showed that trunk mounted accelerometers overestimate the peak vertical and resultant ground reaction force and may be unsuitable as a measure of mechanical loading. Though differences in acceleration have been observed between accelerometers placed on the upper and lower trunk during running (Kawabata et al. 2013) and in PlayerLoad<sup>TM</sup> measured with a trunk and hip mounted accelerometer during running (Barrett et al. 2014). An important shortcoming in these approaches however is the fact that although the relationship between whole body acceleration and ground reaction forces is direct, but the relationship between segmental accelerations and ground reaction forces is not.

### 3.5 Injury

A recent study has attempt to relate PlayerLoad<sup>TM</sup> to injury risk in which data from commercial GPS/accelerometer devices and injury incidence were collected from 46 elite Australian football players during one season. PlayerLoad<sup>TM</sup> showed the strongest relationship to injury risk compared to the variables collected from the GPS (e.g. total distance covered and sprinting distance) (Colby et al. 2014). Furthermore an ongoing cohort study with up to 420 elite junior Australian football players is investigating pre-season training load risk factors (e.g. weekly cumulative workload) for injuries (Lathlean et al. 2014). This study provides initial evidence that mechanical load estimates could be relevant for the prediction of injuries (as illustrated in Figure 3), although to date it is unknown how this would relate to football players and it remains unclear what information a variable such as PlayerLoad<sup>TM</sup> can reveal.

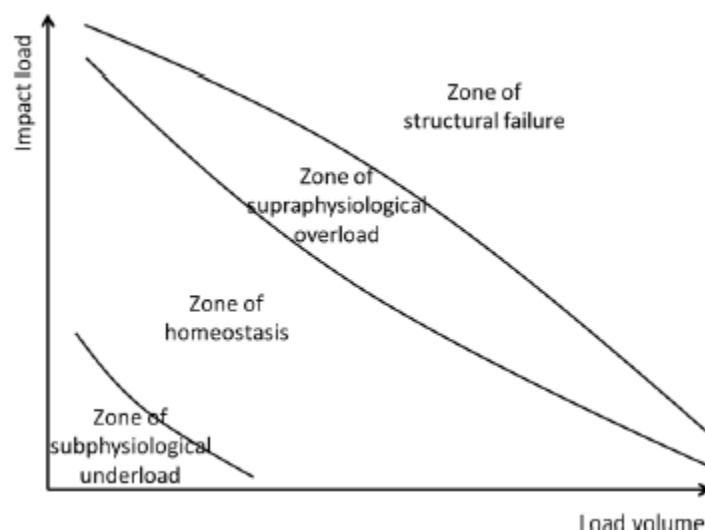


Figure 3: (Bahr 2014) The relationship between structural adaptation and load. Load volume can be characterised as training volume (frequency, duration, intensity), match frequency, etc. Impact load can be characterised as measures of joint (mechanical) load.

## 4.0 Aims

This project will:

- a) Establish the relationship between accelerometry and mechanical cost by evaluating and monitoring players during dynamic tasks as part of match simulations.
- b) Quantify the external and internal forces applied to the body and its musculoskeletal structures
- c) Identify an associative algorithm for body worn accelerometry.

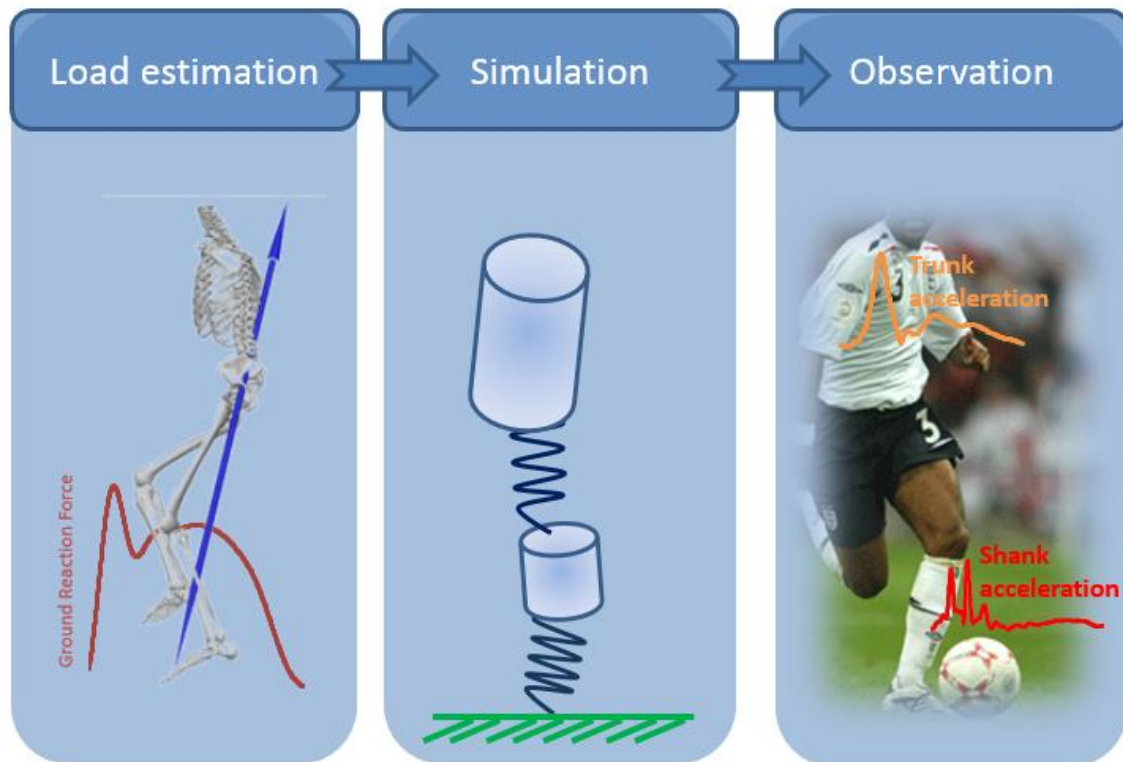
*Aim study 1.* To develop an algorithm that characterises the relationship between triaxial high frequency accelerometry and mechanical cost, including a sensitivity analysis of the algorithm for when the accelerometry is measured at different locations on the body.

*Aim study 2.* To apply the algorithm developed in study 1 to a historical case-control database from training sessions in Premier League football players three weeks prior to an injury incident, to facilitate modelling of mechanical load thresholds in football.

## 5.0 Research

### 5.1 Research Overview

When the body moves external forces are experienced which are caused by our interaction with the environment. During gait-like activities, the only significant external force experienced is the ground reaction force (GRF) (Figure 4). GRF can be measured using highly accurate lab-based force platforms which are rigidly mounted in the floor. The GRF measured by force platforms represents the acceleration of the whole body CoM (the location of average mass distribution about the body) and the weight of the body acting on the platform. When considering the mechanical load experienced by the body, the GRF would be considered the “gold standard”. Ground reaction forces are typically around 1 body weight during low dynamic activities such as walking, they increase to around 2.5 body weights during running but can become much higher during highly dynamic tasks such as jumping. In addition to the type and intensity of activity, the surface on which the tasks are taking place will also affect the mechanical load exerted on the body.



*Figure 4. Visual illustration of the concepts involved in study 1. External forces are experienced during match play, and simulated behaviour of a computer model can help identify the relationship between these forces and measured accelerations of individual body segments.*

The challenge lies in estimating the GRF from accelerations measured on individual body segments. To do this requires a combination of (1) experimental data and (2) computer modelling to establish the value of trunk accelerations currently measured by football clubs. We first need to establish the signal characteristics of accelerations, as they can be measured experimentally. The experimental data then allows us to measure trunk accelerations as well as ground reaction forces during dynamic tasks in the lab typically undertaken during football. The computer modelling then provides a mathematical framework within which we can firstly establish the relationship between accelerations and GRF and secondly, predict ground reaction forces from trunk accelerations only (as this would be only available data within a football environment). This can then help to inform the evaluation of trunk acceleration data against injury data from an applied football context (study 2).

## 5.2 Study 1

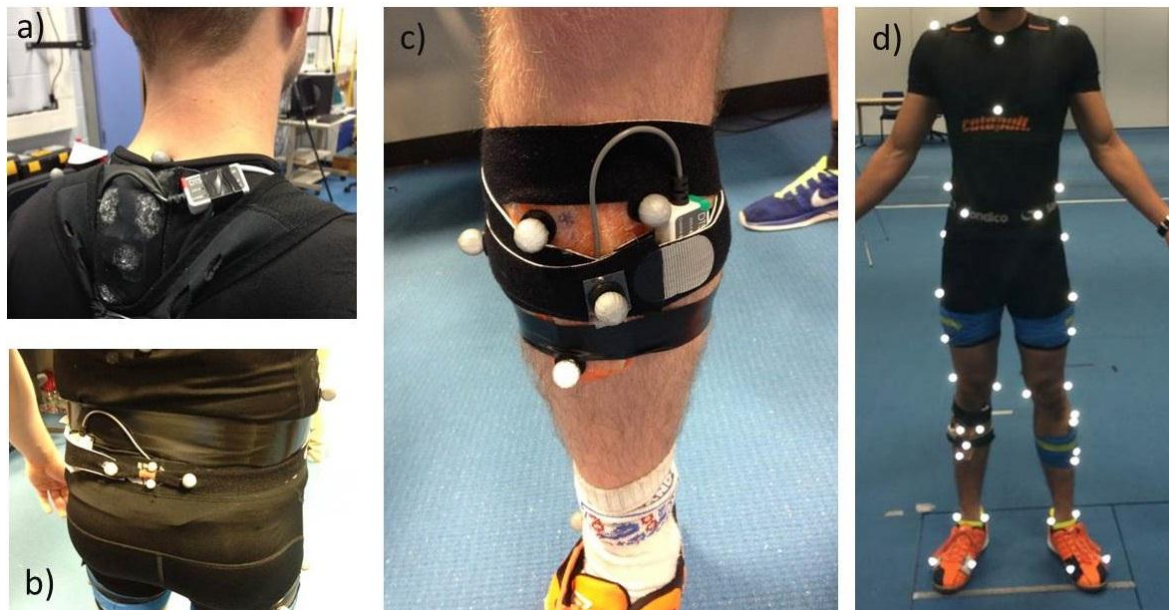
### 5.2.1 Experimental observations

#### 5.2.1.1 Methods

For study 1 we recruited 18 participants and collected the following data:

