Human Running Coordination:

Illuminating a Complex Phenomenon

by

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Abstract

Humans are highly accomplished endurance runners. Yet despite this proficiency, injury rates remain high amongst running populations. Most running-related injuries are overuse in nature, driven by excessive exposure to repetitive impact loadings in the absence of sufficient recovery. The key provocative event, within the running cycle, occurs during contact between the running foot and the running surface. During ground contact, runners manipulate tissue behaviours and limb dynamics, via skilled neuromuscular coordination, to mitigate the threats and capitalise on the opportunities presented by the mechanical shock of impact.

Previous research, however, suggests movement coordination changes subsequent to acute and/or chronic fatigue. In running contexts prior investigations into this phenomenon have variously employed different permutations of measurement tools, fatiguing protocols and assessments, analysis methods and outcome measures. Accordingly, within this field, investigative findings are frequently conflicting. Subsequently, to date, there is no consensus on whether moderate fatigue, such as would be regularly experienced during typical running training programmes, inhibits runners' capacity to proficiently manage repetitive impact shocks.

This thesis is built around 3 related observations. Firstly, to negotiate the challenges imposed by ground contact, runners coordinate movement dynamics to alleviate threats and exploit opportunities. Secondly, moderate fatigue may interfere with coordinated control to the extent that quantifiable facets of the impact event change as

runs progress. Thirdly, how different impact-related metrics change, in tandem with modulating fatigue status, may provide novel insights into the underpinning nature of the running coordination phenomenon. The thesis objectives, guided by these observations, were subsequently to:

i. Examine the influence of running-induced fatigue on impact metrics commonly cited as sensitive indicators of running proficiency

ii. Investigate how these metrics changed over the course of a short, moderately fatiguing run

iii. Determine whether these impact-related metrics changed, in response to moderate fatigue, in a manner that was common and consistent across the running cohort

To accomplish these objectives, 15 regular runners participated in two distinct, yet integrated, running protocols conducted at different intensities, durations and levels of exertion. Runners were fitted with 4 low-mass, integrated measurement sensors. 4 metrics, each advocated as a sensitive measure of running-induced fatigue, were derived from the experimental dataset. Subsequent analyses reveals that:

i. Change was pervasive, across selected metrics, time-points and sensor locations, and in both fatigued and non-fatigued conditions

ii. Running coordination outputs, assessed via changes to surrogate measures of impact shock, exhibited substantial response variability to context-specific conditions

iii. Although prior research has predominantly focussed on lower-limb kinematic changes, in these investigations, high frequencies of meaningful change were also observed at sensor locations on the upper and lower trunk

iv. Vertical tibial deceleration, historically the most extensively investigated acceleration metric, did not change coherently and consistently between conditions

v. Over the duration of the fatiguing run, the selected metrics did not conform to a common pattern of change

The theoretical and experimental conclusions of this thesis emphasise the problems and limitations inherent in investigating such a nuanced, multi-faceted and ever adapting phenomenon, as running coordination, using isolated metrics within conventional, and typically low powered, experimental designs. The diversity of responses, displayed by individual runners, in tandem with the evident sensitivity of impact metrics to specific conditions, suggests that, in the future, greater insights and understanding will demand more insightful consideration of metrics, analysis methods and definitional clarity, and more stringent quantification of fatiguing protocols.

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Glossary of Abbreviations

2-D – Two dimensional

3-D – Three dimensional

ANOVA – Analysis of variance

AP – Anteroposterior

C7 – 7th cervical vertebrae

CoM – Centre of Mass

ES – Effect size

g – with one gravitational unit = 9.81 m/s^2

GRF – Ground Reaction Force

Hz – Hertz, a measure of frequency

ICC – Inter-class correlation

IMU – Inertial Measurement Unit

Kg - Kilogram

km/h – Kilometres per hour

MDD – Minimal detectable difference

MDC – Minimum detectable change

m/s – Metres per second

ML – Mediolateral

mm - Millimetres

MTSS – Medial tibial stress syndrome

N – Newton, a unit of force

PFP – Patellofemoral Pain

PTA – Peak tibial acceleration

ROM – Range of Motion

RRI – Running related injury

SA – Shock attenuation

SD – Standard Deviation

SI - International System of Units

SPSS – Statistical Package for the Social Sciences

ST – Stride time

STEMH - Science, Technology, Engineering, Medicine and Health

SWC – Smallest Worthwhile Change

UCLan – University of Central Lancashire

V3D – Visual 3D. A biomechanics data management system

Chapter 1:

Introduction

1.1 Why Running? Why Coordination?

Whilst in comparison to other mammals we are not especially strong, swift, explosive or supple, we are amongst Nature's most superbly well-adapted endurance runners. Throughout human existence running has been an essential prerequisite for survival. A means to travel, a means to hunt, and a means to escape when hunted (Bramble and Liberman, 2004). The mutually entwined inter-relationships between running and evolutionary survival extends back beyond the beginning of the *homo* genus, and down through our long line of predecessor species. How modern humans now run is shaped by our neural and biological architectures; yet, running was a driving evolutionary force that contributed to shaping these same structures. We evolved to run and running contributed to how we evolved. This entwined evolutionary legacy is reflected in many deeply and irrevocably embedded neurological and biological design features; features which underpin and promote our exceptional running abilities. The same is not true for a golf swing, or a tennis serve, or the backstroke, or track cycling. In fact, from this perspective, there is a clear distinction between running, and how running is coordinated, and the coordination of the majority of sports-related movement skills.

Today, running remains a pervasive activity across human cultures and a cornerstone of contemporary health, fitness and sporting activities. We learn to run as a natural outcome of normal development, without specific practices or specialist coaching, to the extent that, conventionally, we regard running as a mundane, naturally acquired movement skill. In effect, running permeates so many dimensions of normal everyday existence, that we tend to overlook just how remarkably competent we are as runners, and we rarely consider the inherent uniqueness of prolonged, upright, bipedal running within the mammalian kingdom. Yet our alternating single stance running gait demands we bounce from foot to foot, at speed, while retaining stability, steering direction of travel, powering the upcoming stride, whilst simultaneously precariously balancing our fragile brains in the very position in which they are most vulnerable to falling injury. Nevertheless, we typically execute these complex, concurrent feats with ease and within the abbreviated timeframes afforded by short, violent ground contacts.

Within sporting domains, the term coordination is used so frequently, in so many ways and in so many contexts, that it does not have a clear meaning or single all-encompassing shared definition. Instead the sports coach, the biomechanist, the physiotherapist, the skills acquisition professional, commonly all view the coordination phenomenon through different domain-specific lens. In its sparsest interpretation, coordination simply captures the concept of a changing relationship between two variables. In the more complex context of biological systems, perhaps the most concise expression of the essence of coordination is that it demands that multiple components work together to realise an objective (Diedrichsen et al, 2010).

Another factor, further obscuring our perception of the coordination phenomenon, is that in practical contexts, running coordination is an exceedingly difficult capacity to meaningfully measure. Other performance-related capacities, in contrast, are readily evaluated. Thus, although different means of quantifying strength, speed, various endurance measures, and a host of other facets of overall performance, are readily accessible; there is no currently available test capable of adequately quantifying running coordination. Consequently, in practical training contexts, coordination remains largely overlooked; a victim of a focussing illusion, whereby we disproportionately emphasise the measurable, while simultaneously discounting the difficult to quantify. Accordingly, in performance conditioning cultures, although we have substantial evidence bases documenting how to specifically target dimensions of performance, such as strength, endurance, agility, speed and speed endurance; conversely, we have precious little evidence-led information illustrating how practitioners should, or could, design training interventions targeting running coordination. In research contexts, various dimensions of running coordination have previously been quantified using diverse measures, techniques and measurement technologies (DeLeo, Dierks, Ferber, and Davis, 2004; Bartlett, 2007). Crucially, however, all such procedures require specialist equipment, and none are practically amenable. Furthermore, even with access to the appropriate assessment technology, it remains unclear which specific dimensions of running coordination are the most relevant to running proficiency and running resilience.

A further feature of running coordination is that it appears vulnerable to multiple modulating influences. Aging, disease states, injuries, illnesses and insults to either neural or peripheral tissues, all hold the potential to impede fluent movement execution. In running contexts, perhaps the most notable and frequently encountered impediment to proficient coordination is fatigue. That extreme fatigue negatively influences coordination seems clear. Images of Gabriela Andersen-Schiess stumbling into the Olympic stadium in Los Angeles at the end of the first women's Olympic marathon in 1984, or footage of Sian Welch and Wendy Ingraham running, falling and finally crawling across the finish line of the 1997 World Ironman championship in Kona, Hawaii, serve as classic illustrations of how excessive fatigue severely impedes running coordination. These examples are obvious, because they are so extreme. In less extreme cases, however, fatigue-driven erosion of running coordination is not so readily, visually detectable. The premise of this thesis is that new technologies offer opportunities to detect fatigue-induced changes to running coordination, and subsequently provide opportunities to gain a deeper appreciation of the phenomenon by better understanding how it deteriorates.

Perhaps surprisingly given running's priority as a pivotal sporting and health-bestowing activity, it is currently unclear how fatigue impacts different facets of running coordination. This information, however, seems both pertinent and important to professional and recreational runners, and their support staff. If, for example, even moderate levels of fatigue compromise running coordination, and if runners persistently run under moderately fatigued conditions, what are the likely consequences? Is running under fatigued conditions problematic? If so, how do we know?

1.1.1 Does new technology provide an opportunity?

Rapid advances in technology and instrumentation, over recent decades, have enabled the quantification, and easy collation, of many previously unavailable metrics. Within many professional sporting contexts, this has led to a dramatic increase in the availability of accessible quantifiable measures, and an exponential growth in the volume of acquired data. Perhaps the most obvious illustration is the sudden proliferation of GPS systems to quantify external movement loads. GPS use was virtually non-existent 20 years ago, yet now is pervasive across a diversity of sports (Cummins, Orr, O'Connor, and West, 2013). Similarly, more recently, progressively smaller, and more affordable, integrated measurement technologies ensure that all of us have, if we so wish, a ready access to an expanse of suddenly available, kinematic and biological data.

Such technologies have found application within running training cultures. Endurance training cultures, historically, have always prioritised diligent record keeping. Yet the transition from the *pen and paper* recording of basic metrics –times, distances, paces, bodyweight, and Rates of Perceived Exertion–, has exponentially evolved to the extent that, today, an expanse of objective data is easily measured, stored and longitudinally tracked. In fact, access to a spectrum of more complete, more easily collated, more easily managed, and more easily presented, scrutinised and analysed, running-related data has become the ‘new normal’. Recent technological advances have, subsequently, within a handful of years (albeit not universally), dramatically changed many dimensions of running training and running coaching culture.

Currently, accordingly, we have access to a diverse toolbox of measurement devices, yet still do not have a comprehensive, evidence-led explanatory framework through which to conceptualise the essence of running coordination. In endeavouring to develop greater understanding and insight, the broad purpose of this thesis is to, conceptually and experimentally, investigate the underlying nature of the running coordination phenomenon, by examining how it deteriorates subsequent to fatigue.

1.1.2 Quantifying the shock of impact

As will be discussed in the literature review, the highest mechanical stressors applied to the runner, during running, coincide with the ground contact event. Upon ground contact a shockwave of impact travels upwards through the body, towards the brain. Importantly, this impact is associated with a variety of negative outcomes. A diversity of tools and measures are habitually used to quantify the consequences of impact, ground reaction forces assessment via force plates, for example. Yet it is only in recent years that certain wireless-enabled technologies have become sufficiently compact to be comfortably worn by runners. Contemporary lightweight, triaxial accelerometer technologies subsequently offer the opportunity to, in an unobtrusive manner, assess how the accelerations associated with impact change over the course of a fatiguing run. Given the utility of accelerometer technology, recent investigations have employed lightweight, triaxial accelerometers in running contexts Yet other research has used the same technology to quantify metrics, other than acceleration, but similarly relating to the impact shocks imposed during running activity. To date, however, although each of these measures are advocated as surrogates of coordination, given that each is suggested to change subsequent to fatigue, whether these measures all provide accurate indications of fatigue-induced deterioration of coordination remains unclear. Similarly, although each of these metrics has been variously supported within the literature, it remains unknown whether these metrics provide similar indications of fatigue-induced change to running coordination. Specifically, it is not clear if these metrics provide equivalent, or very different, snapshots of fatigue status.

1.2 Objectives of this thesis

The overarching ambition of this thesis is to explore, through literature review and novel experimentation, dimensions of the running coordination phenomenon. More specifically, to investigate how coordination changes subsequent to running-induced fatigue. The steps taken in endeavouring to achieve this ambition were:

1. Firstly, to review and articulate the key, conventionally underappreciated, neural and biological processes which underpin and enable running coordination
2. Secondly, to empirically examine the influence of running-induced fatigue on dimensions of running coordination via the use of commonly recommended acceleration-derived assessment metrics
3. Finally, to synthesise the subsequent learning outcomes against stated thesis objectives

The specific experimental objectives designed to add insights to this broad ambition were three-fold:

1. To investigate the influence of fatigue on running coordination on short pre- and post-fatigue intervals, via the evaluation of a set of surrogate measures, each advocated within the literature as indicative of running fatigue
2. To examine and compare how these measures change, and whether these changes occur in tandem, or asynchronously, over the course of a short, moderately fatiguing run
3. To determine, through analysis and comparison of runner-specific data, whether fatigue-induced deteriorations to running coordination progress in a coherent and shared, or disparate and nonformulaic, manner across the running cohort

1.3 Structure of this thesis

1.3.1 The theoretical backdrop

The topic of running coordination is dauntingly expansive. Providing insight into such a broad and complex topic, clearly requires empirical investigations, but may also benefit from some evidence-based, conceptual re-evaluation and theoretical refinement; some necessary reframing of how we conventionally perceive the running coordination phenomenon. To this end, the literature review section of this thesis is composed of 4 chapters, each dealing with a distinct dimension of running coordination. Each of these chapters is presented as a stand-alone narrative, and each is structured to review the relevant available literature, whilst also offering conceptual refinements to current running coordination explanatory frameworks.

This approach provides the latitude to delve into other relevant research domains, and to investigate key dimensions of the running coordination phenomenon that remain poorly explored, and which have received little attention within the conventional sports science literature. The purpose of these chapters, subsequently, is to provide an overview of the empirical evidence, whilst also providing scope to blend emerging conceptual insights emanating from other academic domains.