- i) Trunk accelerations from a triaxial piezoelectric accelerometer (KXP94, Kionex Inc., Ithaca, NY, USA) within a commercial GPS device (MinimaxX S4, Catapult Sports, Melbourne, Australia).
  - Weight: 67 grams (GPS device)
  - Sampling frequency: 100 Hz
  - Output range:  $\pm 6g$
  - Bandwidth: Not specified by the manufacture
- ii) Accelerations from a triaxial research accelerometer (518, DTS accelerometer, Noraxon Inc., Scottsdale, USA) positioned at the trunk (Figure 5a), the pelvis (Figure 5b) and the tibia (Figure 5c).
  - Weight: 5.7 grams
  - Sampling frequency: 1500 Hz
  - Output range:  $\pm 24g$
  - Bandwidth: 5-1800 Hz
- iii) 3D kinematics (angular motion) of the lower-limbs and trunk (500 Hz) - these segments are sufficient to provide an accurate representation of the whole body centre of mass during dynamic tasks (Vanrenterghem *et al.*, 2010).
- iv) Ground reaction forces (1500 Hz)

Participants performed a variety of locomotor tasks commonly experienced in football including running, 45° change of direction (cut 45), 90° change of direction (cut 90), vertical jumping, single leg drop-landings (SLDL). All tasks were recorded at 4 different intensities (speeds or heights). These tasks could have been completed in a 45 minute simulated match protocol, but to improve repeatability and consistency of the desired task variations they were recorded from separate tasks and intensities.



*Figure 5. The locations of the four accelerometers during experimental data collection a) trunk – both the commercial and the research accelerometer are rigidly coupled then worn in the typical vest, b) the waist, c) the shank – the accelerometer is mounted onto a fibreglass plate which is tightly fixed to the tibia, d) the whole body set-up including reflective markers to track the motion of the body. The participant is also stood on a force platform.*

#### 5.2.1.2 The effect of accelerometer location

It may seem appropriate that if you wish to measure the load exerted on the body then you need to place an accelerometer as close to the point of application of the load as possible e.g. the foot. This is however an incorrect assumption. Individual segments experience different accelerations (Figure 6) but the overall load experienced actually relates to the whole body centre of mass acceleration. This section characterises the accelerations observed at the trunk, pelvis and tibia. These accelerations are also compared to the commercial accelerometer as typically positioned at the trunk. Visual comparison is provided in Figure 6 and exact quantification of common summarising variables is provided in Table 1.

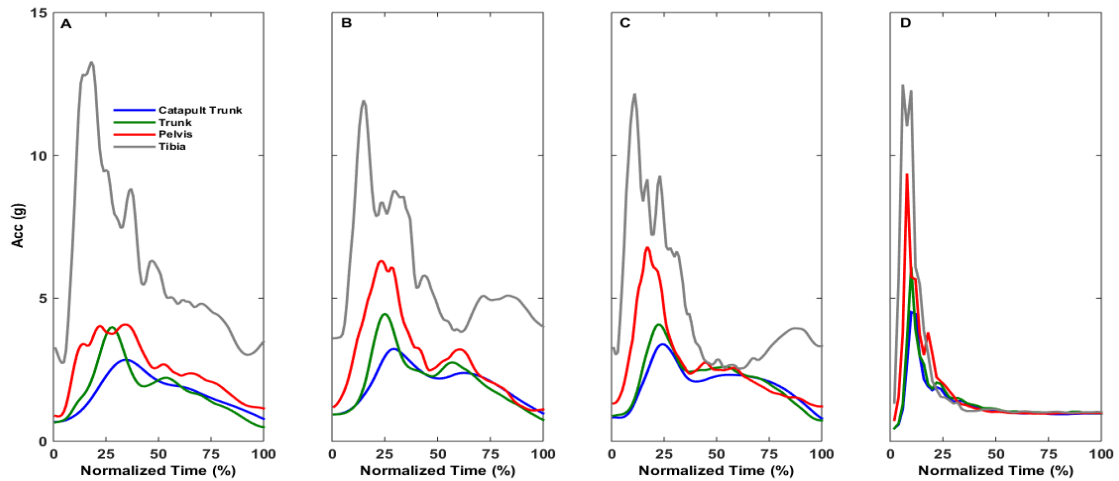


Figure 6. Mean resultant acceleration curves for the different accelerometer locations from one foot-ground-contact for (A) straight forward running (run), (B) 45 degrees change of direction, (C) 90 degrees change in direction all with an approach speed of 5 m/s and (D) single leg drop landings from a 50cm drop height.

Accelerometer location greatly affects the magnitude of the accelerations observed across all tasks (Figure 6). The tibial accelerometer recorded the highest accelerations throughout the tasks and it frequently exceeded 12 g but even got to 22 g following an impact. This is explained by this accelerometer being closest to where impact with the ground occurs, without any shock absorption through multi-body interactions. The pelvis accelerations tended to be of larger magnitude (~8 g) than both trunk accelerometers (~3-5 g). The pelvis data also varied most across the different tasks (Table 1). There were subtle differences observed between the two trunk accelerometers with the commercial unit appearing to slightly underestimate the magnitude of acceleration when compared to the research accelerometer. This is likely due to the lower sampling rate of the commercial unit which has the effect of filtering the acceleration signal.

For all tasks the acceleration profiles were qualitatively similar across all accelerometer locations. The timing of the peak accelerations follow a distal-to-proximal pattern from the tibial to the trunk accelerometers as the impact accelerations are transmitted upward through the body (Table 1). In most tasks, the timing of the peak acceleration also differed between the two trunk accelerometers as the lower sample rate of the commercial unit slightly underestimated the peak value.



Where one places an accelerometer has big consequences for the size and shape of the measured signal. The lower sampling rate of the commercial accelerometer likely underestimates the true loading rates.

*Table 1: Mean ( $\pm$ SD) data across the different approach speeds/drop heights and intensities (288 trials per tasks) for straight forward running (run), 45 degrees change in direction (Cut45), 90 degrees change in direction (Cut90) and single leg drop landings (SLDL).*

Variable / Task	Trunk (Commercial)	Trunk	Pelvis	Tibia
<b>Peak resultant acceleration (g)</b>				
Run	3.43 (0.72)	4.62 (1.28)	5.84 (1.80)	13.31 (3.88)
Cut45	3.60 (0.81)	4.77 (1.32)	7.69 (3.01)	14.18 (4.74)
Cut90	3.79 (0.94)	4.82 (1.45)	9.39 (4.32)	15.18 (5.64)
SLDL	5.11 (1.33)	7.00 (2.42)	12.19 (5.11)	22.22 (6.17)
<b>Time to peak resultant acceleration (ms)</b>				
Run	75 (23)	60 (21)	65 (26)	33 (20)
Cut45	84 (36)	65 (30)	65 (28)	36 (24)
Cut90	88 (47)	75 (47)	62 (45)	40 (31)
SLDL	77 (21)	65 (17)	56 (22)	48 (24)
<b>LOADING RATE (g/ms)</b>				
Run	0.05	0.08	0.09	0.40
Cut45	0.05	0.07	0.12	0.39
Cut90	0.05	0.06	0.15	0.38
SLDL	0.07	0.11	0.22	0.46

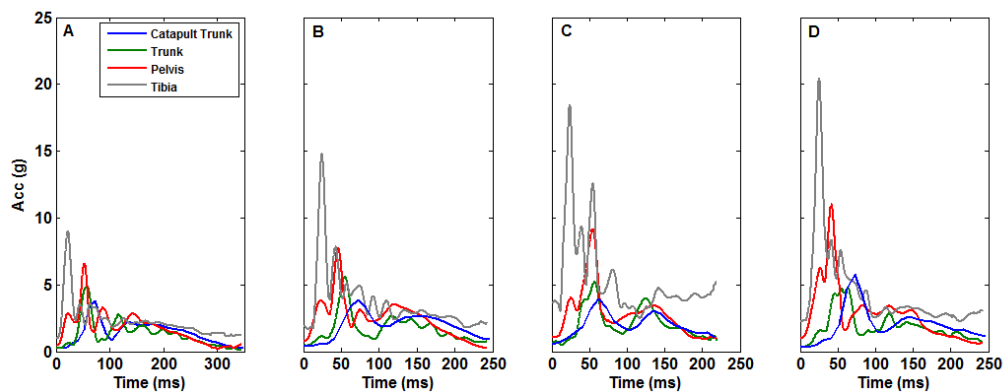
Key observations from Table 1 are that players are exposed to the highest accelerations during the single leg drop landing task, the peak accelerations are very similar during the run, Cut 45 and Cut 90 tasks. The pelvis and the tibia had the greatest standard deviations and are therefore most variable signals.

#### 5.2.1.3 The effect of task intensity

This section looks at acceleration characteristics across the different task intensities for all accelerometer locations. Example visualisation of the effect of intensity for the Cut 45 is shown in Figure 7.

The key features of accelerometer location as discussed in 5.2.1.2 are consistent across the different task intensities. As task intensity increased from 2-5 m·s<sup>-1</sup> during a change of

direction, the peak accelerations also increased. The most prominent increases were observed in tibial acceleration.



*Figure 7. Mean resultant acceleration curves from one subject for the different sensor locations from one foot-ground-contact for a 45 degrees change in direction (Cut45) at four different approach speeds (2 m/s A to 5 m/s D).*

Acceleration increases as task intensity increases. The commercial accelerometer underestimates loading rate across intensities.

#### 5.2.1.4 The relationship between segmental acceleration and whole body centre of mass acceleration

The ground reaction force (GRF) represents the external mechanical load experienced by the body. As such, it reflects the acceleration of the whole body centre of mass (CoM) and not the individual body segment accelerations. This section compares the GRF to whole body CoM acceleration, and then to segmental accelerations.

Comparison to the whole body CoM acceleration (Figure 8) provides an indication of how well experimentally measured accelerations can be used to represent the GRF. The whole body CoM acceleration provides a good representation of the profile of the GRF for most tasks. As the whole body CoM represents the GRF, next to consider is how well segmental accelerations represent the whole body CoM acceleration.

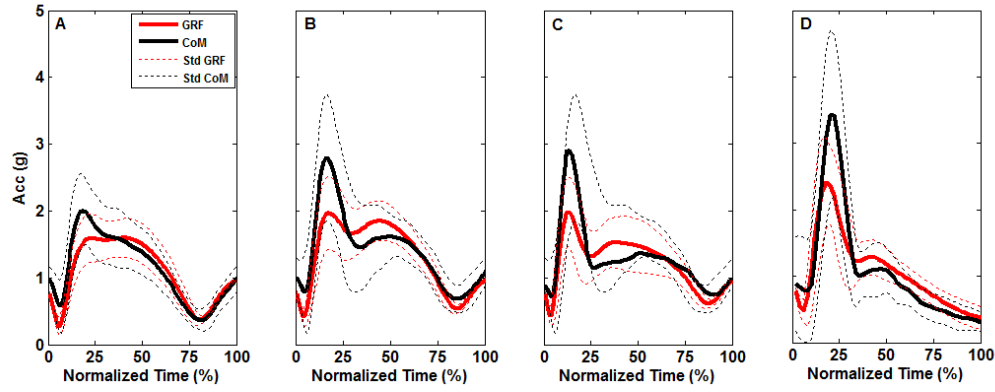


Figure 8. Mean whole body CoM and GRF resultant acceleration curves from one foot-ground-contact for (A) straight running, (B) 45 degrees change in direction, (C) 90 degrees change in direction all with an approach speed of 5 m/s and (D) single leg drop landings from a 50cm drop height.

In an applied situation one can only attach accelerometers to body segments and therefore measure segmental accelerations. The question then is which segment to best attach the accelerometer to in order to represent whole body CoM accelerations. Comparison of the whole body CoM acceleration to the measured segmental acceleration (Figure 9) helps to understand which segmental accelerations could best predict the GRF.

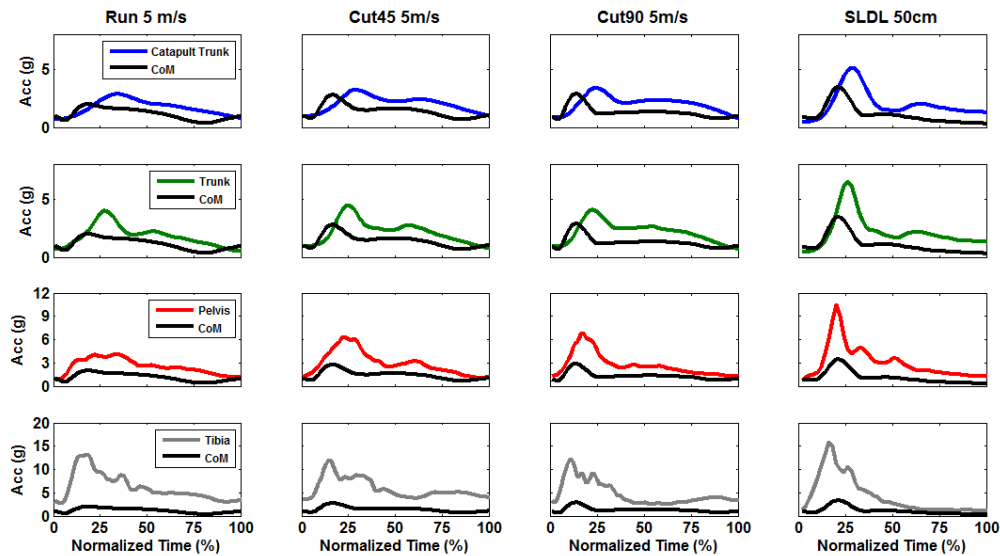


Figure 9: Mean CoM resultant acceleration curves compared to mean resultant acceleration curves from the four sensor locations from one foot-ground-contact for straight forward running (run), 45 degrees change in direction (Cut45), 90 degrees change in direction (Cut90) all with an approach speed of 5 m/s and single leg drop landings (SLDL) from a 50cm drop height.

Observations from Figure 9 are consistent across the different tasks. No acceleration signal adequately represents the CoM acceleration profile. Trunk accelerations most closely follow the magnitude of the whole body CoM accelerations whereas the pelvis and tibial accelerations greatly exceed the whole body CoM acceleration. Only the tibial acceleration characterises the timing of the CoM acceleration peak with the pelvis, and trunk accelerometers tend to have a delayed peak. The rate of loading (initial curve steepness) is overestimated by the tibial accelerometer but underestimated by the commercial accelerometer. The research accelerometers on the trunk and pelvis replicate the rate of loading on the whole body CoM most accurately. Whilst intuitively one may believe that measuring accelerations nearer to the ground is more representative of GRF than being further away, this data clearly shows that both the magnitude and the loading rate of the tibial accelerations will not appropriately match whole body centre of mass accelerations and therefore GRF. This observation has important consequences for the direct prediction of GRF from trunk or pelvis accelerations as it would make the initial ground reaction force peak more difficult to predict given that the peak in the acceleration occurs after the GRF peak.

The commercial accelerometer reasonably approximates the whole body centre of mass acceleration in comparison to the pelvis and the tibia so this suggests that the location of the commercial unit is somewhat optimal. The lower sample rate of the commercial unit however has an important consequence for the rate of loading. Given that it is likely that the rate of loading is related to overload and injury, the commercial unit does not sample the rate of loading adequately (at its' sampling rate of 100 Hz) and so any estimation of loading based on accelerations from these units is likely to underestimate the true whole body loading (also see study 2).

We illustrated the direct relationship between the GRF and the whole body CoM acceleration. Then we showed that segmental accelerations are not directly related. The tibial acceleration clearly does not represent whole body centre of mass acceleration and therefore GRF. The trunk and pelvis accelerations better match the whole body CoM accelerations – in particular the loading rate (initial steepness). The commercial unit accelerations also matched the whole body CoM acceleration but underestimated the loading rate.

#### 5.2.1.5 Summary

Ground reaction forces are inherently linked to the whole body CoM acceleration. The acceleration of individual body segments e.g. the trunk and pelvis provide some representation of the whole body CoM acceleration but this is not a direct relationship, it is a relationship that depends on multi-segmental dynamics (the inter-play of segmental motion and forces). To be able to use segmental accelerations to predict GRF we need a model that represents the body's multi-segmental dynamics.

### 5.2.2 A mathematical model (algorithm)

In parallel with the experimental data collection and analysis we also developed a mathematical model that allows predicting the actual mechanical loading on the body (the GRF) from trunk accelerations. Here we summarise the development and validation of the model.

#### 5.2.2.1 Why is a mathematical model needed?

Ground reaction forces (GRF) are only measurable in a laboratory environment. To be able to estimate GRF when they are not measured, the mechanical coupling between the input data (in this case body-worn accelerometry) and the GRF has to be established. Based on Newton's second law of motion  $F = m \cdot a$ , one way of establishing this relationship is to represent the movement (and therefore acceleration) of the body as a mass coupled with a spring (Figure 10a). This is of course an oversimplification but it provides a way of calculating GRF based purely on the equations used to define the model (an analytical solution). The analytical solution can then be compared to a numerical solution which is calculated using numerical values. We can increase the model's complexity and use the analytical and numerical solutions to validate the model's behaviour and to check that the GRF produced are appropriate (Figure 11). Without such model the trunk accelerometer on its own cannot adequately reflect the whole body CoM acceleration to predict the ground reaction force experienced (5.2.1.4).

#### 5.2.2.2 Model development

Spring-mass models have been used to estimate GRF in animals and humans during running outside of a lab environment modelling the absorption and return of energy in the body as a simple spring (Alexander et al., 1986; McMahon et al., 1987).

The simplest spring-mass model however (Figure 10a) is unrealistic, as the spring acts linearly. All initial energy provided by the acceleration of the mass is returned back to the mass following compression of the spring. This is a fully efficient system – which the human body is not. During impact with the ground there is actually loss of energy or shock absorption that occurs. Adding a damper to the model (Figure 10b) provides a more realistic estimate of whole-body behaviour, but even this model fails to produce the patterns of ground reaction forces seen during typical running type movements (Derrick et al., 2000).