With this objective in mind, Chapter 2 starts with an overview of the deep neural and biological underpinnings of running coordination. Whilst this may be criticised as overly abstract or esoteric, it was written in the expectation that a better appreciation of the neurobiological undercurrents that enable our exceptional running ability, seems likely to promote conceptual insight. Accordingly, the fundamental evolutionary objectives, and constraints, underpinning running coordination are explored within this chapter. Also outlined are the various mechanisms through which running-imposed perturbations and de-stabilisations are countered and remediated, such that safety and stability are maintained within an acceptable bandwidth, for an acceptable cost; in terms of energy expenditure and neural control commitment.

This chapter also details the distributed nature of running coordination and outlines how this multi-centre control underpins our capacity to instantaneously respond to running-imposed perturbations and de-stabilisations. Importantly, this chapter details the multi-level, multi-module collaborations that ultimately underpin running coordination and subsequently enable humans to run in such an unusual, and apparently risky, unstable, bouncing gait.

Chapter 3 specifically focuses on the processes facilitating how we learn, and subsequently embed, personalised running coordination habits and proclivities. Also described are the mechanisms through which healthy coordination habits are progressively refined, and subsequently inevitably erode in the face of the gradually accumulating wear and tear of life. In fact, the evidence presented illustrates that running coordination is a constantly adapting phenomenon, persistently modulating under the influence of multiple shifting constraints. This chapter also provides a framework for conceptualising the many routes through which life experiences, and running and injury histories, irrevocably shape individualised running coordination signatures. The chapter concludes by offering a comprehensive definition of running coordination.

Chapter 4 addresses the challenges and opportunities imposed by the repetitive, cyclical impact shocks imposed during the ground contact event. Specifically, 2 inter-related phenomena, running stiffness and impact shock absorption, are considered. Accordingly, this chapter explores the various inter-related variables, emanating from collision between the running foot and the running surface, which have previously been implicated in running performance and/or running-related injury. While the historical perspective, emerging from the literature, commonly presents the regulation of impact shock absorption as primarily a physical capacity, the range of evidence reviewed here illustrates that stiffness and shock attenuation is a closely regulated, context-specific, coordinative output. This chapter concludes with an evidence-led reconceptualization of the running stiffness construct.

The final narrative within this review section, contained in Chapter 5, examines the, under-appreciated but potentially relevant, concept of running smoothness and its diametrically opposed counterpart, kinematic jerk. Evidence, across a diversity of motor control applications, illustrates that movement smoothness changes subsequent to natural aging, general health, training and injury histories, and levels of fatigue. As smoothness reflects sensorimotor control, smoothness metrics potentially provide a quantifiable window into running coordination proficiency. Supporting this conjecture, preliminary research suggests that running proficiency is reflected in smoother running control (Hjerlac, 2004). In contrast, prior injury and/or excessively accumulating fatigue drive diminishing smoothness and increasing jerk. Relevantly, recent advances in contemporary assessment technologies provide the opportunity to sensitively detect changes in running smoothness and may, subsequently, bestow unique insights into running coordination proficiency.

At the end of this chapter, a brief summary synopsises the key outcomes of the literature review chapters. Within this synopsis a set of distinct acceleration-derived measures, all supported to varying degrees within the literature as reflecting running fatigue, are noted. These selected measures will subsequently be deployed within the experimental protocols contained in chapters 7 and 8.

1.3.2 The experimental core

Chapter 6 documents both the methodological rationale and decision-making considerations that guided experimental design, as well as the specific experimental protocol used to collate, treat, and analyse the relevant experimental data. Within this chapter the twin premises underpinning the experimental design are outlined; firstly, the premise that accumulating fatigue drives coordinative change and, secondly, that fatigue-induced changes in running coordination will be reflected in changes in how we manage different facets of the mechanical challenge imposed during ground contact. Focussing directly on the kinematic ‘outputs’ emanating from the ground contact event and examining how these outputs change over the course of a moderately fatiguing run, provides a window into how running coordination may, or may not, deteriorate subsequent to running-induced fatigue. Such an approach enables comparison between what could be considered ‘better’ coordination, i.e., coordination while fresh, and ‘worse’ coordination, i.e., coordination in the fatigued condition.

Chapter 7 focusses specifically on examining how the magnitudes of a set of related, acceleration-derived measures changed before and after a moderately fatiguing run. Hence, this chapter seeks to establish how different metrics may, or may not, differentially change subsequent to fatigue.

Chapter 8 takes a different approach by collecting and analysing acceleration data collected across the duration of a moderately intense, fatiguing run to determine whether fatigue, as evaluated via a battery of acceleration-derived metrics and a conventional Category-Ratio scale, the Borg CR-10 (Borg, 1982 & 1998).

1.3.3 Synthesis

Finally, in Chapter 9, the key themes emerging from the literature review and the findings of the empirical investigations are synthesised. The limitations of this work are identified, and potentially productive avenues of future research are suggested. Subsequently, the key messages emanating from this thesis are critically reviewed against the stated objectives. Potential associated implications both for future research and for professional practice within sports sciences are also discussed.

1.4 Conclusion

It is perhaps worth noting at the outset that the thesis title, and the inter-related topics addressed, are broad and extensive and can, potentially, be approached from a diversity of perspectives and drawing from distinct, occasionally overlapping, literatures. In acknowledging the potential breadth and multi-dimensional nature of the running coordination phenomenon, the approach adopted within this thesis placed an emphasis on an extensive review of the literature, drawing from distinct scientific disciplines. The rationale for this approach was, in effect, that the scope of the questions posed by the thesis were not directly answerable via a short series of empirical investigations and required some necessary conceptual reframing to rationalise and complement the thesis ambitions of generating new insights and understanding.

Currently, in the context of insightful running metrics, we are still struggling to discern which stride related metric, or combination of metrics, best reveals the most pertinent, running proficiency-relevant information. As a practical sports scientist, my ultimate objective is to better understand the coordination phenomenon, with the intention of using this enhanced insight to derive a more practically amenable, evidence-informed framework for conceptualising running coordination.

Chapter 2:

Uniqueness of Human Running Coordination

(This chapter was published, in large part, as "Kiely, J., & Collins, D. J. (2016). Uniqueness of human running coordination: the integration of modern and ancient evolutionary innovations. *Frontiers in psychology*, *7*, 262.)

2.1 Introduction: Human Running Ability

Running is such a pervasive activity, across human cultures, that we often fail to appreciate how extraordinarily gifted we are as runners. We lack the swiftness of cheetahs; the power of charging bulls; the agility of cats. Yet we are exceptional running generalists, capable of running at moderate speeds for prolonged periods; readily modulating pace without changing fundamental gait pattern; seamlessly adapting to varying terrains and climatic conditions (Bramble and Lieberman, 2004).

Unlike other mammals who—thanks to embedded fixed action patterns and rapidly myelinating nervous systems—can quickly execute a limited repertoire of stereotypical movements, we remain helpless for prolonged periods after birth (Langen et al., 2011; Miller et al., 2012). This initial early life deficit, however, underpins a remarkable, slowly emerging coordinative proficiency. A proficiency, ultimately, enabling us to master a staggering diversity of skills unrivalled within the animal kingdom.

A sometimes overlooked distinction between running and more modern sporting movements is that running has been essential for survival across the expanse of hominid evolution (Bramble and Lieberman, 2004). How we run is shaped by our anatomy, neurology, and physiology. Yet, in a mutually reciprocating manner, how our long line of hominid ancestors once ran similarly contributed to sculpting current structural, neurological, and biological characteristics (Bramble and Lieberman, 2004). Throughout the countless blind “trial and error” experimental iterations of evolutionary deep-time, the mutually entangled co-evolution of bio-structures and running ability has led to the creation of deeply integrated coordinative solutions to the running challenge. Yet despite the ubiquity of running within human cultures, and the everyday use of the term “coordination” within sporting domains, the running coordination phenomenon remains vaguely explored, perhaps overlooked as a key facilitator of our species unique running abilities.

Conventionally, movement coordination is viewed through the lens of one of a number of competing theories—Dynamical Systems; Equilibrium Point Hypothesis; Optimal Feedback Control— all of which exhibit both substantial overlap, and points of distinction; each variously explaining many, but not all, observable behaviours (Todorov, 2004). Discerning between these theories, as they apply to running, however, lies outside of the focus of this thesis. Instead, here, the objective of this chapter is to describe how the inter-mingling of both modern and ancient evolutionary innovations blend to underpin and facilitate human running coordination.

2.2 The Evolutionary Purpose of Coordination

Evolutionary survival demands that biological systems—operating in unpredictable environments using unreliable components and finite energy sources—are robust to the challenges to which they are most commonly exposed (Kitano, 2004). In evolutionary terms the “threat,” imposed by running, takes many forms. If energy depletes; if mechanical tissue tolerances are exceeded; if neural processes are overloaded to the extent that movement precision and/or cognitive clarity declines, then inevitably, survival probability diminishes (Todorov, 2004; Niven and Laughlin, 2008; Skoyles, 2008; Miller et al., 2012). No single imperative necessarily predominates in any given context. Instead, the neurobiological system seeks to satisfactorily and simultaneously resolve multiple partially overlapping, yet partially competing, organizational constraints (Wolpert et al., 2011).

2.2.1 The Running Robustness Challenge

Singularly within the mammalian kingdom, humans favour a prolonged upright, bouncing, bipedal running gait. Although other primates are capable of running for short distances, they are highly inefficient and, hence, reluctant runners (Bramble and Lieberman, 2004). We, however, run in an inherently unstable bouncing gait; managing impacts of multiple times bodyweight; steering direction of travel; retaining stability; generating sufficient propulsive forces to facilitate vigorous rearrangement of limb positions in preparation for upcoming contact; and accomplish all of these tasks within the abbreviated timeframes afforded by short ground collisions, interspersed between long flight periods.

Further complicating the running challenge are inevitable signal transmission and processing delays, in feedback and feedforward communication loops; delays impeding the rapidity with which the motor system can formulize, and action, responses to arising sensory information (Wolpert et al., 2011). Additionally, the dynamic multi-limb, multi-muscle nature of running produces unavoidable sensory “noise”; discrepancies between intended and actual muscle activations; errors in predicting behaviours of fragile soft-tissues, and mis-estimations of characteristics of the external environment (Skoyles, 2006; Wolpert and Flanagan, 2010; Wolpert et al., 2011); all factors theoretically conspiring to ensure bipedal running is precariously unpredictable, and energetically and computationally expensive. Nevertheless, despite these apparent limitations, human running exhibits remarkable robustness under diversely challenging conditions.

2.2.2 The Highly Evolved Top-Down Coordinated Control of Human Running

From an evolutionary perspective, upright locomotion appears a bizarrely risky survival strategy. We precariously balance our fragile brains over the narrow base of support provided by our disproportionately skinny feet, and in the very position where they are most vulnerable to falling injury. Our ability to safely run, in such an apparently dangerous manner, is facilitated by a comparatively recent evolutionary innovation. An innovation facilitated by our uniquely, in comparison to all other mammals, expanded cerebello-cerebral cortical circuitry (Todorov, 2004; Skoyles, 2008; Wolpert et al., 2011). Specifically, this cortical expansion has dramatically enhanced our ability to construct high-fidelity, temporally-resolved internal models capable of accurately predicting the likely consequences of upcoming interactions between the body and the external environment (Skoyles, 2006, 2008; Wolpert et al., 2011). This predictive capacity enhances anticipation of potential sources of upcoming perturbation and underpins our ability to pre-prepare, and seamlessly integrate, advance-planned, multi-level compensatory postural adjustments—customized to repel potential destabilizations—into on-going movement instructions (Skoyles, 2008).

We take such abilities for granted. Reliably predicting the future consequences of on-going muscle activations, however, is a complex task requiring accurate estimation of current kinetics and kinematics, relative tissue behaviours and mechanical loading tolerances. All of this dynamically emerging information must be integrated with embedded experience to project how the shock load, imposed by ground contact, will interact with fallible bio-composite tissue structures. Developing such high-level predictive accuracy demands the painstaking construction—over the course of our extensively prolonged maturation—of highly detailed, experientially driven, internal models (Skoyles, 2008). Once matured these models permit the skilful blending of sensory-informed estimates of current internal and external conditions, with accurate forecasting of upcoming de-stabilizations, to shape the emergence of the anticipatory postural adjustments necessary to preserve stability in the face of dynamically shifting running conditions (Skoyles, 2006; Wolpert and Flanagan, 2010; Wolpert et al., 2011).

2.2.3 Multi-Sensory Cross-Correlated Mapping

As we repeatedly activate muscles, and receive sensory feedback on subsequent movement consequences, cross-correlated correspondences are gradually engrained and refined; correspondences which ultimately map the relationships between movement intentions, expectations, activations, and outcomes (Skoyles, 2006; Wolpert and Flanagan, 2010; Wolpert et al., 2011). Driven by persistent repetition, these practice acquired relationships capture, in detail, the integrated relationships between activation and sensation. Gradually, this constant triangulation of intention, activation, and sensory information drives a detailed mapping of the multi-dimensional, sensorimotor landscape.

Slowly, with continued practice, discrepancies between projections and outcomes are progressively resolved, and accuracy, sensitivity and efficacy of subsequent motor unit activations become ever more finely calibrated. Ultimately, these elaborately detailed cortically-located, internal models enable us to virtually simulate upcoming interactions, between runner and environment, and to formulise advance-planned remedial solutions (Todorov, 2004; Skoyles, 2006; Wolpert et al., 2011).

Consequently, when we run, the primary motor commands initiating and directing movement are accompanied by activation instructions, in the form of anticipatory feedforward motor adjustments, tailored to counteract forecasted upcoming destabilizations. Hence anticipated perturbations can be skilfully offset by the active orchestration of multiple micro-movement permutations, managed in a centrally determined, precisely timed manner (Skoyles, 2008; Shadmehr et al., 2010; Wolpert et al., 2011).

The accuracy of perturbation predictions is dictated by the refined interpretation of emerging sensory information, facilitated by these richly detailed, practice-acquired models (Shadmehr et al., 2010; Wolpert and Flanagan, 2010). Ultimately, our evolved capacity to construct elaborative internal models underpins the skilled anticipation, and efficient remediation, of looming de-stabilizations; thereby enabling the CNS to sensitively, rather than clumsily, calibrate micro-movement adjustments to best fit emerging running contexts. Subsequently, therefore, offsetting the need for periodic, energy costly, emergency corrections. The calibrated clarity of these mapped relationships dynamically modifies in response to shifting circumstance; chronically, in response to factors such as practice-induced learning and accumulating neural or peripheral “wear-and-tear” and acutely, in response to mounting fatigue, soreness's and sensitivities (Wolpert et al., 2011).