During an activity such as running, the shape of the GRF curve is characterised by an early passive impact phase as the foot hits the floor, and this is then followed by an active peak as the downward motion of the centre of mass is reversed. Both active and passive structures in the body contribute to this complex energy absorption and return. The spring-mass models in Figure 10 a and b do not allow for both a passive and active peak in their dynamic behaviour. This is thought to be because the passive phase is an independent phenomenon from the active phase and specifically associated with the dynamics of the support leg. To get a realistic estimate of the GRF force then, two masses are required (Figure 10c). The lower mass is typically around 20% body mass – therefore likened to the mass of the stance leg and the upper mass is the remainder of mass in the body. The spring between mass 2 and the ground can be thought of as the lower limb stiffness whereas the upper spring can be thought of as whole-body compliance. Care should be taken with interpretation because this system is still a simplified system and cannot be directly related to a particular joint e.g. knee joint stiffness but rather provides an estimate of the overall stiffness of multiple joints that absorb energy during impacts.

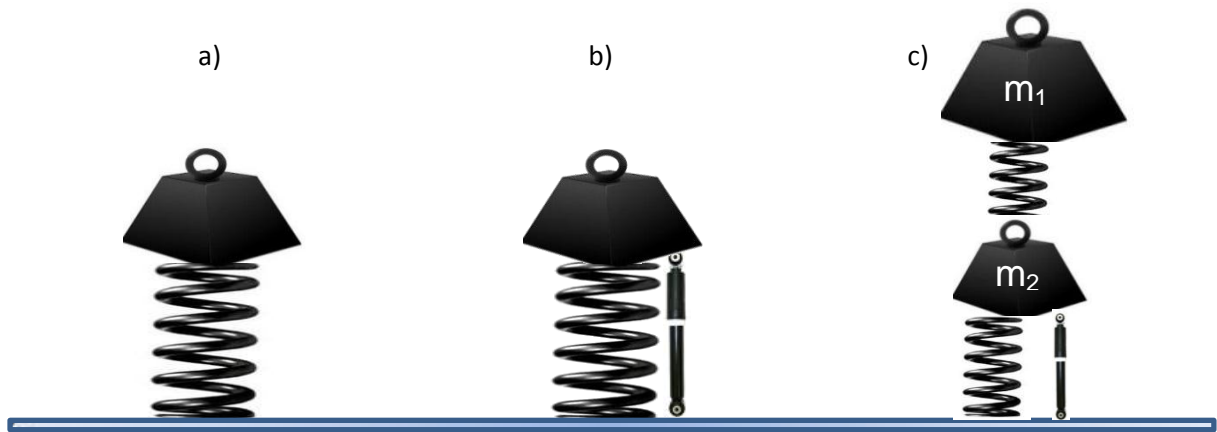


Figure 10. Three mass spring models of increasing complexity which can be used to calculate GRF. a) A simple mass-spring system, b) A mass-spring system with damper, c) A two mass system with a damper on the lower spring.

Spring-mass models have been used in the field to investigate the running economy of different running techniques (McMahon et al., 1987), changes in spring-mass behaviour during exhaustive running and over 5-km running (Derrick et al., 2002; Girard et al., 2013), as well as changes during sprinting and repeated sprints (Girard et al., 2011; Morin, Jeannin, Chevallier, & Belli, 2006). The use of these models to predict GRF in team sport movements is unknown yet they provide arguably the best approach to use information from body-worn accelerometry for estimating GRF.

#### 5.2.2.3 Model results

The equations to describe the behaviour of all mass-spring models in Figure 10 have been written and implemented in Matlab and are provided in Appendix 2. This allows us to provide both analytical and numerical solutions which validate the equations and the GRF which the models generate (Figure 11).

For a two mass system, the solution to calculate GRF can be given by the following equation.

$$GRF = m_2 \ddot{x}_2(t) + m_1 \ddot{x}_1(t) - (m_1 + m_2)g \quad (\text{Equation 1})$$

Where –  $m_1$  and  $m_2$  are the masses (see Figure 10c),  $\ddot{x}$  is the acceleration of  $m_1$  and  $m_2$ , and  $g$  is gravitational acceleration. So for a two-mass system GRF can be estimated according to the above equation. If the accelerations of  $m_1$  and  $m_2$  are known then the GRF can be estimated (Figure 12).

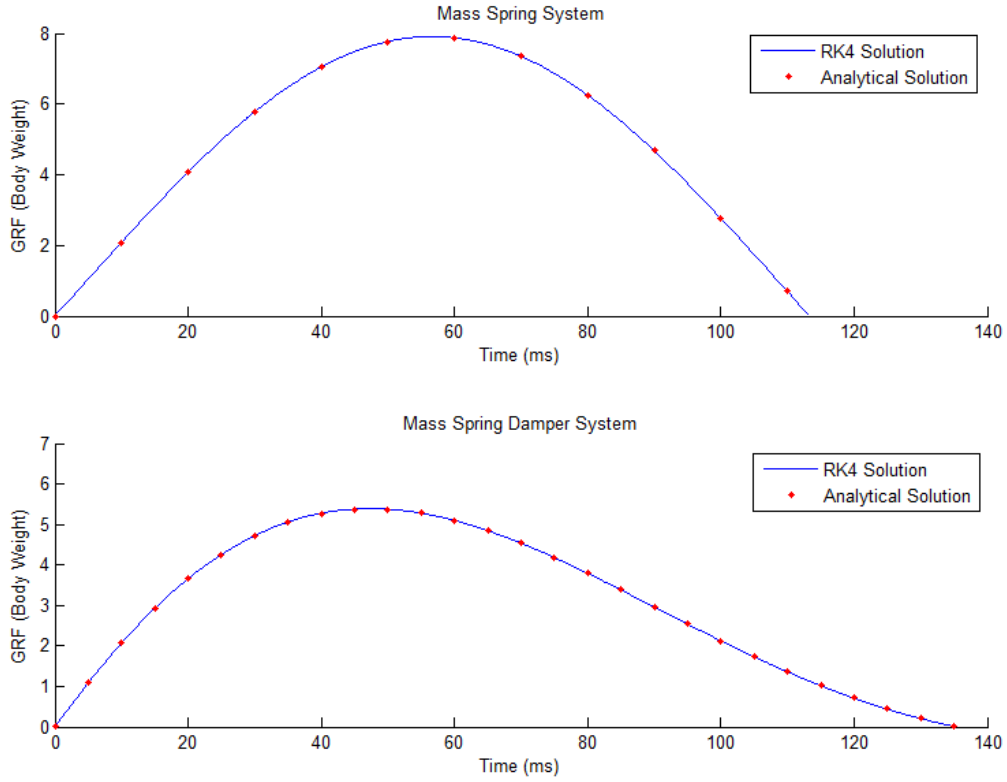


Figure 11. Analytical and numerical solutions for the mass-spring (upper) and mass spring damper (lower) systems calculation of GRF. The analytical solution is identical to the numerical solution for the GRF validating the implemented equations. The mass-spring system provides a symmetrical ground reaction force profile whereas the damped system incorporates energy loss and therefore this is reflected in the altered profile of the ground reaction force.

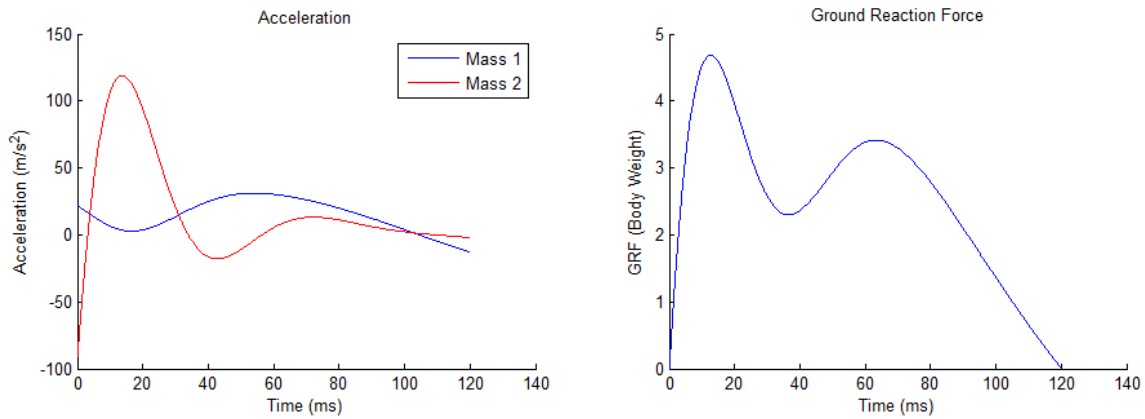


Figure 12. Example data from a two-mass damped model. (Left) Accelerations of mass 1 and mass 2, (Right) The estimated ground reaction force.

Equation 1 has important consequences for the use of trunk acceleration as the sole source of information for the prediction of mechanical load. It demonstrates that from the acceleration of a single mass (this could be the measured trunk acceleration from a football player) the GRF could not be appropriately predicted using either a single-mass or a single-mass-damped



model as the active component of the ground reaction force is not present (compare the shape of the curves in Figure 11 against Figure 12-right). The active component of the ground reaction force is important to the model as this is typically when the peak force occurs and when certain acute injuries are expected to occur (e.g. anterior cruciate ligament injuries are thought to occur during the weight acceptance phase of sidestepping).

The two-mass model with damper seems to provide the minimally complex model for the calculation of GRF.

Equation 1 requires an estimate of the acceleration of both mass 1 and mass 2 to calculate GRF. With the experimental data generated we can use the trunk accelerations to calculate the ground reaction forces and validate the predicted ground reaction forces against the measured forces from the force platform. Discrepancies can be identified and tested with a sensitivity analysis of the model parameters, such as the stiffness settings of the springs.

Ultimately though, we wish to be able to predict GRF from trunk accelerations only as this is the only data available to football clubs. This is equivalent to having the acceleration of mass 1 only. It is therefore clear that we need to have a two-mass model but that we need an estimate of the behaviour of the second mass, otherwise we will not get an accurate GRF prediction.

#### 5.2.2.4 Investigating model sensitivity

To allow the model to predict ground reaction forces from the acceleration of the upper mass, we need to know 8 different parameters. Four of those parameters describe the general mechanical behaviour of the body / model and four describe the state of the body upon impact. The equation showing how the parameters relate to the ground reaction force is given in equation 2.

$$GRF = -\frac{M\omega_2}{1+\lambda}(\omega_2 P_2 + 2\zeta \dot{P}_2) \quad (\text{Equation 2})$$

However, before GRF can be predicted for individual players we need to know how player-specific the model parameters are. This is investigated by firstly identifying an appropriate set of parameters that allow the model to match experimental ground reaction forces. Secondly, the model's upper mass acceleration is matched to the experimentally measured trunk acceleration to see if this then reproduces the GRF. There is no guarantee that (1) by

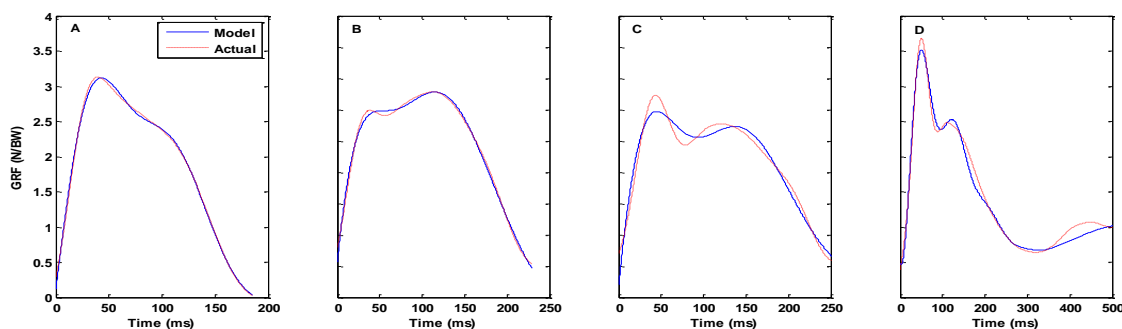
matching the modelled versus measured ground reaction forces, and (2) by matching the acceleration profiles, the same parameters will be determined. However we will have a better understanding of how sensitive the predictions are to changes in any of the parameters.

A parameter sensitivity analysis will help to determine the error with which ground reaction forces can be predicted from trunk accelerations and which practical steps football clubs could use to get the most appropriate loading estimates for their players.

### Step 1: Parameter estimation by matching modelled versus experimental GRF

Ranges for each parameter were split into equal intervals and each possible combination of the values were used to solve the model. Each of the solutions were compared with the actual ground force as measured experimentally. This resulted in interrogating 65,536 different models and a computation time of nearly 2 hours. This process was repeated a number of times which was highly computationally demanding. In order to speed up the parameter search we fixed certain parameters at specific values then performed a search on remaining parameters. Further analysis showed that the parameters describing the state of the body upon impact were hyper sensitive and even small variations can mean that the GRF is not predicted accurately. This means that the four parameters determining the state of the body at impact need to be estimated from the acceleration data for each contact.

Importantly, the parameters describing the mechanical behaviour of the body can be consistent for each subject. Figure 13 shows that as a result of this process, by finding the optimal parameters for a subject, the model can recreate the GRF very accurately for all tasks. By reproducing the GRF experienced, this shows that we can mimic the physical behaviour of the human body during dynamic tasks.



*Figure 13: Measured and modelled GRF for A) straight forward running, B) 45 degrees change in direction, C) 90 degrees change in direction all with an approach speed of 5 m/s and D) single leg drop landings from a 50cm drop height.*

Table 2 shows how sensitive the eight parameters are between subjects. In summary the four parameters determining the state of the body at impact are very sensitive. This means that the model's reproduction of GRF is sensitive to the task that is being performed. In this study the model was validated against a range of highly dynamic and less dynamic tasks for robustness. As an example consider that the model (and player) stiffness is likely to change dramatically for an “elastic” movement such as running where energy is absorbed and returned and this is likely to be quite different from an “inelastic” movement such as single leg drop landings where energy is absorbed by the player.

The practical implication of the model parameter sensitivity analysis is that there is no need to attempt to calibrate the model for individual subjects (except for their mass) but a high quality acceleration signal is needed so that the initial state of the body can be determined accurately.

*Table 2. Optimal parameters to recreate the ground reaction forces during running. This illustrates the inter-subject variation in parameter estimation.*

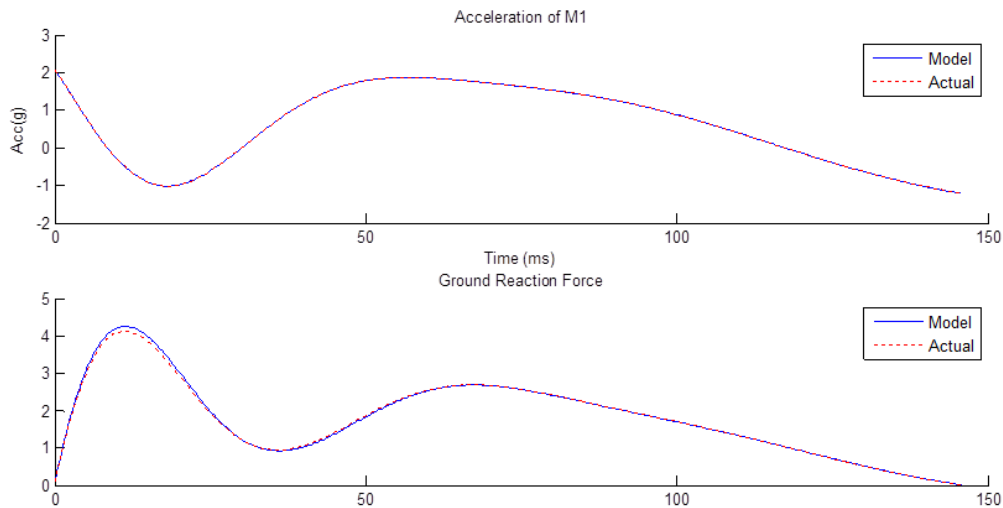
Subject	Trials	Mass (kg)	P1 (m)	P2 (m)	v1 (m/s)	v2 (m/s)	$\lambda$ (au)	$\omega_1^2$ (Nm <sup>-1</sup> )	$\omega_2^2$ (Nm <sup>-1</sup> )	$\zeta$ (au)
1	16	64	0.005	0.017	-1.15	-0.79	3.610	670.2	2520.9	0.649
2	16	95	-0.001	0.010	-0.99	-0.71	3.356	582.0	2584.2	0.413
3	16	68	0	0.014	-1.34	-0.59	2.658	450.7	2850.6	0.700
4	16	80	1.63e-17	0.016	-1.10	-0.77	1.929	540.1	2252.7	0.560
5	16	76	8.02e-18	0.012	1.10	-0.63	3.341	764.4	2427.9	0.569
6	16	72	1.93e-18	0.010	-0.96	-0.46	2.006	554.9	2814.9	0.472
<b>Median</b>			<b>5.28e-18</b>	<b>0.012</b>	<b>-1.14</b>	<b>-0.66</b>	<b>2.515</b>	<b>576.5</b>	<b>2515.9</b>	<b>0.549</b>

Now that the model can represent the physical behaviour of the body during dynamic tasks, the model is next used to help predict from trunk accelerations what the expected ground reaction forces are. Therefore the parameters were estimated that allowed the prediction of the GRF from trunk acceleration.

#### Step 2: Parameter estimation by matching modelled versus experimental accelerations

A similar process to step 1 was undertaken for step 2. At first, the parameters found by fitting the model to the acceleration of the top mass (which is representative of the trunk accelerometer) was very successful at fitting the acceleration but these parameters did not

accurately represent the ground reaction forces. Following 1000's of iterations the GRF was accurately produced. This demonstrates that the model can predict GRF by fitting the acceleration of the top mass (Figure 14). This is evidence that the trunk-worn accelerometer can be used to predict mechanical loading (GRF) experienced by the body. However, at present it is computationally expensive and further research will be required to improve the speed in which these optimal parameters are discovered.



*Figure 14. Validation that the acceleration of the top mass of the model can be used to predict the ground reaction force. The model (blue) accurately mirrors the experimentally measured accelerations and GRF (red).*

#### 5.2.2.5 Summary

The model has taught us that for dynamic tasks, the subject characteristics are not very sensitive to change so to estimate GRF there is no need for any player specific calibration, such as everyone doing some specific tasks once they put on the trunk accelerometer. On the other hand the model was very sensitive to the type of task being observed. Nevertheless this is not a major concern because the tasks of interest from an overload and injury perspective have similar acceleration and deceleration profiles. Getting good quality acceleration signals to estimate the state of the body at contact is important and could be improved with accelerometers that have a higher sampling rate.

This study has overcome the conceptual and biomechanical challenges to estimate the external forces acting on the body from measuring trunk mounted accelerations. This required the modelling of multi-body dynamics. It is no longer necessary to try to associate segmental accelerations with whole body loading in an arbitrary way, for example using

linear regression approaches as have been used in the literature. This study has instead determined the true biomechanical relationship between trunk acceleration and GRF (see appendix 2 for full details).

### 5.3 Study 2

Study two builds on the analysis from Study 1 by investigating elite players' accelerometer signals in a retrospective case-control hamstring injury scenario. This study provides an exploratory examination of acceleration data to demonstrate how acceleration signals may be used to investigate player load and overload in a football context.