2.2.4 Spinally-Mediated Control

2.2.4.1 Central Pattern Generators: Enhancing Processing Efficiency

Locomotion, in terrestrial and marine life forms, is characterised by automated, cyclical patterns of muscle activation. Critical to the fluent execution of rhythmical gaits are spinally located neural networks, Central Pattern Generators (CPG's) (Thoroughman and Shadmehr, 2000; Lacquaniti et al., 2012). CPG's contain, embedded within their neural architecture, the undulating rhythmical patterns of motor neuron firing necessary to drive cyclical locomotive behaviours (Thoroughman and Shadmehr, 2000; Lacquaniti et al., 2012; Dzeladini et al., 2014). Although experimental difficulties remain a barrier to fuller understanding of CPG's in humans, recent work highlights their importance in evolutionary-prioritized gaits (Lacquaniti et al., 2012; Dzeladini et al., 2014).

The out-sourcing of evolutionary-critical activation templates to spinal CPG's economises information storage and signal transmission efficiency; thereby providing a means through which sparsely detailed low-dimensional inputs can be translated into coordinated patterns of richly-detailed, high-dimensional rhythmic outputs

(Thoroughman and Shadmehr, 2000; Dzeladini et al., 2014). Subsequently, therefore, unburdening higher cortical centers from having to meticulously specify routine rhythmical activation patterns. Accordingly, CPG's dramatically reduce the need for highly elaborative descending motor commands, from supra-spinal to spinal centers; thereby minimizing precious communications bandwidth and providing a mechanism through which higher-cortical centers, rather than micro-managing movement specifics, need only fulfil an overseeing function (Thoroughman and Shadmehr, 2000). Once initiated CPG's are capable of autonomously sustaining locomotive activity, even switching between gaits with minimal descending guidance (Thoroughman and Shadmehr, 2000; Lacquaniti et al., 2012; Dzeladini et al., 2014). However, on-going supra-spinal monitoring, and consistently updated sensory feedback, integrate to adapt motor performance to any changes in running-context (Thoroughman and Shadmehr, 2000; Sidhu et al., 2012; Dzeladini et al., 2014).

Accordingly, when running demands are predictable, higher cortical resources are spared, allowing supra-spinal centers to devote resource to higher-order neural functions, such as cognition and “executive-level” decision-making. If, however, the coordinative challenge escalates—due to, for example, unpredictable surfaces or mounting fatigue—, descending top-down direction intervenes to specifically customize CPG activity to current context (Zehr et al., 2007; Ijspeert, 2008; Sidhu et al., 2012). Accordingly, top-down intervention is more necessary, and strongest, in unpredictable environments which impose non-formulaic running challenges (Suzuki et al., 2004; Slobounov et al., 2006; Jahfari et al., 2012).

2.2.4.2 Running and Reflexes

An unknown number of reflexes proliferate the brainstem and spinal cord. Each of these reflexes drives automated, perturbation-stabilizing responses in the absence of top-down supra-spinal commands (Heng and de Leon, 2007; Wolpaw and Chen, 2009; Dimitriou, 2014). In recent decades, it has become apparent that reflexes are more pervasive; more widely distributed; more adaptive to context; more fluidly integrated with and manipulated by higher-level processes and sensory feedback than historically envisaged (Jahfari et al., 2012). Given their automated action, reflexes offer a supplement to supra-spinal control, providing the CNS with an additional mechanism to speedily action remedial responses to emerging sensory information.

Reflexes are conventionally categorized along a spectrum of response times. Long-loop reflexes are highly modifiable and, as repeat practice adjusts inter-neuronal bias, are customized to favourably regulate gain between afferent inputs, and motor outputs. Through these mechanisms, regularly encountered movement permutations, of timings and positional cues, can be programmed to accentuate or dampen activation thresholds and response magnitudes, depending upon whether reflex activation helps or hinders desired movement outcomes (Heng and de Leon, 2007; Wolpaw and Chen, 2009; Dimitriou, 2014).

The stretch reflex, a reaction provoked when muscle spindles are suddenly stretched, serves as useful illustration. In comparison to non-runners, trained runners have readily triggered stretch reflexes, responding with heightened reflexive counter-actions (Ogawa et al., 2012). In contrast, ballet dancers, who habitually cushion ground reaction forces to finely control postures, substantially suppress stretch reflexes during practiced landing activities (Nielsen et al., 1993). Unlike their more slowly responding longer-loop counterparts, fast-acting monosynaptic reflexes are less readily modifiable by experience (Wolpert and Flanagan, 2010). Nevertheless, their inflexible pre-programmed reactions are predictable, and can therefore, with practice, be productively harnessed to contribute to upcoming movement tasks.

2.2.5 The Inherent Limitations of Top-Down Neural Control

Together these hierarchical neural processing modules, dispersed throughout supraspinal and spinal branches of the CNS, sensitively and responsively blend their collective outputs to direct running actions. There are, however, innate limitations to top-down neural control. Specifically, there are inherent signal transmission delays in cortical communication and spinal reflex loops; there are unavoidable misestimations of tissue positioning's and capacities; there are inevitably unpredictable changes in surface integrity and impact conditions, and there is the ever-present and unavoidable sensory noise implicit when controlling dynamic multi-limb, multi-tissue activity (Blickhan et al., 2006, 2013; Haeufle et al., 2012).

Such factors should, theoretically, greatly detract from the efficiency of bipedal running. Nevertheless, despite these apparent design flaws, the human neuro-mechanical system behaves remarkably robustly and proficiently during running. A proficiency which is facilitated by an incredibly ancient and primitive evolutionary innovation.

2.3 Preflexes: Evolution's Movement-Management Shortcuts

Intriguingly, when evolutionary-relevant impact activities—running, jumping—are closely scrutinized, it is evident that compensatory stabilizing reactions occur in advance of the fastest acting mono-synaptic reflexes (Brown and Loeb, 2000). Similarly, when surprised by suddenly changing surface compliance, leg stiffness adjusts and compensates in advance of altered EMG-signal; thereby suggesting that initial leg stiffness adjustments occur in the absence of top-down neural activation (Moritz and Farley, 2004; Daley et al., 2007; van der Krogt et al., 2009).

These mysterious instantaneous responses, as they occur “pre-reflexively,” have been termed “preflexes” and, as they operate without neural direction, are neither the same as, nor a sub-set of, reflexes (Brown and Loeb, 2000; Dickinson, 2000; Moritz and Farley, 2004). The preflex phenomenon is such an elegantly simple evolutionary innovation that its contribution to running remains, conventionally, overlooked.

2.3.1 Solving the Preflex Puzzle: The Biotensegrity Solution

Over the span of evolutionary deep-time we have evolved from single cell entities, to dexterously skilful masters of our physical universe. At every step of this journey, Nature's blind tinkering has been pressurized by Darwinian imperatives to save energy, reduce uptake of precious neural resources, and discomfort and damage. Every dimension of our structural, morphological design has been shaped by these ever-present imperatives; resulting in the creation of highly innovative and integrated solutions to evolutionary survival problems (for review, see Turvey and Fonseca, 2014).

2.3.1.1 Biotensegrity essentials

The individual components of a tent become structurally stable only when the covering sheet is draped over a lattice of stiff poles and appropriately tensioned by strategically placed lines and pegs. Biological organisms are, needless to remark, vastly more complex. Nevertheless, when scaled to the level of biological complexity, this generalized theme—whereby tensile and compressive components are collectively pre-stressed in a specific configuration enabling disproportionate self-stabilizing resilience subsequent to deformation—is a ubiquitous evolutionary innovation (Fuller, 1961; Ingber, 2008; Turvey and Fonseca, 2014). In engineering contexts, such configurations have been termed tensegrity systems (Fuller, 1961; Ingber, 2008).

Experimental work, over the past three decades, reveals that molecules, cells, peripheral tissues, organs, and our entire bodies use such self-equilibrating design principles to repel suddenly imposed deformation (Levin, 2002; Schleip and Müller, 2013; Turvey and Fonseca, 2014).

Thus, for example, at the level of the cell, actin microfilaments stiffen cell structures and serve as conduits for mechanical stress; actomyosin microfilaments transmit forces continuously throughout the whole cell; and intermediate microfilaments function as tensioned guide-wires stabilizing the cell. On the macro-scale, skeletal structures can absorb large compressive forces; muscle tissue generates contractile forces, and the fascial web of connective tissue conveys tensile forces (Turvey and Fonseca, 2014). Subsequently, the innate deformation-resistance of any biotensegrity system, at the instant of load application, arises simply from the relative configuration of tensioning and stiffening elements, and the structural integrity provided by a pervading binding pre-stress. This binding pre-stress provides a background “tone”, pervading the structure, which serves to strategically compress stiff rod-like, and tension taut cable-like, elements of the system in a state of dynamic equilibrium.

This background pre-stress is not a product of neural activity, and is hence invisible to EMG (Turvey and Fonseca, 2014). Muscle, for example, has an electrically invisible intrinsic “tone,” ensuring tissue is never completely lax. Similarly, the collagen lattice, of the all-encasing fascial network, imparts a tensioned structural integrity which binds and stiffens bundles of tissues through a unifying pre-stressed tautness (Schleip and Müller, 2013; Turvey and Fonseca, 2014). Through this pre-stressed medium each tissue cell is bound to the next. A mechanical deformation in one tissue component, is instantaneously transmitted to its neighbour. Subsequently, mechanical deformations are transmitted and absorbed by the all-enveloping, pre-tensioned connective web, connecting the mechanical state of each cell to that of the whole body (Levin, 2002; Schleip and Müller, 2013; Turvey and Fonseca, 2014). Throughout the musculoskeletal system, tissues of this bio-composite connective net variously press and pull, stiffen and strain, against other tissues. Crucially, this arrangement is not haphazard but meticulously evolutionarily configured to resist, accommodate and productively harness the mechanical stresses and strains most pertinent to our species survival.

From micro- to macro-scales, our biological structures represent long series of nested biotensegrity systems capable of, individually and collectively, eliciting disproportionate and non-linear restorative responses to imposed biomechanical disruptions (Levin, 2002; Schleip and Müller, 2013; Turvey and Fonseca, 2014).

2.3.2 The Running Biotensegrity System

During running, impact forces swirl in a multi-directional vortex; subjecting tissues to various degrees of compression, stretch and twist, as the shudder of impact reverberates through the system. To move safely, these forces must be dispersed to alleviate the risks of exceeding critical tissue loading limits; to move efficiently, these forces must be channelled and re-deployed to optimally contribute to stabilization and propulsive power requirements.

By manipulating the kinematics and kinetics of the running leg, the skilled runner can exploit their innate material and geometrical properties to create conditions whereby the sudden shock of ground contact is absorbed and channelled for minimal effort, in terms of top-down direction and energetic demand, and for minimal risk of destabilisation and tissue damage.

Driven by evolutionary imperatives and repeat practice, we progressively become more skilled at exploiting these built-in mechanical efficiencies. We gradually become more proficient at leveraging biotensegrity structures to more productively capitalise on “cheap” sources of control and propulsion, merely by harnessing the innate deformation-repelling structural and material properties of the body, and appropriately matching them to the physics of the running context (Brown and Loeb, 2000; Daley and Biewener, 2006; Biewener and Daley, 2007). Importantly, when appropriately harnessed, the judicious deployment of our structural and morphological properties remedies the inherent information-processing delays implicit in top-down control (Biewener and Daley, 2007). In essence, thereby providing an instantaneous non-neurological, yet skilled, response to sudden perturbation; automatically buffering, stabilizing, re-directing, and re-cycling momentums for little energetic or neurological investment (See Figure 2.1).

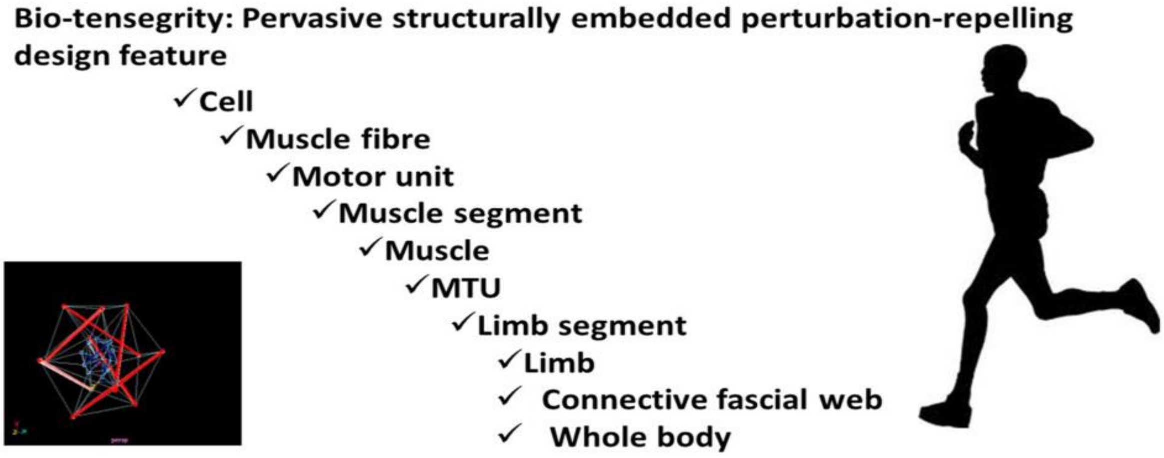

Figure 2.1: **Biotensegrity: Our innate perturbation-repelling design feature**

2.4 Running Coordination: Integration of the Ancient and the New

2.4.1 Practice-Driven Plasticity

This blending of archaic and comparatively recent evolutionary innovations is enabled by a pervasive characteristic of the human condition: activity-dependent plasticity. Activity-dependent plasticity represents the capacity, both within the CNS and tissues of the periphery, to adapt—structurally, chemically, electrically, materially, and ultimately functionally—to repeated experience (Knikou, 2010; Taubert et al., 2010).