#### 5.3.1 A brief introduction

Hamstring strain injury can result in a variety of functional deficiencies, altering aspects such as motor control (Cameron et al., 2003), activation patterns (Sole et al., 2011), isokinetic torque development (Opar et al., 2013) and load distribution during contraction (Silder et al., 2010). They have also been associated to movements involving high acceleration or deceleration of the body (sprinting), and in a recent biomechanical analysis of an actual injurious event the prominent role of excessive loading rates was confirmed (Schache et al., 2009). In elite football clubs the most common way of evaluating hamstring risk is by observing asymmetries in loading between the injured limb and the contralateral side. For example, sprint tests on a non-motorized treadmill (NMT) have revealed that previously injured players can achieve their pre-injured levels of speed, albeit whilst employing compensation mechanisms (Brughelli et al., 2010). If a treadmill-based sprint test can reveal asymmetry in force generation post-injury, then it may also be possible to characterise injured from uninjured players prior to the injured players getting a hamstring injury based on their rate of force development as measured from trunk-worn commercial accelerometers.

#### 5.3.2 Methods

Nine elite players from one premier-league football club experienced a hamstring injury in the time period from 2012-present. Seven players injured their biceps femoris muscle and one player the semitendinosus and semimembranosus muscles and they were diagnosed as either

grade I or grade II tears. Three injured players were defenders, five were midfielders and one was a forward, injuries occurred during either sprinting or stretching.

Trunk acceleration data (STATSports) was retrospectively collected from training sessions in the three weeks prior to a player getting a hamstring injury. Upon injury, data for uninjured players who were matched as closely as possible by training undertaken and position, was also collected. A summary of the activities undertaken by the injured and uninjured players in the three weeks prior to injury is provided in Table 3. As Table 3 shows, a good match between players was generally observed. The training loads and matches played were similar throughout the three weeks.

*Table 3. A coloured visualisation of players' activity three weeks prior to hamstring injury.*

Pair	Group	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21
1	Inj																					
1	Con																					
2	Inj																					
2	Con																					
3	Inj																					
3	Con																					
4	Inj																					
4	Con																					
5	Inj																					
5	Con																					
6	Inj																					
6	Con																					
7	Inj																					
7	Con																					
8	Inj																					
8	Con																					
9	Inj																					
9	Con																					

Training
 Game played/minutes
 International game
 Off/no training

R

 Rehab

Game not on squad
 Game on bench/minutes
 International duty
 Away training

To provide an example of the complexity of acceleration signals, an example acceleration profile over one full training session is shown below (Figure 15). This figure shows that the commercial accelerometer captures the different acceleration intensities experienced by the player during the session in three different directions.

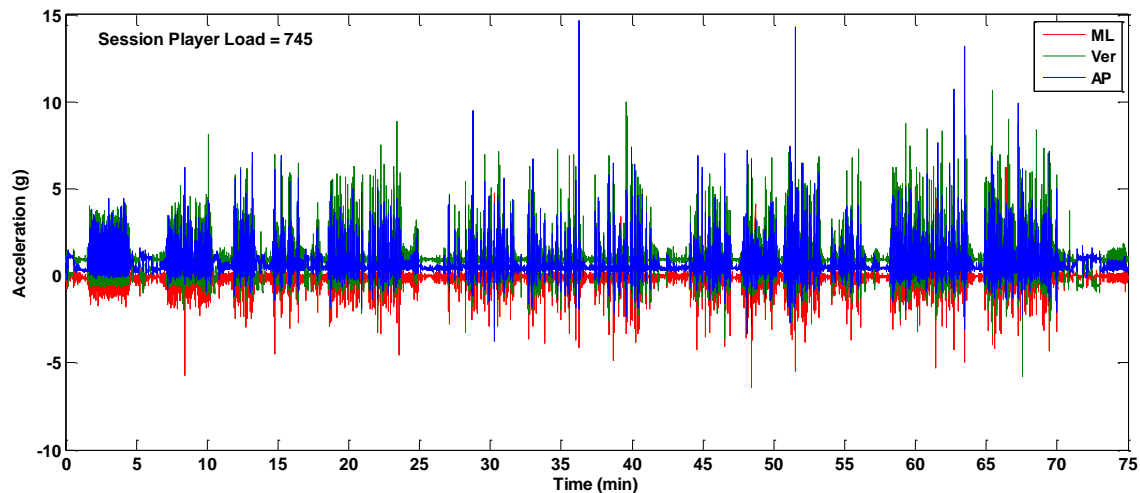


Figure 15. An example acceleration signal from a 75 minute training session for one player. Medio-lateral (ML – red), Vertical (Ver – Green) and anterior-posterior (AP - blue) accelerations are shown. These three accelerations can be combined into a resultant acceleration vector for further analysis.

From a player load perspective, it is important that the high accelerations experienced by the player are retained for further analysis whilst the less relevant “noise” within the signal, such as when the player is inactive, standing still or not being substantially loaded is removed as these accelerations are unlikely to provide meaningful information about hamstrings injury (see Figure 2 for an analogy). To characterise the accelerations experienced during dynamic tasks a commercial accelerometer was used to collect data from multiple ground contacts in a lab, and this was compared to the measured gold standard ground reaction forces

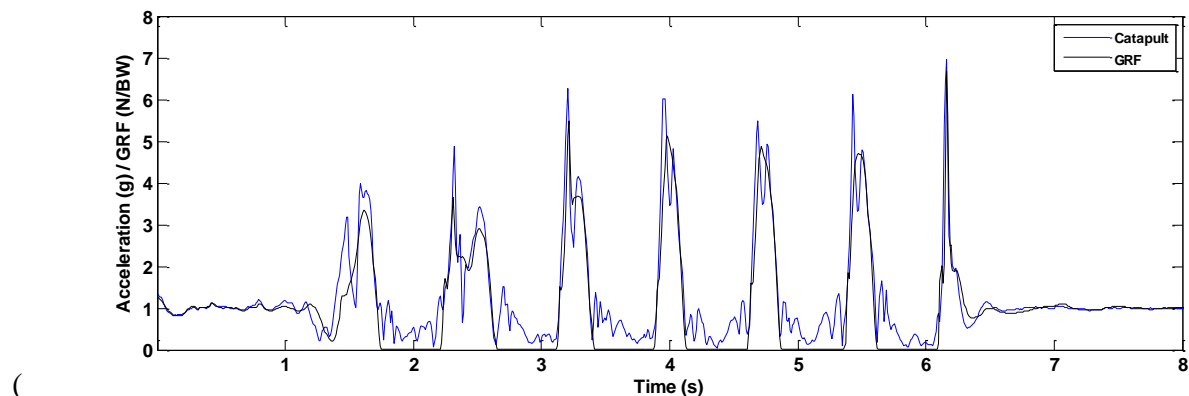


Figure 16).

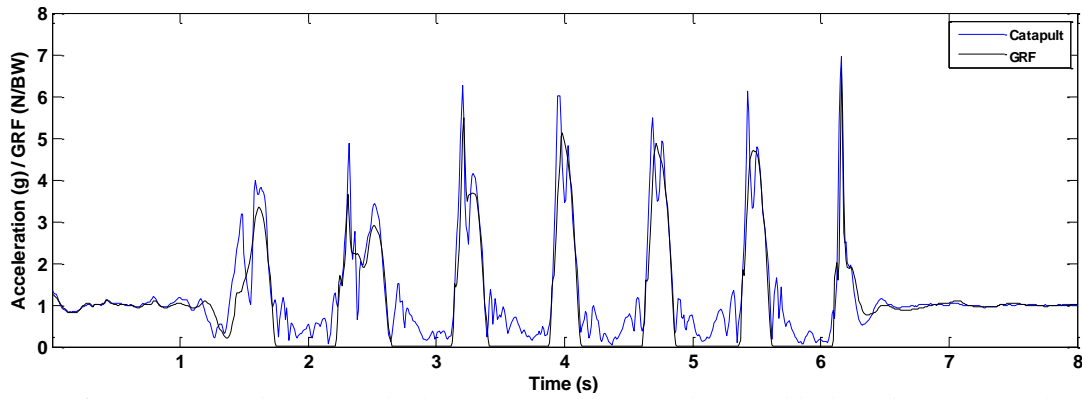


Figure 16. A comparison between multiple contacts with a force platform (black) and a commercial acceleration signal.

We therefore used the following criteria to select the relevant acceleration signals from the resultant acceleration signals of each player from each training session:

1. The acceleration signal was greater than 1 g for a minimum of 100 ms.
2. The acceleration signal reached a peak of at least 1.5 g having also met criteria #1.

An example acceleration signal meeting the above criteria is shown in Figure 17. Approximately 25-30% of all acceleration signals from all players and sessions met the above criteria.

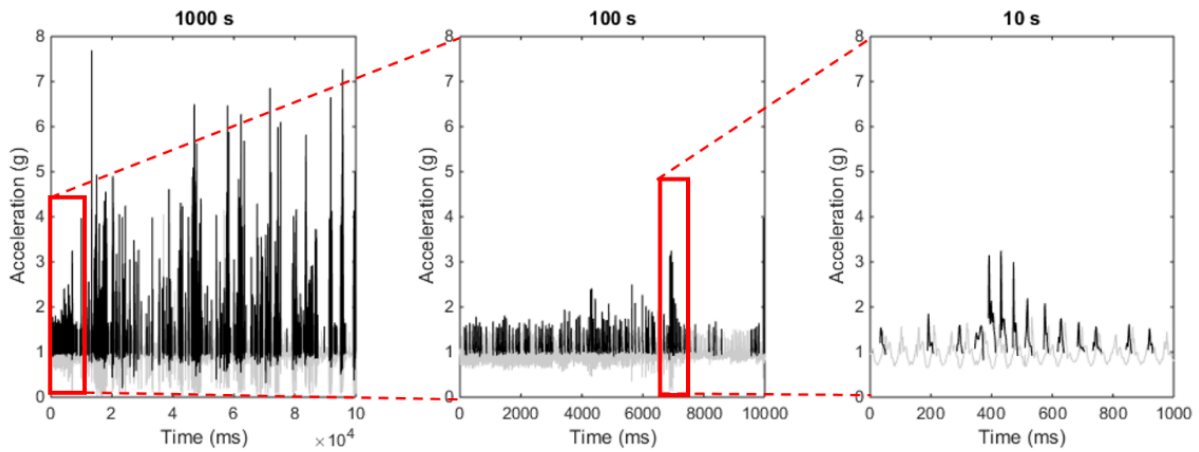


Figure 17. Example acceleration profiles from a commercial accelerometer. 1000 s of acceleration (left) is shown for the raw acceleration (light grey) and chosen acceleration signals according to the previously described criteria (black). The left image is then enhanced to show 100 s (centre) and 10 s (right).

All acceleration data meeting the above criteria were retained for further analysis. In some sessions particularly long acceleration signals were observed which had met both criteria #1 and #2. These signals were likely either the result of a faulty accelerometer or cross-talk between the acceleration components. As these were not expected to be representative of a relevant mechanical loading profile, these signals were not retained for further analysis.



Acceleration signals between 100-300 ms (the length of a typical ground contact phase) were examined further and the rate of acceleration (rate of loading) was calculated by differentiating the acceleration profile then averaging the differentiated signal from the start until the peak acceleration. Approximately 34,000 acceleration profiles for the control group and 36,000 acceleration profiles for the injured group were analysed per player. The median loading rate for each player was obtained. Statistical analysis was not undertaken across pairs due to the small sample size.

### 5.3.3 Results

Figure 18 shows the frequency distribution of contacts for different loading rates for all pairs of players. The average median loading rate was higher in the control group ( $28 \pm 9 \text{ g s}^{-1}$ ) than the injured group ( $22 \pm 9 \text{ g s}^{-1}$ ) indicating that the control group experienced a greater rate of force development. An example acceleration profile showing the different loading rates is shown in Figure 19.

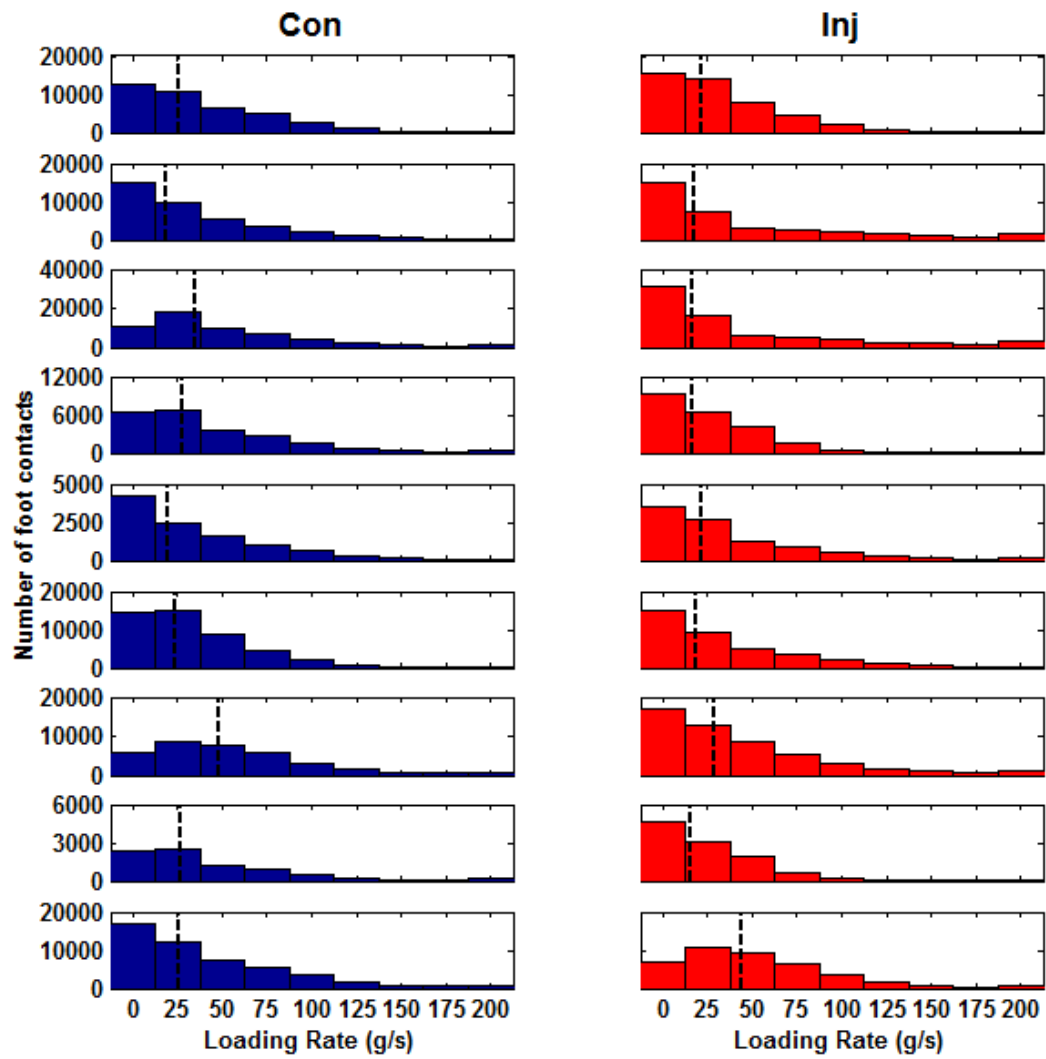


Figure 18. A histogram of loading rates for the control (CON) and injured (INJ) groups. The median loading rate is shown as a black vertical line.

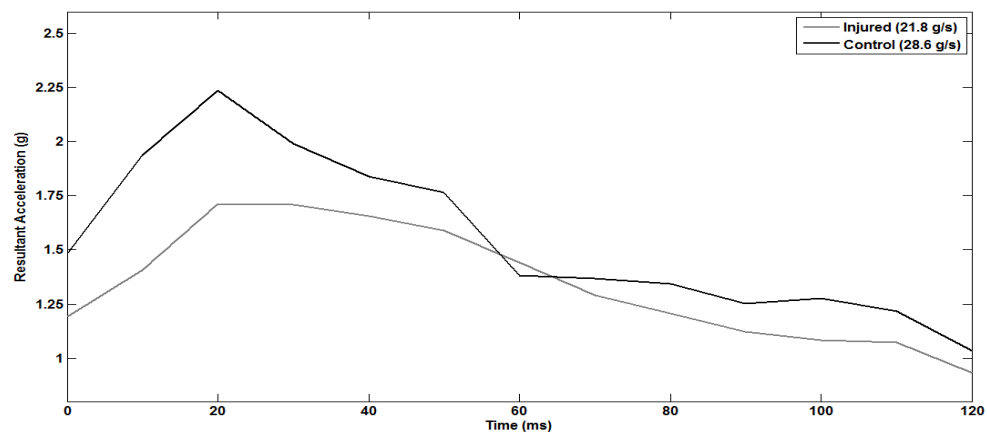


Figure 19. An example acceleration profile highlighting the difference in loading rate between the injured (grey) group and the control group (black).

### 5.3.4 Discussion

This study provided an initial exploration of commercial accelerometry data in a retrospective case-control hamstring injury scenario. A greater average median loading rate was observed in the control group compared to the injured group and this was evidenced in seven out of nine player pairs.

Trunk-worn commercial accelerometry signals have the potential to provide meaningful injury relevant information. The observation that the control group had greater loading rates than the injured group is novel and suggests that differences in behaviour may be observable prior to the occurrence of an injury. Speculatively, one might attempt to justify these results in the context of preventative behaviour in the injured players. If one assumes that the load experienced by players across the same training session should be similar, then the lower loading rates experienced would be indicative of altered behaviour in the injured group. This however should be confirmed in future studies. The behaviours associated with lower loading rates, are generally seen as beneficial, such as having a decreased vertical stiffness of the body (Milner et al., 2006) however, given that lower loading rates led to an undesirable injury outcome in seven out of nine players, a decreased loading rate could also reflect behavioural alterations consciously or unconsciously in task intensity, movement kinematics e.g. more ankle / knee flexion upon landing or neuromuscular control e.g. the inability to adequately stiffen the body. Given that a previous hamstring injury is highly predictive of injury recurrence (Engebretsen et al., 2010) it is also possible that these behaviours could be observed in a post-injury return to play biomechanical assessment.

The loading rate detected by the commercial accelerometers underestimates the actual loading rate (see study 1 – experimental observations), however given that it is relative changes within the same units that are examined in this study, accurate absolute loading rate magnitudes are not essential. Characterisation of commercial accelerometer loading rates compared to the lab accelerometer loading rates were demonstrated in study one. Greater sample rates (increasing from 100-1000 Hz) would be required from the commercial accelerometer to accurately capture the true loading rates observed, and predict GRF from the model described in section 5.2.2. The characterisation of loading rate from GPS measured data would not be possible due to the poor temporal sample rate of this signal. However given that the above results appear to indicate a behavioural difference in the players prior to injury there may be relevant behavioural information such as the number of sprints or high

intensity bursts that could also be observed that would complement the acceleration information. Given that the accelerometers detected a reduction in median loading rate in injured players there may also be correlates to decreased metabolic loading rates that are detectable with GPS analysis.