Throughout supra-spinal and spinal branches of the CNS persistent patterns of neural activations induce plastic re-configurations; modifications serving to micro-architecturally concretise relationships between regularly co-operating neural components, and between neuronal apparatus and activated motor units (Dickinson, 2006; Lemon, 2008; Taubert et al., 2010). Plasticity in the CNS is mirrored in the periphery as tissues modify in response to habitual loading patterns. Muscle, in particular, is highly plastically evolvable, as habitual loadings progressively sculpt the non-linear, visco-elastic, length-velocity-force relationships of muscular sub-compartments; thereby tailoring material and architectural characteristics to best fit regularly encountered movement contexts (Flück, 2006; Harridge, 2007; Hoppeler et al., 2011).

2.5 Conclusion: Coordinated Blending of Top-Down and Bottom-Up Control Processes

The coordinated control of human running is enabled by the finely tuned, tightly integrated blending of primitive evolutionary legacies, conserved from reptilian and vertebrate lineages, and comparatively modern, more exclusively human, innovations (Lemon, 2008). The operations of neuronal top-down, and mechanical bottom-up, control processes are so seamlessly integrated that describing their functionality in isolation is to obscure the true nature of coordinated running. There are no discontinuities where one ends and the other begins, instead organizational levels are irrevocably functionally entangled (Biewener and Daley, 2007).

When we run, top-down feedforward control responds to emerging multi-modal sensory information to strategically orientate tissues to exploit our nested biotensegrity design (See Figure 2.1). On ground contact, immediate perturbation buffering is provided by “dumb,” but skilfully manipulated, preflexive morphological responses. These responses dampen disturbances through tactical deployment of passive tissue properties; thereby providing simple, but effective, attenuation of imposed decelerations. As stance progresses, shorter-loop, then longer-loop, reflexes are layered over initial preflexive responses; thereby further supplementing and customising initial control demands. Repetitive, cyclical activation patterns are delegated to spinally-located CPG's; thereby reducing the control burden imposed on evolutionary precious and energetically costly supra-spinal centers (Todorov, 2004). The spinal cord thus serves, not as a rigidly hardwired communications conduit, but as a plastically modifying extension of higher neural centers, capable of independently integrating CPG and reflex interactions with descending commands, and ascending sensory information (MacKay-Lyons, 2002). In the event of non-formulaic coordinative challenge, higher-order neural resources intervene, exerting top-down executive direction to specifically tailor activation signals to offset emerging instabilities and threats.

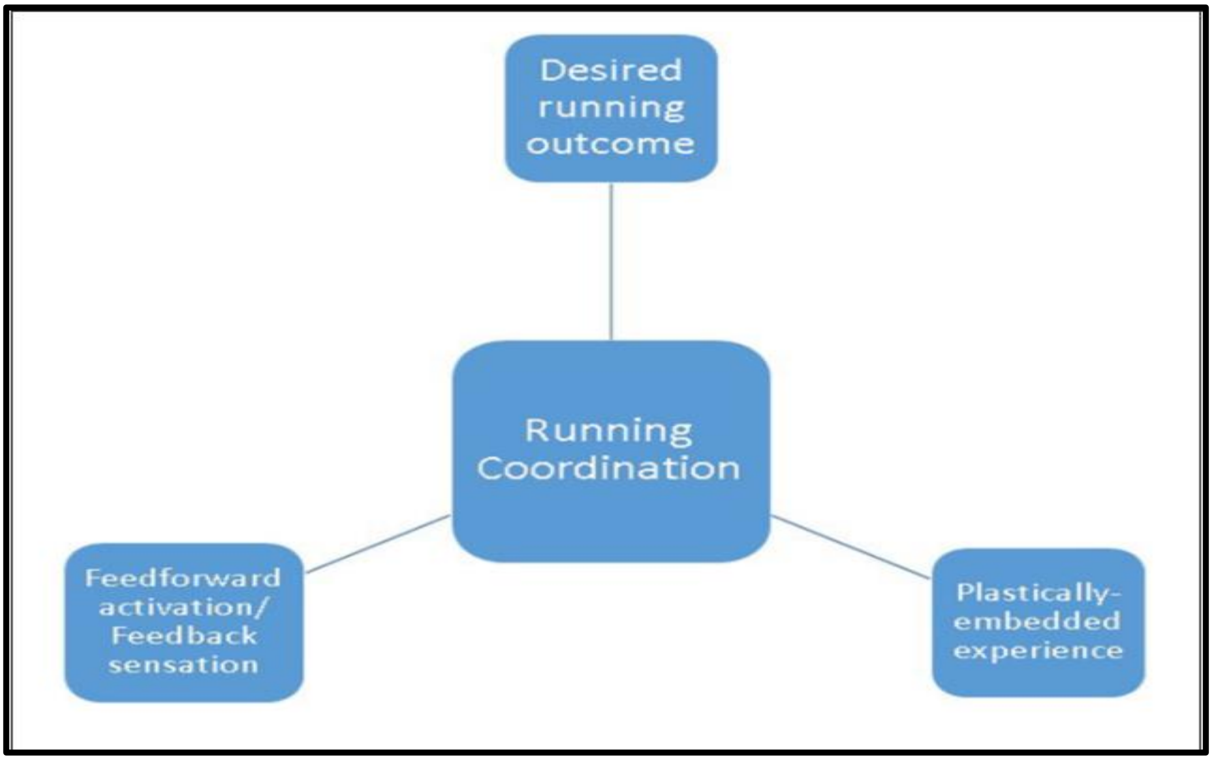

Figure 2.2: Running coordination as the blend of plastically-embedded experience; sensorimotor integration of feedforward activation and feedback information; desired running outcome

Over countless gait cycles, evolutionary-bestowed protective mechanisms persistently seek to extract more benefit, for less cost. Thus, as we practice, we progressively learn to more astutely poise bio-composite tissue structures in response to more sensitively interpreted sensory information (Haeufle et al., 2012). This deeply integrated blending of top-down and bottom-up strategies provides a robust system of collective, collaborative, distributed control; a system permeated with built-in overlapping degeneracies and compensatory fail-safes; a system enabling deficits, errors or failures emanating from any specific control module, to be remediated by changing contributions from others (Whitacre, 2010; Mason, 2015). Ultimately, this system of distributed, multi-level control enables the human runner to negotiate varied challenges and terrains, for a reduced neuronal investment; a reduced energetic cost, and a reduced exposure to survival threatening trauma.

Without question the various coordinative undercurrents, summarised here, encompass many complex and unresolved academic puzzles. Although such a conceptual framework may, at first glance, seem far removed from real-World running performance and injury considerations, it does provide an insightful theoretical lens through which to conceptualize the underlying nature of human running coordination.

Other species similarly rely on coordinated processes to run, yet the unique demands imposed by the bouncing bipedal nature of human running presents a specific set of coordination challenges, solved using a novel configuration of evolved solutions.

2.6 In summary

This exploration provides a basis for conceptualising the evolutionary engrained processes serving to embed and constrain human running coordination.

Importantly, from the perspective offered here, running coordination is clearly a multi-module collaboration serving to seamlessly integrate elements of both neural and structural control. Coordination, accordingly, is vulnerable to the progressive erosion of both sensorimotor processing network and biological, load-bearing tissue collectives. More acutely, running coordination is vulnerable to perceptions of pain or discomfort, gradually increasing sensitisation, and the detrimental consequences of fatigue manifest in the multiple neural networks responsible for regulating motor control.

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Chapter 3:

The Robust Running Ape

(This chapter was published, in large part, as “Kiely, J. (2017). The robust running ape: Unravelling the deep underpinnings of coordinated human running proficiency. *Frontiers in Psychology*, *8*, 892.”)

3.1 Introduction

Running is the most primitively ancient of athletic movements is critical to competitive success in many sports and, in evolutionary contexts, critical to survival. Uniquely amongst mammals, humans employ an upright bipedal, bouncing gait when running. A gait characterized by long flight times, interspersed with brief ground contacts during which the shock of impact –equating to multiple times bodyweight–, is absorbed, re-cycled and steered through the narrow stabilizing platform provided by a single supporting foot. Nevertheless, despite these apparent limitations, we are amongst Nature’s most supremely well adapted runners (Bramble and Liberman, 2004).

The evolutionary innovations bestowing human running proficiency do not, however, render us invulnerable to breakdown and running-related injuries are common (Van der Worp et al, 2015). Runners seem particularly exposed to Overuse injuries, with up to 70% suffering such injury each year (Van der Worp et al, 2015; Clarsen et al, 2013; Saragiotto et al, 2014). Various definitions exist, amid some inconsistency, and confusingly 'Overuse' describes both a 'mechanism' and 'type' of injury (Clarsen et al, 2013). Although definitions vary, published consensus agrees that Overuse syndromes arise consequent to progressively mounting micro-trauma accumulated over a protracted period, exacerbated by insufficient recovery leading to increasing tissue sensitization in the absence of single catastrophic events (Clarsen et al, 2013; Saragiotto et al, 2014). Commonly cited risk factors include: elevated running volumes, prior injury, fatigue and background psychosocial stress (Van der Worp et al, 2015; Clarsen et al, 2013; Ivarsson et al, 2016); yet how these factors synergistically interact, leading to Overuse injuries, has yet to be clarified (Van der Worp et al, 2015).

A frequently overlooked distinction between running and many other sporting movements is that running is one of a limited sub-set of gaits —along with crawling and walking— that are so evolutionary ancient as to have mutually co-evolved in tandem with human neural and biological infrastructures (Kiely and Collins, 2016). In short: how we run is shaped by, yet has also contributed to shaping, modern human morphology, in ways that other sporting movements —a golf swing; a tennis serve; rowing; the butterfly stroke—, have not. An implication of this synergistic co-evolution of form and function is that the adaptations underpinning human running permeate every dimension of our anatomical, biological and neurological being. Our capacity to withstand the extraordinary mechanical and stability challenges imposed during our bouncing bipedal running gait, consequently, is not attributable to any single evolutionary adaptation. Instead, human running robustness emerges subsequent to our slowly developing capacity to seamlessly harness, orchestrate and integrate the outputs of multiple biological and neurological sub-systems, to effectively accomplish running objectives. In short: our ability to coordinate the running action.

The core defining feature of coordination is that multiple components work together to realize an objective (Diedrichsen et al, 2010). Conventionally, within the sports sciences, coordination is perceived through the lens of Dynamical Systems Theory (DST). Elsewhere, however, through the lens of Optimal Feedback Control Theory (OFCT), conventional interpretations of DST have been criticized for obscuring the fundamental priority of sensory feedback in shaping effective movement coordination (Todorov, 2004; Todorov, 2009). The OFCT framework subsequently claims to more prominently highlight the relationship between high-level goals, and the real-time sensorimotor control strategies most suitable for accomplishing those goals. Recent ecological dynamics perspectives have similarly advocated the prominent role of emerging sensory ‘information’ in regulating on-going motor behaviour (Seifert, Button and Davids, 2013). As in other scientific domains, however, debates and disagreements proliferate, and the need for on-going argument, skepticism and scrutiny remain obvious. Various perspectives, accordingly, have been expertly and extensively discussed within their respective motor control and neuroscientific literatures (see for example: Davids and Glazier, 2010; Nagengast et al, 2010; Proske and Gandevia, 2012; Kelso, 2012). The problem, for the vast majority of practical sports scientists, sports medicine practitioners and evidence-led coaches, is that while these academic debates are essential, by necessity they are abstract, highly technical, typically obscured by the in-house terminology of the specific academic realm, and often too narrowly focused to provide practically implementable insight. Accordingly, the overarching objective of this chapter is to provide an updated, evidence-led synopsis of the linked dimensions of the running coordination phenomenon deemed most relevant to performance, resilience and injury rehabilitation.

3.2 The evolutionary undercurrents of Coordinated Running robustness

Evolutionary survival demands that biological systems, operating in unpredictable environments using unreliable components and finite energy sources, are robust to the challenges to which they are most commonly exposed (Kitano, 2004). Accordingly, from an evolutionary perspective, running coordination's overriding imperative is to deploy available resources to satisfactorily achieve desired outcomes, while preserving an acceptable robustness to any running-imposed 'threat' serving to reduce survival probability.

This ‘threat’ takes many forms. If energy depletes; if mechanical tissue tolerances are exceeded; if neural processes are overloaded to the extent that movement precision and/or cognitive clarity declines, then inevitably survival probability diminishes. No single survival imperative necessarily predominates. Instead, the neurobiological system seeks to satisfactorily resolve multiple partially overlapping, partially competing organizational constraints (Miller et al, 2012; Skoyles, 2008; Hodges and Tucker, 2011). In negotiating this complex organizational problem, evolution has arrived at a typically ingenious resource-sparing set of solutions.

3.2.1 Interpretation of sensation shapes movement

As running increases in severity, we are made consciously aware of mounting ‘threat’ through increasingly discomforting interpretations of arising sensory information (Marcora et al, 2009; Smirmaul, 2012). At the whole-body level growing discomfort influences psycho-emotional state, amplifying perceptions of anxiety, ‘pain’ and diminished attention, which in turn intensify the inner conflict between motivational drive and perceived effort –what we, collectively, interpret as mounting ‘fatigue’ (Marcora et al, 2009; Smirmaul, 2012; Seay et al, 2011). At the local level, muscle activation patterns are subtly modulated to offload sensitized tissues; thereby moderating regionalized discomfort and alleviating irritation (Seay et al, 2011; Gerlach et al, 2005; Mizrahi et al, 2000). Through these mechanisms our interpretation of arising psychobiological discomfort informs us of increasing risk —of impending tissue damage, elevating metabolic costs, increasing neural processing demands and cognitive effort—; thereby providing a direct means through which the perceived relevance of changing sensation directly influences running behavior (Marcora et al, 2009; Wolpert et al, 2011).

Prompted by subtle, but persistent, sensory signals, the CNS continually searches for economic trade-offs between desired outcomes, available resources and discomforting perceptions of 'threat' (Miller et al, 2012; Hodges and Tucker, 2011). As we accumulate running experiences, we learn to more precisely triangulate between sensory feedback, feedforward activation and desired running outcomes, and gradually gravitate towards coordinative solutions more satisfactorily resolving these multiple competing constraints.

Accordingly, progressively, with practice, sensory information and muscular activation strategies co-evolve into a seamlessly integrated sensorimotor system; whereby changes in sensation directly modulate muscular activations, and changes in activation directly modify sensation (Wolpert et al, 2011). Through this elegantly efficient process, sensory feedback information and feedforward activation instructions become irrevocably mutually entangled; subsequently preserving running robustness, within acceptable limits, through an integrated sensorimotor process of 'self-organizing optimality' (Glazier and Davids, 2009).

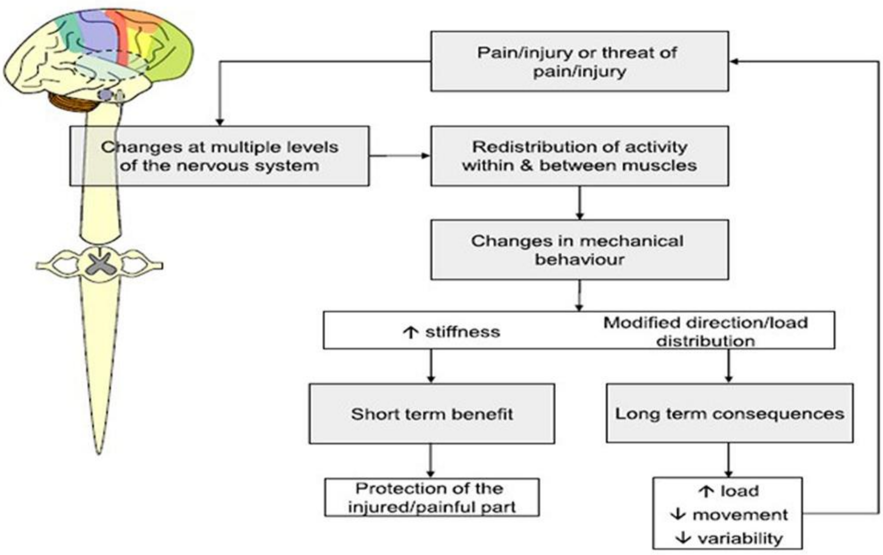

Figure 3.1: Mechanisms through which pain leads to re-distribution of activity within, and between, muscles

3.2.2 Organizing Neuro-biological Complexity; Modularity facilitates degeneracy

Biological lifeforms are reflectively characterized as complex adaptive systems. Complex, as the behaviours of individual components are inextricably linked to those of multiple others through arrays of processes, cycles and regulatory feedback loops. Adaptive, as the behaviours and collaborative outputs of collections of components flexibly modify their concerted contributions to best fit current context (Manor and Lipsitz, 2013). In 1992 Lewis Lipstiz and Ary Goldberger published a landmark paper, in *JAMA*, suggesting that our capacity to robustly and agilely respond to the broad spectrum of multi-dimensional challenges to which we are continually exposed, is founded on a platform of extensive neuro-biological complexity. Subsequently, as complexity gradually diminishes, in response to the slings, arrows and insults of a normal life, our capacity to robustly respond to repel multi-source challenge, inevitably contracts. Accordingly, this *loss of complexity hypothesis* proposes that age-related functional decline is driven by a progressive loss of complexity across multiple, integrated neuro-biological processes (Lipstiz and Goldberger, 1992). More recently, however, a review updating the *loss of complexity hypothesis*, by Busa and Van Emmerik (2016), suggested that although diminishing complexity is commonly associated with aging, the root cause is more accurately perceived as increased sensorimotor impairment. As old age increases the likelihood of excessively accreting micro-trauma and subsequent control deficits, aging may be associated with accumulating damage, but is not the critical driver. Instead, Busa and van Emmerik coherently argue that it is the gradual accumulation of damage, deficits and deteriorations which inevitably erodes the micro-machinery underpinning every dimension of neurobiological function (Busa and Van Emmerik, 2016).

Each individual entity within the complex organism is linked, physically or functionally, to every other. Nevertheless, there remains an evident modularity; whereby collections of elements are more densely networked to each other, than to elements within other modules (Whitacre, 2010; Mason, 2015). All modules are subsequently inter-connected yet are simultaneously partially-insulated and functionally semi-autonomous. Modularity, accordingly, facilitates robustness as modules can evolve, reshape, rewire and repair in tandem, or independently, without jeopardizing the survivability of the entire organism (Mason, 2015; Maleszka et al, 2014).

Modularity, accordingly, is a fundamental neuro-biological organizing principle, greatly simplifying otherwise overwhelmingly disordered complexity. Related modules exhibit extensive *functional overlap*, such that alliances of neural networks and peripheral tissues can spontaneously modify behaviours to achieve equivalent ‘outputs’ through a multiplicity of pathways. This functional agility is often conflated with redundancy but is perhaps more reflectively termed *degeneracy* (Glazier and Davids, 2009; Mason, 2015; Seifert et al, 2016). Degeneracy describes the ability of multiple, alternate structural pathways to achieve either similar or dissimilar functional outcomes, dependent on current context (Seifert et al, 2016). Degenerate systems are subsequently capable of performing similar or overlapping operations or fulfilling similar or overlapping functions. Subsequently, degenerate systems are fundamental facilitators of biological complexity, robustness and evolvability (Whitacre, 2010; Mason, 2015, Maleszka et al, 2014). Redundancies, in contrast, occur when sub-sets of identical elements combine to achieve similar outcomes. True redundancy is subsequently rare, as there are few identical neural and/or biological entities. Degeneracy, however, describes a more flexibly adaptive phenomenon; whereby collaborating communities of fundamentally distinct components produce reliably consistent outputs under fluctuating conditions (Mason, 2015; Seifert et al, 2016).

The human runner represents a highly degenerate system. Consider the phenomenon of leg stiffness during ground-contact —the accurate calibration of which promotes the protective dampening, and economic re-cycling, of impact shocks. Our highly-degenerate neuro-biological design ensures that equivalent leg stiffness values can be realised using a potentially limitless expanse of organisational permutations. As illustration, muscle-tendon units (MTU's) may vary their individual movement power contributions whilst, collectively, whole-leg power outputs remain constant; individual MTU's may realise equivalent force outputs, by summating different muscle and tendon contributions; individual muscles may activate different motor unit populations, under differing contractile conditions, to produce identical tensions; and varying and re-combining diverse combinations of torso, leg and foot postural orientations, with different background levels of pre-set tensions, can deliver equivalent propulsive outcomes (Roberts and Azizi, 2011; Wickham and Brown, 1998; Turvey and Fonseca, 2014). Accordingly, this option-rich, highly degenerate movement landscape provides a multiplicity of avenues through which collaborating modular alliances combine, and re-combine, to flexibly satisfy dynamically shifting demands.

This degenerate design offers multiple means to accomplish running objectives. Historically, the apparently overwhelming complexity presented by this proliferation of movement 'options' was famously interpreted as a control 'problem' (Bernstein, 1967). This potentially complex problem, however, is reduced by the gradual construction of synergies —coordinative structures comprised of highly context-specific, context-sensitive functional linkages serving to temporarily constrain collaborating elements, such that they act as single coherent units (Latash, Scholz and Schöner, 2007; Wu and Latash, 2014). Through the formation of synergies, the control 'problem' is greatly simplified; while simultaneously retaining the benefits of complexity and degeneracy. As such, the 'problem' of excessive choice has subsequently, and more recently, been reframed as the 'bliss' of motor abundance (Latash, 2012). When running, this abundance of potentially over-whelming movement 'choice' can be, through effective coordination, productively managed to disperse the running work-burden among networks of collaborating tissues; thereby promoting efficiency and robustness. This rationalisation also corresponds to another prevalent means of conceptualising movement control, the hypothesis of the uncontrolled manifold (for example: Scholz and Schöner, 1999). This hypothesis suggests that, as 'control' comes at an inevitable cost (energetically and neurologically), only parameters that need to be constrained within tightly acceptable limits will be closely monitored and subsequently corrected, in the event of deviations outside of acceptable limits. Conversely, movement elements that are not currently critical to current movement outcomes are not tightly controlled and can subsequently deviate within a more expansive range. Non-critical movement parameters are, consequently, given freer rein to deviate, than outcomes that are critical to current movement outcome success (for example: Maldonado, Bailly, Souères and Watier, 2018).

3.2.3 Fractal variation: Deploying coordinative abundance

Conventionally, we equate skilful running with metronomic regularity. As proficient runners achieve reliably consistent stride outcomes, it seems sensible to assume experts precisely replicate running stride characteristics. In recent years, however, close scrutiny of running behaviours illustrates that, even when experts run at steady paces, movement parameters persistently vary (Stergiou and Decker, 2011). Through the lens of traditional motor control paradigms such variability was initially interpreted as 'noise' —meaningless error arising from the intricacies of the engineering challenge, measurement inaccuracies and fallible biological components. Intriguingly, however, more recent investigations reveal the structure of gait variability to be neither randomly erratic, nor independent of prior events. Instead, the architecture of past, current and future stride variabilities appear statistically linked through some, as yet incompletely understood, long-range correlations (Stergiou and Decker, 2011; Haursdorff, 2007; Hamill et al, 2012).

3.2.4 Structured non-random variability

Mandelbrot's classic work, *The Fractal Geometry of Nature* (1982), first popularized the term 'fractal' to describe the phenomenon, pervasive in Nature, of recurrent structural self-similarity (Mandelbrot, 1982). The unifying characteristic of fractals is scale-free structural replication; whereby individual entities are composed of sub-units of a shared structure, while themselves forming super-ordinate entities conforming to a similarly patterned design. Examples include the branching networks of the vascular system and the convoluted folding surfaces of the neo-cortex; both fractally replicating architectures exponentially increasing tissue surface area.

Fractal self-similarity is not, however, confined to physical architectures and also manifests as time-series and/or organizational replications. Thus, sub-regions may be exact or distorted copies of the all-encompassing over-arching structure, or may simply share quantitative, qualitative or statistical properties (Newell et al, 2005; West, 2010; Goldberger et al, 2002; Vazquez et al, 2016). Fractal signatures are ubiquitous in neurophysiology, with multiple phenomena exhibiting self-similarity across observational scales. Famously, the time series of inter-heartbeat intervals — heart-rate variability— is a fractal phenomenon. Although each beat is unique, its uniqueness is not random, but shaped by an innate, neurally-embedded background algorithm. This algorithm effectively blends the organism's unique idiosyncrasies with past experiences, current status and transient momentary demands, to collectively shape the time-series architecture of the emergent heartbeat (Goldberger et al, 2002). Accordingly, the beat-to-beat 'solution' to the circulation 'problem' is neither tightly prescribed, nor loosely erratic.

Expert running coordination is similarly characterized by the tuned inter-play between predictability and responsiveness bestowed by the fractally-fluctuating deployment of option-rich, functionally overlapping degenerate networks. Together, these networks provide the diverse repertoire of behavioural responses essential for survival in chaotic, unpredictable environments (van Orden, 2007; Stergiou and Decker, 2011; Nakayama et al, 2010; Vasquez et al, 2016).

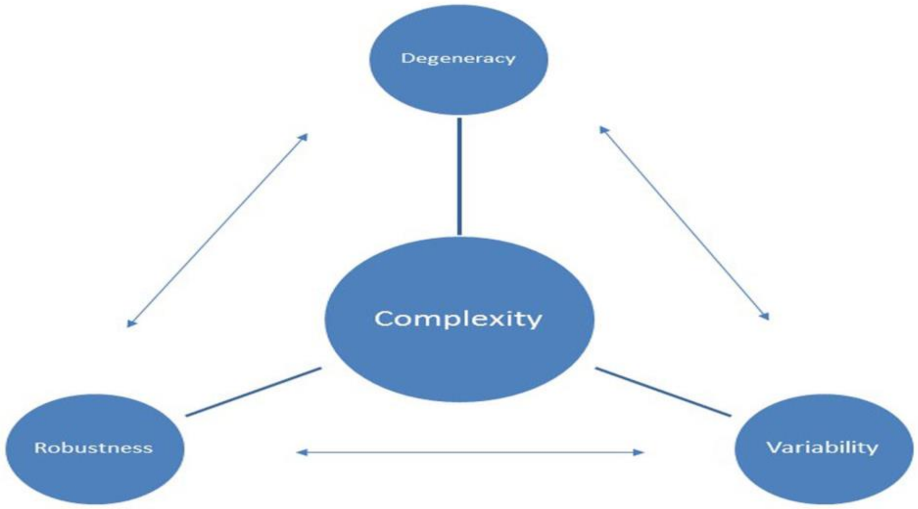

Figure 3.2: Inter-relationships between Complexity and Injury Resilience

3.3 Running variability: Sharing the running work-burden

As with other neuro-biological processes, running dynamics exhibit robust fractal characteristics; thereby suggesting stride-to-stride variability is neither random, nor dictated by the fluctuating idiosyncrasies of current conditions. Instead, on-going stride variations are meaningfully related —in a decaying Power law fashion— to past variations stretching back over thousands of strides (Hamill et al, 2012; Meardon et al, 2011). This pervasive fractal variation ensures that the mechanical stress of running is distributed in ever varying, yet non-randomly organized, patterns; patterns tuned, through practice, to the runner’s unique architectural and experiential peculiarities.

This structured variability enables the well-trained runner to disperse the running ‘work burden’ amongst expanded networks of biological tissues, whilst simultaneously retaining the agility to spontaneously respond to emerging challenge**.** Healthy running, accordingly, is characterized by an optimal bandwidth of movement variability: neither too much, nor too little (Hamill et al, 2012; Meardon et al, 2011). Accordingly, proficient running coordination is not the capacity to monotonously replicate an idealized stride pattern, but the ability to continuously recombine expansive, yet conditioned, populations of collaborating neural and biological components; thereby enabling the achievement of reliably consistent running outcomes through a diversity of subtly shifting movement permutations.

3.3.1 Diminishing complexity, drives dysfunctional variability

As we move through a running life, accumulative cycles of *‘wear and tear’* — of injury, overuse, misuse and disuse— gradually degrade both the material integrity of biological components and the networked richness of neural connectivity (Taubert et al, 2010; Hoppeler et al, 2011; Elbert and Rockstroh, 2004). As neuro-biological complexity contracts, the landscape of viable degenerate permutations, capable of satisfying running demands, deteriorates. Now, the mechanical stress of running must be distributed amongst shrinking networks of collaborating components (Pelletier et al, 2015).

Reductions in viable degeneracies do not, however, inevitably decrease running variability. Instead, as the neuro-biological systems struggles to proficiently manage imposed loadings, mechanical stress becomes either more tightly focused on restricted populations of working tissues or is erratically dispersed amongst expanded webs of unconditioned tissues (van Orden, 2007; Hamill et al, 2012). As illustration, ACL-deficient knees typically exhibit reduced, whereas ACL-reconstructed knees exhibit dramatically expanded, inter-stride variability (Stergiou and Decker, 2011; Hamill et al, 2012). Such deviations from habituated variability ranges, oscillating between overly formulaic constancy and wild randomness, signify an impaired capacity to absorb, disperse and purposefully recycle and re-direct impact momentums (Nakayama et al, 2010). As coordinative fluency deteriorates, vulnerability to Overuse syndromes and unexpected perturbations, subsequently, escalates.