The criteria applied to the acceleration data in this study showed that only around 25-30% of the acceleration signal was kept for further analysis. This highlights that the accelerations measured contain a lot of data that is unlikely to be relevant to player overload or injury. This can have important practical consequences as a typical session would capture around 100 mb data per player or 2 Tb data per team per session. This volume of data requires quite some processing power and is not easily handled in typical software such as Microsoft Excel. If the application of simple criteria allows file sizes to reduce to around a quarter then this makes further analysis much more manageable. The other consequence of analysing only 25% of the captured data is that summary statistics such as the commonly used PlayerLoad<sup>TM</sup> which use the entire acceleration profile would contain a lot of “noise”, making its’ use less clear for a predictive injury context.

In study 2 we used an acceleration threshold to remove acceleration signals that we did not deem relevant to the acquired injury. The consequence of this was that we were left with short high-intensity bursts of acceleration that likely represent single contacts with the ground. Whilst the model developed in study 1 is capable of calculating ground reaction forces from single contacts it has not been implemented in a way which will reproduce estimated ground reaction forces for the continuous acceleration trace. Whilst this may seem desirable it is perhaps the accurate modelling of the high intensity accelerations that is likely to reveal the meaningful injury related information.

This study was a retrospective case-control study. The benefits of this study type are that you can build up a database of injuries in a time-efficient way compared to a prospective study design. A retrospective study allows the exploration of differences between the injured and uninjured players prior to injury and so is a good model that football clubs could adopt for interrogation of their own data and injuries. More data of this type is required for different injuries so that clubs can have some injury indicators against which they could profile their players and implement preventative measures.

Finally, the results of this study are limited to a small number of injured players from one football club therefore the generalisability of these results across other clubs and other injuries is unknown.

## **6.0 Implications for elite football**

### **6.1 Implications**

There are a number of implications from this report that are applicable to elite football clubs. Foremost is that this report should change the perceptions of how trunk accelerometry is related to player loading. Whereas previous research has focussed largely on metabolic aspects of player loading (see literature review) or have assumed a direct relationship between trunk acceleration and external forces acting on the body, this report can serve to generate a change in mentality around what can realistically be expected from accelerometry and in particular how this differs from GPS-based data.

This report then provides a starting point for the development of appropriate software that will allow professional clubs to use their accelerometry data with a valid model of mechanical loading. As many football clubs already have access to accelerometer data from their players, these studies can empower practitioners in a number of ways, including:

1. Providing an understanding of how trunk-worn accelerations relate to mechanical loading of the musculoskeletal system,
2. Providing a theoretically and biomechanically validated method to estimate mechanical loading,
3. Providing an example of how one can use the monitoring of mechanical loading to investigate injury and overload data

This empowerment will provide tangible benefits to teams who through appropriate use of accelerometer data would be able to:

1. Plan and monitor training intensity from a mechanical (tissue) loading rather than metabolic loading perspective only,
2. Monitor and evaluate loading profiles into and out of games,

3. Conduct their own local investigations into how mechanical load profiles of individual players relate to the injuries they experience

**Some answers to typical questions that practitioners might ask are provided below.**

*Question. How does trunk acceleration relate to mechanical loading on the body?*

Answer. Trunk acceleration only gives you information about one part of a complex multi-segment system so on its' own a direct link is difficult to make. Whilst trunk accelerations are related to whole body accelerations and therefore mechanical loading, the interaction of different body segments also has to be taken into account for trunk accelerations to predict mechanical loading accurately.

*Question. Are the commercially available units able to provide as relevant information as more expensive accelerometers?*

Answer: As long as commercial accelerometer units continue to sample at 100 Hz then a commercial trunk acceleration signal is unlikely to estimate loading well because it tends to underestimate the loading rate experienced by players.

*Question. What do I need to be aware of to get good acceleration data?*

Answer: To use trunk acceleration signals to predict mechanical loading no individual player calibration or correction is needed, except the player mass. Predicted loading is not sensitive to between subject differences. It is however very important to get the best quality acceleration signal possible (>250-500 Hz would be best) so that the state of the body at contact can be estimated. Secure the accelerometer as tightly as possible to minimize movement artefact – the more secure the accelerometer the better the signal will represent the player's behaviour.

*Question. Is the trunk the best place to put an accelerometer?*

Answer. Accelerations measured at the trunk tell us something about how we interact with the environment and how the external forces we experience pass through the body and put internal tissues under stress. Putting an accelerometer on the trunk accelerometer is likely to be the optimal location given that securing an accelerometer to the pelvis (the next best place) is likely to be difficult.

*Question. Are trunk accelerations valid for player overload and/or injury prediction?*

Answer. At this stage it is not known what variable/s from trunk accelerations might predict injury. This report shows how trunk accelerations can estimate mechanical load and it also gives an example of how trunk accelerations might be used in an injury context, however both areas are highly novel and require further work for the full benefits to be established. Trunk accelerometers provide a lot of signal that is unlikely to be meaningful if you use it all. This report suggests that focussing on higher magnitude accelerations and the rate of loading is likely to be more meaningful.

*Question. How might acceleration data be used for injury analysis?*

Answer. As the trunk acceleration characteristics of injured players are as of yet unknown a retrospective case-control analysis would allow characteristics between injured and uninjured players to be explored.

## 6.2 Reflections on PlayerLoad™

As described in the literature review the variable PlayerLoad™ (the sum of the resultant differentiated acceleration) has been used as a summary measure against which load related variables have been correlated. This report deliberately avoided using PlayerLoad™ as the starting point or focus for this research because there is no clear understanding what this variable really measures or represents. All of the correlational studies described are attempting to find the usefulness of this measure. Recent correspondence with the PlayerLoad™ trademark holder indicated that they do not expect relationships between such measures and whole body loading (ground reaction forces) yet in the first study we have demonstrated the mathematical relationship between trunk acceleration and whole body loading and highlighted the differences between the trunk segmental acceleration and the whole body centre of mass acceleration which is directly related to the ground reaction force.

From study 2 we found that only approximately 25% of the measured acceleration signal provides meaningful overload / injury-related information. We would therefore question how sensitive and insightful a cumulative measure such as PlayerLoad™ can be in an overload / injury context given that prolonged walking could quite conceivably have the same PlayerLoad™ as a few short sprints. The inability of this summary measure to distinguish task intensity and duration means that it is unlikely to be able to represent the mechanical demands on the body leading to overload / injury.

## **7.0 Limitations**

All studies capturing accelerometer data are reliant on the quality of the fixing of the accelerometers for the quality of the acceleration signals achieved. For all of the accelerometers utmost care was taken with mounting. The commercial unit was mounted in the manufacturer recommended chest harness yet there is likely to be some relative movement of the accelerometer relative to the body. As this is the way in which such data is collected in football clubs no additional effort was made to reduce the effects of any relative movement. Consequently the representation of the trunk acceleration may be influenced by this artefact but the validity of the measure in the context of the applied football environment is high.

The model used in study one is considered a “passive” two-mass model. This can be likened to the model being dropped onto the ground and its’ response being observed. In reality a human ground contact should be considered as passive around impact, but the push-off from the ground is an active process. To more accurately fit ground reaction forces, the model may either need an active component, allowing it to push off the ground and may also require additional masses to be able to more accurately model the complex behaviour of the interaction of the body with the ground. Furthermore the model will predict ground reaction forces based on the input characteristics to the model including player-specific stiffness-related variables. To predict ground reaction forces for an individual player it will assume that the player has a constant stiffness. This may not be the case for example if the player gets a knock and chooses to reduce loading on the body by altering joint stiffnesses. It is not known how players input parameters to the model would change over a prolonged period of time e.g. a season and therefore how often the parameters would need to be validated or the effect this would have on the predicted ground reaction forces.

## **8.0 Future Directions**

In study one, the tibial accelerometer was shown not to reflect the whole body CoM acceleration and therefore ground reaction forces, there is still likely to be relevant injury related information that can be gained from such an accelerometer. The usefulness of a tibial accelerometer in an injury context should therefore not be discounted for example for metatarsal fractures and shin splints, but would require an accelerometer with an approximate



range of +/- 24 g and a sample rate > 1000 Hz to appropriately capture the high frequency accelerations this segment experiences.

The usefulness of trunk acceleration for predicting player overload or injury occurrence is likely to be based on the capturing and interrogation of a large database of a variety of injuries. This could either be done in a prospective study design or perhaps more appropriately in a retrospective case-control analysis. A collaboration with elite football clubs across Europe to gather such data could allow this.

In this study the biomechanical and mathematical relationships were established to allow informed interpretation of data from trunk mounted accelerometers through the prediction of external forces on the body. Improvements in accelerometer technology will allow better quality trunk accelerations to be measured and these accelerations will provide a better input to the model. The model can be further refined, developed/integrated into software that allows the prediction of external forces acting on players based on their trunk acceleration profiles. Once ground reaction forces can be reliably estimated, the next steps would be to move towards the determination of joint specific loading.

Future exploration of the sensitivity of mechanical load to variations in training surface, training periodisation, footwear, pitch size, small versus large sided games will help practitioners to distinguish mechanical from metabolic load / overload indicators. This will ultimately help them to differentially monitor and adjust players' exposure to metabolic and mechanical loads.

## 9.0 References

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## **10.0 Appendices**

Attached separately are two appendices:

10.1 Appendix 1. Calibration and synchronisation of the accelerometers.

10.2 Appendix 2. Spring-mass damper systems from scratch.