3.3.2 Global accommodation of local perturbation

The entangled nature of complex neurobiology ensures that when variability changes at any discrete location, accommodating compensatory behaviours are introduced elsewhere in the system. As illustration, active injuries typically reduce habitual running variability in the injured leg —constraining control to protect sensitized tissues—, while simultaneously inducing expansions of variability in the non-injured leg (Hamill et al, 2012). Such evidence illustrates that, although running injury is a site-specific event, the accommodation of injury is a system-wide phenomenon occasioning system-wide coordinative adjustment. Importantly, these behavioural modifications, although temporarily functional, inevitably expose compensating tissues to unhabituated loadings. What has not been discussed, within the running-related literature, however, is how such remedial compensations gradually become more progressively micro-architecturally embedded within neural and biological structures.

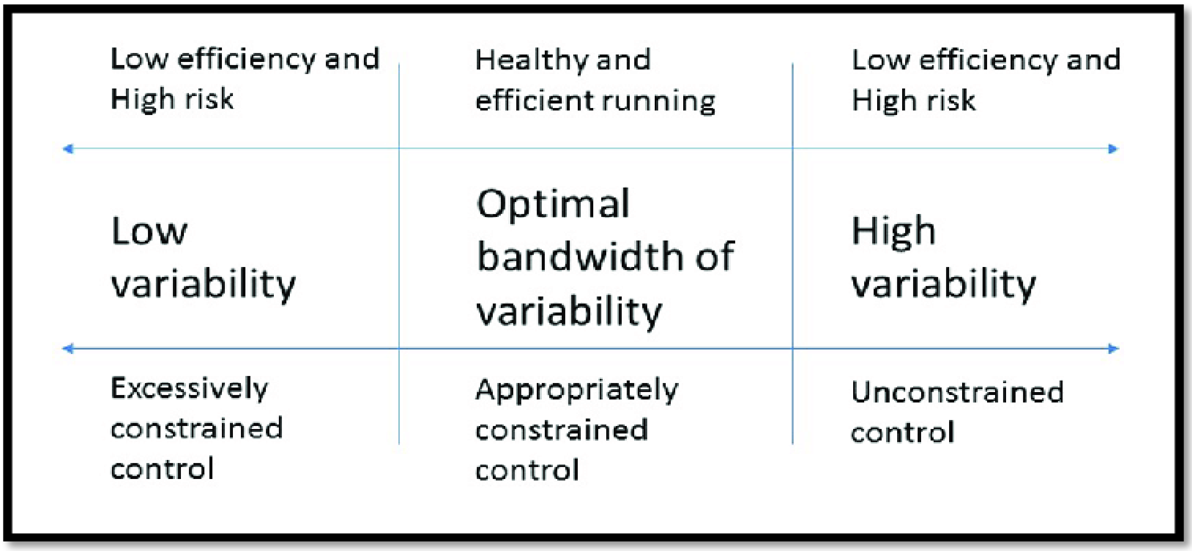

Figure 3.3: Relationships between running gait variability, risk and efficiency

3.4 Pervasive bio-plasticity: The embedded legacy of prior events

A fundamental dimension of human neurobiology is life-long *experience-dependent plasticity*; the capacity within the CNS and tissues of the periphery to lastingly respond —structurally, chemically, electrically and materially— to repeat experience (Taubert et al, 2010; Elbert and Rockstroh, 2004). Throughout supra-spinal and spinal branches of the CNS persistent patterns of neural activation induce experience-dependent plastic re-configurations that micro-architecturally embed the relationships between regularly co-operating neural components, and associated motor units. Plasticity in the CNS is mirrored in the periphery, as tissues re-model in response to habitual loading patterns (Hoppeler et al, 2011). Experience-dependent plasticity, subsequently, constructively refines and economizes communications linkages between collaborating neural networks, and conditions peripheral tissue structures to better cope with regularly encountered movement contexts (Pelletier et al, 2015).

As we converge on our individually unique running styles, pervasive neuro-biological plasticity embeds movement habits; thereby constraining the landscape of degenerate movement options to manageable proportions and increasing the probability previously successful ‘solutions’ will be recycled in the future. Plasticity, accordingly, drives the physical embodiment of coordinative change; thereby sculpting the micro-architectural basis of coordinative synergies, linkages and attractors.

Inevitably, however, plasticity is both blessing and curse, and the engraining of new habits inevitably degrades old habits.

3.4.1 The plasticity of over-specialization

As running experience accumulates, the sensorimotor apparatus becomes ever more efficient at executing the running task. Neural resources, however, are evolutionarily precious and fundamentally limited commodities and, as such, are persistently re-deployed to fulfil varying roles within diverse tasks. Such conflicting usage patterns drives *competitive plasticity* as neural networks strive to persistently re-model neurological *'form'* to best fit currently prioritized *'function'*.

Consequently, as we progress from novice to 'skilful', a by-product of on-going neuro-plastic refinement is that fewer networked collaborators are required to manage the evermore highly practiced running pattern (Pelletier et al, 2015; Coq and Barbe, 2011; Avanzino et al, 2014). Subsequently, it becomes evolutionarily wasteful to continually dedicate expanded sensorimotor networks to task execution. Accordingly, when the range of running behaviours to which we are regularly exposed becomes monotonously stereotypical, evolutionary pressure to economize resource uptake ensures that the landscape of conditioned neural and biological collaborators, dedicated to executing highly-practiced running patterns, progressively diminishes as under-utilized resources are re-allocated elsewhere (Elbert and Rockstroh, 2004;

Avanzino et al, 2014). A drawback, subsequently, of engaging in only a narrow band of overly stereotypical running tasks, is that we become hyper-efficient at deploying reduced populations of degeneracies to execute a narrowing band of self-similar running patterns. As direct consequence, we become increasingly vulnerable to both overuse syndromes, and unhabituated running challenges.

3.4.2 The plasticity of disuse

Prolonged abstinence from running drives a progressive loss of physiological conditioning, whilst also serving to dim the regular flow of running-related sensorimotor information. The critical cortical circuitry, normally maintained by consistently processing running-related sensorimotor information, is subsequently eroded as voracious *competitive plasticity* re-models neuronal inter-connectivity to best fit current usage patterns (Elbert and Rockstroh, 2004; Pelletier et al, 2015). Subsequently, when we return to regular running, coordinative control is slightly less proficient and slightly less resilient.

3.4.3 The plasticity of misuse

When we run in injured or irritated states, we subtly alter coordination patterns to divert discomforting mechanical stress away from sensitized tissues; thereby alleviating negative sensation, tempering structural damage and facilitating healing. If, however, we continue to run in compromised patterns for prolonged periods, newly adapted remedial strategies become progressively more plastically engrained within CNS and tissue architectures (Engineer et al, 2012; Avanzino et al, 2014). Accordingly, the dynamic inter-play between *experience-driven* and *competitive* plasticity processes ensures that traces of temporarily functional coordinative compensations –originating to remediate negative sensitisation–, commonly remain plastically embedded within neuro-biological structures. Subsequently, therefore, becoming the new ‘normal’, and exerting a legacy not easily erased within the abbreviated timeframes offered by conventional rehabilitation paradigms (Pelletier et al, 2015).

3.4.4 Promoting positive plasticity

Ultimately, plastic re-modelling, as it consumes precious material and energetic resources, is evolutionarily expensive (Merzenich et al, 2014; Clark, Schumann and Mostofsky, 2015). Within the adult brain it is not evolutionarily economical to plastically adapt to all stimulation —valuable neural reserves would be immediately depleted. Accordingly, structures in the mature cortex plastically remodel only when specific criteria —regulated by modulatory neurotransmitters such as acetylcholine, dopamine, serotonin, and norepinephrine— are satisfied (Merzenich et al, 2014; Clark, Schumann and Mostofsky, 2015). Operating collectively these neuro-modulatory enablers act as “on-off” switches, engaging excitatory and inhibitory processes and temporarily opening plasticity-enabling ‘windows of opportunity’; windows within which sensorimotor inputs contributing to ‘success’ are selectively amplified, while signals from competing inputs, uncorrelated with that success, are selectively dampened (Merzenich et al, 2014).

Over time, the continued amplification of relevant sensorimotor inputs provides a competitive advantage; greatly enhancing the representational detail embedded in the cortical territory dedicated to processing running-related sensorimotor information (Wolpert et al, 2011; Engineer et al, 2012). Crucially, a core finding emanating from this research domain is that repetitively non-varying, non-challenging ‘mindless’ movements —those not demanding focused attention for satisfactory execution— are insufficiently stimulating to reliably release the cocktail of neuro-modulating chemical catalysts necessary for plastic re-modelling within the mature motor cortex (Merzenich et al, 2014; Clark, Schumann and Mostofsky, 2015). In contrast, positive plastic re-modelling is optimized in response to behaviourally relevant, and intense, practice; practice that is typically executed at the limits of current abilities and therefore demands high attentional and motivational drives (Avanzino et al, 2014; Merzenich et al, 2014). Consequently, positive neural re-modelling is only promoted when tasks are neither so easy that they fail to stimulate focused attention, nor so difficult that continuous failure undermines motivation. In essence, coordination improves through engaging challenge, not mindless routine. *(A rationale perhaps explaining why rehabilitation processes employing non-challenging coordinative tasks typically fail to generate optimal recovery* (Elbert, and Rockstroh, 2004; Merzenich et al, 2014; Clark, Schumann and Mostofsky, 2015)).

3.5 Conclusions

As we accumulate running experiences, sensory feedback biases us towards personalized coordinative styles more satisfactorily resolving achievement of the running objective, against an acceptable investment of survival-relevant resources. Guided by innate evolutionary influences, individualized coordinative habits progressively form around our unique anatomical, biological, neurological and experiential idiosyncrasies.

As we progress from “novice” to “skilled” runners we more sensitively and smoothly respond to small perturbations, thereby offsetting the need to periodically and clumsily respond to larger challenges as minor errors accumulate. We adjust activation patterns to navigate away from discomforting sensation, thereby moderating tissue aggravations. We gravitate towards activations more proficiently poising bio-composite tissue structures to productively absorb and re-cycle impact momentums, thereby reducing energetic investment and dampening the negative consequences of excessive shock decelerations. We learn to exploit our layered landscape of degenerate movement options by fractally varying stride parameters under the integrated influence of historical events and current context, thereby dispersing running work-burdens amongst expanded webs of conditioned tissues. As we accumulate running experience, plasticity-processes progressively embed working relationships between regularly collaborating neural components, and embed the tissue features most adaptive to running-specific loadings. As such, plasticity is the mechanism that engrains synergies and linkages, and embeds the attractor states underpinning running coordination habits. A key observation, accordingly, is that running coordinative change is founded on a platform of plastic neuro-biological modification.

The evolutionary neuro-economics that embed efficient habits, however, eventually encase us within limiting constraints. Plasticity facilitates learning by engraining efficient habits, yet also retains the residues of past traumas and prolonged sensitivities; thereby, subsequently ensuring that injuries are rarely transient peripheral events, but long-lasting insults etched into the cortical tissues of the CNS (Elbert and Rockstroh, 2004; Pelletier et al, 2015; Coq and Barbe, 2011). Similarly, the enduring traces of repeated cycles of over-specialization, overuse, disuse and misuse, impose plastic re-configurations which do not automatically revert to original conditions once discomfort diminishes, and pain-free running is resumed.

Consequently, as we progress through our running lives, the sensorimotor landscape is in perpetual plastic flux as the integrated influences of general health, training and injury subtly re-configure neural connectivity and biological tissue architectures. Chronically, the progressive accumulation of plastic mal-adaptations drives the creeping decay of networked neural connectivity; thereby compromising sensorimotor information flow, blurring cortical representations of peripheral structures, prompting mal-adaptations in neuronal excitability, and driving disorder within the primary motor cortex (Coq and Barbe, 2011; Avanzino et al, 2014). Consequently, coordinative control inevitably degrades.

When other lifestyle and training considerations —background psycho-emotional stress, monotonous running volumes, generalized and localized fatigue—, are overlaid on already compromised operating conditions, access to expansive populations of viable movement degeneracies further diminishes. As this self-perpetuating cycle escalates, coordinative proficiency decays, susceptibility to tissue irritations grows and we become increasingly fragile to Overuse syndromes and non-formulaic perturbations.

3.5.1 Practical insights and relevance

Deeper appreciation of the various phenomena underpinning running coordination potentially informs many aspects of conventional theory and practice. The topics discussed below are offered as tentative examples:

3.5.1.1 Overuse injury

Documented incidence rates suggest running-related Overuse injury is neither a ‘solved’, nor perhaps even a clearly articulated, problem. Contextualising Overuse as a direct consequence of chronically compromised coordination emphasizes the necessity of balancing the monotonous stagnation, often implicit in conventional endurance running programs, with the unhabituated challenging stimulation essential to promoting positive neuro-plastic re-modelling. Furthermore, this rationale suggests that introducing coordinative diversity into high-volume running programs may be an effective prophylactic against the occurrence of Overuse injury.

3.5.1.2 Enforcing technical change

A deeper appreciation of the embedded undercurrents, shaping running coordination, also questions the long-standing coaching practice of striving to change technique simply by instructing the runner to consciously re-configure established coordination patterns, so as to better conform to an aesthetic ideal. Suddenly altering engrained running habits diverts mechanical stress along unhabituated pathways; thereby inevitably exposing unconditioned tissue to unaccustomed loadings and elevating injury risk. Although validating empirical evidence remains scarce, there is a suggestion of rising injury rates following short-term technical interventions (see, for example, Tucker, 2007).

3.5.1.3 Driving neuro-plastic change

Crucially, the perspective presented here suggests we should perhaps pay less attention to how running styles look, and more attention to designing interventions that provide the coordinative challenge necessary to stimulate the neuro-plastic re-modelling necessary to refine communicative clarity between the CNS, and the peripheral musculature. Although such interventions typically fall outside the scope of conventional run-training dogma, many coaches, past and present, have intuitively designed training practices fulfilling the criteria for optimally stimulating neuro-modulatory processes (see for example: Kiely, 2013; Pfaff, 2014; Smith, 2014). What emerging scientific insight does add, however, is a growing appreciation of the value of regularly challenging running coordination through the design and implementation of appropriately constructed practices.

3.5.1.4 Visual evaluation of running technique

Conventionally, we associate running coordination with running technique — the visual evaluation of running style evaluated against an aesthetic ideal. This pervasive assumption, however, has never been satisfactorily demonstrated, and no empirical evidence supports a direct relationship between looking ‘better’, and actually being ‘better’.

When we visually assess a runner’s technique and extrapolate these observations to running efficiency and injury risk conclusions, we make judgments based on very superficial information. Typically, we fail to acknowledge the unseen underlying terrain —the idiosyncratic neurology; the embedded fractal signatures; the unique anatomical architectures and tissue structures; the plastically-personalised legacy of historical habits and traumas—upon which all coordinative habits are founded. Although it may be feasible that, to the highly practiced eye, visual evaluation may provide clues; generally, how these clues are interpreted is rooted in assumptions lacking an evidence base. Certainly, visual assessments of running proficiency seem unavoidably subjectively biased, and previous investigations demonstrate extensive inter-individual differences in technical ratings between coaches, and even when the same coach evaluates the same footage at different times (Norris, Anderson and Kenny, 2014).

Would performances improve if running form more closely conformed to perceived technical ideals? Are more aesthetically pleasing runners less injury prone; more economical? While opinions are plentiful, evidence is scarce. *(Anecdotally, renowned coach and physiologist, Dr. Jack Daniels, once sent video of 20 physiologically-evaluated competitive runners to a selection of coaches and exercise scientists, asking them to —on the basis of visual inspection— rank athletes in order of running economy but, "they couldn't tell, no way at all" (Kolata, 2007).*

3.5.2 What is running coordination?

The perspective offered within this chapter is that coordination is the overarching super-capacity ultimately orchestrating how proficiently neural, muscular, cardiovascular and metabolic reserves are purposefully harnessed, or wastefully squandered. Specifically in relation to running, coordination is the learned deployment of available neuro-biological resources to satisfactorily realize running objectives for an acceptable 'cost' —in terms of depletion of energetic and neural reserves, and exposure to risk. Running coordination, accordingly, is the physical expression of a confluence of psychological, emotional, neural and biological constraints emerging in response to the on-going interplay between intention, motivation and perception of risk; informed by emerging sensory feedback; modulated by prior experiences and expectations; biased towards repeatedly re-employing plastically-embedded coordinative solutions to current running 'problems'.

Ultimately, running performance is underpinned by a conglomeration of assorted capacities. Yet it is the super-capacity of coordination that regulates how proficiently these overlapping performance contributors are collaboratively expressed; to generate propulsive power, to promote efficiency and preserve robustness, and to accomplish running objectives for an acceptable exposure to discomfort and risk. A deeper appreciation of the underpinnings of the running coordination phenomenon may enable practitioners to more judiciously design interventions to promote, nurture and preserve coordinative proficiency in the face of the inevitably accumulating *'wear and tear'* endured over the course of a running lifetime.

3.6 In summary

1. Dimensions of coordinative control persistently vary, in structured, non-random ways, to strategically manage both the control and mechanical burdens associated with running
2. The sensorimotor landscape, underpinning running coordinative control, is in a state of perpetual flux in response to both acutely and chronically modulating constraints
3. Coordination is the over-arching, super-capacity responsible for regulating the expression of all other performance-related sub-capacities

This chapter highlights the ways and means that coordination proficiency gradually accrues, and gradually decays. A core running robustness enabling capacity is the ability to distribute the running 'work burden' amongst conditioned populations of neural control modules, and biological tissue collectives. Of further relevance, current evidence illustrates that when coordination is compromised at one location, due to injury for example, remedial coordinative adaptations may occur elsewhere. In essence, changes in local function are likely to induce coordination compensations dispersed throughout the running system. Finally, the evidence and rationale within this chapter suggests that coordination is the over-arching, super-capacity, ultimately responsible for regulating running performance and injury resilience.

Chapter 4:

Running Stiffness: The elegant resolution of multiple threats and opportunities

4.1 Managing the shock of running impacts

Uniquely amongst mammals, humans run in an upright bouncing gait characterized by long flight times, short ground contact times and single leg landings imposing impacts of multiple times bodyweight. While this running style creates considerable challenges, in terms of managing the repetitive destabilizing shocks imposed by ground contact, while simultaneously maintaining directional coherence and forward momentum, it also offers benefits. Specifically, mechanical energy is absorbed during the shock of contact, as bodily tissues deform during the braking phase, and is partially restored by an elastic rebound when the body reaccelerates during push-off (Schepens, Willems, Cavagna, and Heglund, 2001; Legramandi, Schepens, and Cavagna, 2013). Thus, impact energy is partially conserved during human running, thanks to an *elastic bounce,* which serves to greatly reduce metabolic costs (Legramandi et al, 2013).

The physical deformation, imposed during impact, is manifest at multiple levels. At the macro-level, ground collision causes a partial postural collapse; at the level of the muscle-tendon unit (MTU) visco-elastic tissue structures are elongated or compressed; while at the cellular level, micro-filaments are similarly distorted and distended. Collaboratively, these nested deformations elicit disproportionate, non-linear, restorative responses to suddenly imposed mechanical perturbation (Kiely and Collins, 2016; Maloney and Fletcher, 2018). During running, the rate and extent of these multi-level structural deformations, and subsequent restorations, are regulated by running stiffness –the capacity of the runner to resist deformation following the sudden application of ground contact forces.

Consequently, the running body can be modelled as a point mass, balanced on a compressible spring of specific stiffness (k), and the joints modelled as torsional springs each having a distinct stiffnesses (Lorimer and Hume, 2016). The bipedal spring-mass model is a very simple model of legged locomotion, in contrast to other more complex representations, requiring only two essential features: bipedalism and leg compliance (Pandy 2003). Despite its simplicity, however, this model accurately captures key dimensions of gait mechanics, and usefully approximates how the lower body interacts with the ground during landing (Lorimer and Hume, 2016).

There remain, however, within the literature, multiple inconsistencies relating to how distinct dimensions of the stiffness phenomenon are defined, measured, modelled, labelled and interpreted (Serpell et al 2012; Hébert-Losier and Eriksson, 2014; Lorimer and Hume 2016). Consequently, many investigations have modelled and measured stiffness using different methodologies, evaluated performance using different assessments and movements, and frequently used specific terms interchangeably and/or in varying contexts.

Amongst sports coaches, clinicians and conditioning professionals, there is a general appreciation that the term stiffness broadly describes an athlete's ability to resist postural deformation during the sudden vertical deceleration caused by ground contact, and their subsequent capacity to harness and recycle some of this mechanical energy to contribute to on-going movement. Consequently, athletic stiffness is considered a core attribute across the range of sporting activities involving running and jumping; accelerating and decelerating; agility and change of direction. Given the terminological debates, methodological distinctions, and the subsequent confusion of context-specific empirical observations; unsurprisingly, while there is a general awareness that stiffness is an important attribute, there is no coherent, shared, explanatory framework encapsulating exactly 'why' stiffness is so important and precisely 'how' stiffness is regulated.

Ultimately, stiffness is a deceptively simple, albeit ambiguous, concept that adequately describes a complex phenomenon. Yet while technical conversations relating to definitional distinctions, assessment methodologies and measurement technologies are necessary (and, needless to remark, more research is needed); to optimally foster insight and understanding, we also need a deeper appreciation of the phenomenology underpinning the 'why' and 'how' of stiffness. The focus of this article, accordingly, is to explore the deep roots of the running stiffness phenomenon and to suggest how a refined conceptual vision may contribute to a more holistic appreciation of human running coordination.

4.2. Stiffness: Confusion, controversy and inconsistencies

4.2.1 What is stiffness?

Conventionally, the concept of athletic stiffness is typically decomposed into a number of variously defined ‘stiffness’ parameters; commonly segregated into three categories:

1. Vertical (or system) stiffness – the resistance of the body to vertical displacement after application of ground reaction force–; typically calculated as the quotient of ground reaction force and centre of mass (CoM) displacement
2. Leg stiffness – the resistance to changes in leg length on application of internal or external forces; calculated as the quotient of ground reaction force and change in leg length
3. Joint stiffness – the resistance to change in angular displacement subsequent to the application of joint moments calculated as the quotient of joint moments and change in joint angle (Brughelli and Cronin, 2008; Butler et al, 2003)

4.2.2 Definitions, measurements and calculations

Evident, within the literature, however, are significant discrepancies in how distinct manifestations of stiffness are defined, measured, labelled and interpreted (Serpell et al, 2012; Lorimer and Hume, 2016). Multiple methods for modelling and calculating stiffness exist, and stiffness values typically vary widely depending on both the activity under consideration, and the assessment technologies and computational methods employed. Furthermore, investigations in this field are typically characterized by small sample sizes and large confidence intervals; resulting in effect sizes which remain open to wide-ranging interpretations (Serpell et al, 2012; Hebert-Losier et al, 2014; Maloney and Fletcher, 2016; Lorimer and Hume, 2016).

Most notably, although vertical and leg stiffness are definitionally distinct phenomena, these terms are often conflated and used synonymously and interchangeably. In such cases Centre of Mass (CoM) displacement is used to calculate, what is elsewhere defined as, vertical or system stiffness; yet is labeled leg stiffness. In contrast, in other contexts, leg stiffness is calculated directly via changes in leg length (Serpell et al, 2012). There is, however, an important distinction between vertical and leg stiffness, in that employing CoM displacement as a surrogate of leg stiffness assumes a rigid body superior to the hips which, given the inevitable flexion and extension of the trunk during running, is clearly an unrealistic assumption (Serpell et al, 2012; Hebert-Losier et al, 2014; Maloney and Fletcher, 2018). Further adding to this confusion, some consider vertical stiffness across the entire gait cycle (i.e. stance and flight); whereas others contend that stiffness can only be measured during the active structural compression occurring during ground contact (Farley and Gonzalez, 1996). It is worth noting that a diversity of local stiffness measures –the stiffness's of specific muscles, tendons and/or particular joints— are also commonly assessed. The relationships, influences and effects of these local variables on global stiffness, however, remain ambiguous (Lorimer and Hume, 2016).

There is also disagreement surrounding the question of which joint plays the most important role in modulating leg stiffness. Various studies, for example, have reached contrasting conclusions concerning whether the knee or ankle joint exerts the most influence on leg stiffness; whereas other investigations could not differentiate between the two (Serpell et al, 2012). As a final complication: the fundamental purpose of running stiffness remains contentious. Some, for example, consider stiffness as primarily a means of recycling elastic energy (Fiolkowski, Bishop, Brunt and Williams, 2005; Roberts and Azizi, 2011); whilst others emphasise its importance in maintaining stability subsequent to perturbation (Lewis, MacKinnon and Perrault, 2010). These perspectives, however, assume that stiffness evolved as a behavioural innovation to solve a specific movement problem, or at least a narrow range of movement problems. This position may, however, be limited in the scope of its vision.

4.2.3 Stiffness and injury

It seems sensible to suspect that, in impact-related sporting activities such as running and jumping, the key injury-inducing event occurs during ground contact. Accordingly, a number of impact-related variables, measured at ground contact, have been implicated with an increasing injury risk (Serpell et al, 2012; Lorimer and Hume, 2016). Prominent amongst these variables are the magnitudes of ground reaction forces, sudden decelerations, and the frequency of tissue vibrations subsequent to ground collision (Matijevich et al, 2019). Each of these variables are clearly influenced by the momentary stiffness of the athletic system during ground contact and, as you might expect, previous studies and research reviews have concluded that stiffness plays some, as yet unresolved, role in modulating the incidence of athletic injuries (Brazier et al, 2014).

Historically, the pervasive assumption was that elevated stiffness values provoked an increasing likelihood of injury. More recently, however, a number of retrospective studies have suggested that excessively high or low levels of leg stiffness both serve to increase injury likelihood. Interestingly, the available research suggests that the nature of injuries, caused by either high or low stiffness, differs. Specifically, it is suggested that excessive stiffness leads to an increased prevalence of bony and ligamentous pathologies, such as tibial stress fractures and lateral ankle sprains; whereas low stiffness values are associated with a higher incidence of soft tissue injuries, such as muscle strains and patellofemoral pain syndromes (Williams et al, 2004; Napier et al, 2015; Maloney and Fletcher, 2018).

These conjectures are logically appealing. If lower body stiffness is excessive during ground contact, then shock loadings transmitted to bony and soft structures are inevitably elevated. Conversely, an unduly compliant landing, caused by a lack of stiffness, inevitably results in excessive postural collapse, tissue length changes and deformational accelerations; thereby leading to an increased likelihood of repetitive soft-tissue micro-trauma. Nevertheless, it should be noted that, given the inconsistencies evident in the existing research, these associations remain inadequately supported and, currently, there are no prospective studies directly correlating stiffness with injury, or type of injury, incidence.

4.2.4 Stiffness and performance

High levels of lower extremity stiffness –during running, jumping and hopping tasks– have been positively associated with a variety of athletic performance outcomes, such as sprinting and long jump performance (Brazier et al., 2014). Previous studies also established that vertical stiffness increased in tandem with both running velocity and stride frequency (Farley and González, 1996). However, although some studies have concluded that leg stiffness increases in line with vertical stiffness and running speed; other studies have not observed parallel increases in leg stiffness and running velocity (Cavagna, et al., 2005; Morin, et al., 2005). As further complication: currently it remains unclear whether increases or decreases in stiffness are dependent on the site assessed. Thus, for example, while Arampatzis and colleagues (1999), and

Kuitunen and colleagues (2002) observed increases in whole-body and knee stiffness with increasing running velocity, there was little or no coincident change in ankle stiffness. Elsewhere, however, ankle stiffness increased in tandem with running velocity (Günther and Blickhan, 2002; Maloney and Fletcher, 2018).

Furthermore, previous work demonstrated that power athletes exhibited greater leg stiffness than endurance trained athletes (Hobara et al, 2008), and that endurance trained athletes demonstrated greater stiffness, at the ankle, knee and hip—assessed over 5 consecutive double-legged hops—, than untrained controls (Hobara, Kimura, et al, 2010). Similarly, amongst a group of 22 sub-elite footballers, those having higher vertical stiffness values demonstrated better performances across a number of athletic assessments –comprised of sprint, jump and agility activities–, than their less stiff peers (Kalkhoven and Watsford, 2018).

4.2.5 Section summary

The predominance of existing literature suggests that regulation of stiffness exerts some influence on both athletic performance potential, and injury susceptibility. In relation to performance, the weight of evidence suggests that higher stiffness measurements are associated with enhanced athletic outcomes. Furthermore, lower extremity stiffness seems to increase –across impact-related activities such as hopping frequency, jumping height, and running speed–, in tandem with the demands of the activity (Butler et al., 2003).

More specifically, in relation to running, the majority, but not all (for example: Heise and Martin, 1998), studies suggest that increased lower limb stiffness promotes running economy (Butler et al, 2003). In contrast, others conclude that excessive stiffness impedes running performance; via interference with the runner's capacity to optimally absorb, re-direct and recycle elastic energy to purposefully contribute to upcoming movement demands (for review see: Legramandi et al, 2013). Such rationalization provides a mechanistic route through which an excessively stiff system may interfere with the runner's capacity to derive an optimally efficient *elastic bounce.* Furthermore, the up-regulation of joint, limb and/or whole-body stiffness, via increased muscular co-contractions, inevitably adds to the metabolic cost of transport.

4.3. Running stiffness: Moving towards a more nuanced perspective

Evolutionary survival demands that biological systems, operating in unpredictable environments using unreliable components and finite energy sources, are robust to the range of challenges to which they are most commonly exposed (Kitano, 2004). These 'threats' take many forms. Historically, and perhaps arbitrarily, it was assumed that the central over-riding constraint shaping running coordination was the conservation of energy, and that this conservation was achieved by reducing the amount of muscular work necessary to power running actions. This perspective was encapsulated within the *Minimum Energy Hypothesis* and was widely accepted in the relevant literatures (for example: Cavagna et al, 1964; Shen and Seipel, 2015). Although this position was sporadically challenged, it has long remained the predominant dogma.

In recent years, however, accumulating evidence illustrates that other considerations exert an influence on how complex human movements are coordinated. Currently, although it is clear that energy expenditure –relative to specific performance criteria– is typically reduced as movement proficiency improves with practice, it is certainly not minimized (Wolpert and Diedrichsen, 2011). Instead, energy conservation appears to be but one of a number of competing *cost functions*; all of which must be astutely and simultaneously resolved in a complex series of negotiations and trade-offs to facilitate safe and proficient running ((Wolpert and Diedrichsen, 2011).

4.3.1 The multiple overlapping challenges of running

At the macro-task level, the core control imperatives of running are typically described as the maintenance of desired speed and direction (Daley and Biewener, 2006), injury avoidance (Roberts et al, 1998; Birn-Jeffery et al, 2014) and the conservation of energy (Shen and Seipel, 2015). At a more micro-level of observation, however, a number of other notable imperatives, or *optimality criteria,* have been suggested. Such considerations include, for example: minimum jerk (Flash and Hogan 1985); minimum torque-change (Uno et al. 1989); minimum effort (Hasan 1986); minimum discomfort; minimum muscle activation; the regulation of movement accuracy; as well as other, more complex, cost functions (for example, Rosenbaum et al. 2001). Similarly, existing evidence highlights that sparing precious neural resources and reducing cognitive load —achieved by out-sourcing routine running-related dimensions of control to progressively lower neural control centres and/or innate tissue properties— also serve to shape and constrain running coordination (Kiely, 2017).

4.3.2 The mechanical threats imposed by impact

Most obviously, however, the fundamental mechanical threat posed by running arises from the impact associated with ground collision. This violent deceleration creates an impact shockwave that reverberates, via the bones and soft tissues, throughout the runner’s body; thereby presenting a series of immediate, and simultaneously overlaid, challenges and threats.

4.3.2.1 Ground reaction force

The most readily apparent, and most empirically amenable, of these mechanical threats is the sudden imposition of ground reaction forces (GRFs) upon collision between the ground and the running foot. However, the assumption that the GRFs measured at the point of ground contact directly and accurately reflect the subsequent internal mechanical stresses inflicted on bony and soft tissues, remains speculative. As previously noted, the magnitude of GRFs applied at the interface between ground and the running body can be immediately and dramatically dampened, re-directed and re-deployed via a number of coordinative mechanisms (Wilson and Kiely, 2016; Nigg et al, 2017; Matijevich et al, 2019). Accordingly, in relation to running injury applications, externally imposed forces may be of little diagnostic consequence (Nigg et al., 2017). Supporting this conjecture, a recent empirical investigation found only weak correlations between GRFs during running – assessed across a range of speeds and slopes–, and tibial bone loading (Matijevich et al, 2019). The authors subsequently concluded that GRFs provide limited utility for predicting running-related injury risks. This position, however, remains controversial, and it may simply be the case that GRF data needs to be more perceptively blended with other running-related metrics to provide greater interpretative value (Clark et al 2017).

4.3.2.2 Tibial accelerations

Peak tibial shock —the highest vertical acceleration recorded at the tibia during stance—, is typically assessed via accelerometers attached to the anteromedial tibial surface and is a commonly used approximation of mechanical stress during ground contact (Milner et al, 2006; Sheerin et al, 2018). This measure is closely related to both GRFs and running kinematics, and is strongly correlated with vertical loading rates (Milner et al, 2006; Gruber et al, 2014). Although shock is a surrogate measure of loading, previous work suggests tibial shock actually provides a more direct and informative estimate of the mechanical stress exerted on the tibia than GRFs (Milner et al, 2006).

Specifically, in relation to peak tibial accelerations and injury, prior evidence has linked increasing vertical tibial shocks to an increasing incidence of tibial fatigue fractures in runners (Milner et al, 2006; Pohl et al, 2008). Similarly, differences in the magnitudes of vertical tibial shock adequately distinguished between runners with, or without, tibial fatigue fractures and enabled differentiation between runner’s previously injured and uninjured limbs (Pohl et al, 2008). Consequently, it is suggested that the likelihood of tibial fatigue fracture increases by a factor of 1.4, for every 1 g increase in peak vertical tibial acceleration (Milner et al, 2006). Supporting this perspective, previous investigations concluded that greater vertical shocks were more strongly associated with increased running-related injury (RRI) risks than peak vertical forces (Van Gent et al, 2007; Liberman et al, 2010), and suggest that runners who habitually experience greater vertical shocks are more vulnerable to overuse injury; subsequently leading others to conclude that shock loading is the single most informative indicator of RRI probability (Davis, Bowser and Mullineaux, 2016; Zifchock et al, 2008; Hreljac, 2004).

Despite such conjecture, however, the relationships between peak vertical tibial accelerations and running performance remain unclear and under-explored (Zifchock et al, 2008). Confusingly, peak tibial accelerations appear to vary under the influence of multiple modifiers. For example: tibial accelerations modulate based on individual runners footstrike characteristics (Giandolini et al, 2016); females typically demonstrate higher mediolateral accelerations when menstruating, as opposed to when ovulating (Clark et al, 2010); and peak tibial accelerations have been noted to change in tandem with varied stride parameters (Laughton et al, 2003). Notably, and relevantly, however, investigations evaluating changes in tibial shock over the course of fatiguing runs have exhibited mixed results. Abt and colleagues (2011), for example, reported no changes in any kinematic or acceleration-dependent variables subsequent to an exhaustive treadmill run (Abt et al., 2011). In contrast, other studies concluded that tibial shocks did increase with mounting treadmill running-induced fatigue (Mizrahi et al, 2000). Unclear findings were reported in a study where fatigue effects on vertical tibial shock were compared when runners ran both overground, and on a treadmill (García-Pérez et al, 2014). These illustrations highlight that the available evidence, relating to whether fatigue induces changes to locally experienced shock accelerations, remains unclear. This ambiguity recently led Sheerin and colleagues (2018) to conclude that, due to confounding influences –such as the use of different fatiguing protocols and different running populations–, there is no clear consensus on how global and localized fatigue influence accelerative shocks experienced at the tibia (Sheerin et al, 2018).

Over twenty-five years ago Lafortune and Hennig (1995) suggested that accurately quantifying the total accelerative shock applied to the musculoskeletal system, during running, required the integration of all three directional (vertical, mediolateral and anterior/posterior) components via the calculation of a peak triaxial resultant acceleration (RA).

Nevertheless, twenty years later Gandolini and colleagues (2015), following review of all studies investigating running impact shock over the previous 2 decades, concluded that the overwhelming majority of published research projects had focussed exclusively on vertical impact, and although a small few had considered mediolateral accelerations none had reported antero-posterior peak accelerations (Gandolini et al, 2015). A potential additional benefit of using the RA method is that, as resultant metrics incorporate accelerations in 3-dimensions, the alignment of the accelerometer axes is not as sensitive to error as is be the case when assessing a single acceleration measure (Sheerin et al, 2018). Accordingly, the device may be used by coaches, runners or others, who may not have experience or expertise in aligning measurement devices and underlying anatomy.

However, despite this theoretical substantiation, due to the sparsity of available evidence it remains unclear whether the integrated product of vertical, medio/lateral and anterior/posterior tibial accelerations, represented by the RA metric, provides a useful metric for assessing fatigue during running activities (Sheerin et al, 2018). Similarly, the question of whether triaxial resultant accelerations modulate, subsequent to fatigue, at anatomical locations other than the tibias remains unanswered. Although some investigations have assessed uni-, bi- or triaxial accelerations at the pelvis, the literature is sparse. As yet, no published study has quantified the RAs experienced simultaneously at multiple anatomical sites during fatigued running.

4.3.2.3 Vibration load

Immediately after ground contact an impact shockwave propagates upwards through the runner's body, causing the viscera to bounce and adipose tissues and un-tensioned muscles to 'wobble' (Christensen et al, 2017; Chadefaux et al., 2019 REF). Inevitably, bodily tissues of different architectures, masses, densities and background tensions, will experience different accelerations, relative displacements and structural deformations. These differential mechanical experiences ensure that skeletal sites and soft tissue compartments vibrate relative to each other (Friesenbichler et al, 2011). These vibrations, inevitably, expose tissues to yet another form of mechanical stress, and excessive exposure to vibration loads –in terms of magnitude and/or duration–, during running, has been linked to numerous negative outcomes (Wakeling et al, 2003). Interestingly, and relevantly, vibrations have also been observed to increase in tandem with mounting running-related fatigue (Friesenbichler et al, 2011; Castillo and Lieberman, 2018).

4.3.2.4 Maintaining dynamic stability

The placement of our delicate brains in the very position where they are most exposed to falling injury seems, from an evolutionary perspective, an uncharacteristically risky survival strategy. It seems inconceivable that *Homo Sapiens* would not simply survive, but actually thrive as a species of regularly and voluntarily enthusiastic runners, unless Nature installed significant safeguards mitigating against the probability of neurological trauma. Subsequently, it seems clear that the preservation of dynamic stability is an essential, and highly prioritized dimension of human running coordination. Supporting this perspective, recent work demonstrated that energy efficiency and dynamic stability both serve as mutually integrating influences shaping the leg stiffness of running quadrupeds (Birn-Jeffery et al, 2014). Here, the authors concluded that leg stiffness values typically hover within a range which simultaneously deliver both a reasonable energetic cost, and an acceptable level of dynamic stability (Shen and Seipel, 2015).

4.3.2.5 Disruption of neural communication clarity

An additional, but less commonly considered, constraint, relates to the fact that sensorimotor processing capacity —our neural systems ability to integrate sensory feedback information with feedforward activation commands— is highly sensitive to excessive supra-spinal accelerations (Gruber et al, 2014). Accordingly, prior investigative work has concluded that –to preserve the functional integrity of critical neural processing during running–, the brain must be adequately insulated from the deleterious consequences of insufficiently attenuated impact shockwaves (Mercer et al., 2002; Gruber et al, 2014).

4.3.3 Section summary

Theoretically, it seems sensible to speculate that if the runner's musclo-skeletal system is too rigid upon ground contact, then neural and biological tissues will be exposed to excessively severe, un-attenuated shock. Experimentally supporting this conjecture, prior investigations have demonstrated that high impact shocks increase injury risk in runners (Zadpoor and Nikooyan, 2010). Nevertheless, although the weight of evidence seems to support this perspective, it remains unclear which specific dimension of the impact event is most responsible for driving RRI incidence (Legramandi et al, 2013; Pantoja et al, 2016). What is clear, however, is that the available evidence base remains incomplete, inadequate, ambiguous and frequently conflicting (Serpell et al, 2012; Hebert-Losier et al, 2014; Maloney and Fletcher, 2016; Lorimer and Hume, 2016).

4.4 Regulating running stiffness: Resolving the multiple challenges of impact

The preceding discussion illustrates that our conventional rationalization of the stiffness phenomenon, within academic and coaching domains, requires re-alignment with contemporary evidence. 5 observations seem especially worthy of note.

4.4.1 Stiffness is a 3-dimensional phenomenon

Clearly, although stiffness is commonly measured as a single axis, vertically-oriented phenomenon, the mechanical challenges emanating from ground contact are inevitably 3-dimensional. Recent research has evaluated this perspective, during running, by including measures of deformation in the medio-lateral plane. Notably, Liew and colleagues (2017), comparing traditional and multiplanar measurements, reported that the inclusion of medio-lateral measurements significantly increased estimated leg deformation subsequent to ground contact; thereby lowering leg stiffness estimations. These researchers concluded that the inclusion of multiplanar data is likely to provide a more accurate and comprehensive estimate of leg stiffness (Liew, Morris, Masters and Netto, 2017).

4.4.2 Stiffness is a context-specific coordinative skill

Conventionally, stiffness is presented as a product of the viscoelastic properties of the joint, and/or the connective tissues, and/or the co-contraction of antagonist muscles (Ludvig and Kearney, 2007). These local factors are typically considered to be dictated by the structural capacities of specific tissues or specific tissue collectives. However, the perspectives presented here emphasise a more complex picture; whereby stiffness is more aptly perceived as a complex coordinated collaboration between passive tissue and active neural control mechanisms; which, when skilfully blended, can deliver instantaneously reactive (zero lag-time) responses to suddenly imposed mechanical perturbations (van der Krogt et al, 2009; Kiely and Collins, 2017). Subsequently, the skilled regulation of stiffness provides a mechanism whereby – simply by manipulating the geometry and the background tensions of tissue collectives– energy can be variously stored, transferred, dissipated, re-directed and subsequently re-cycled via the practiced deployment of structural tissue properties.

4.4.3 Stiffness is a multi-faceted coordinative collaboration

Accordingly, a fundamental dimension of the skill of running is the careful regulation, via directed muscular activation, of the behaviors, postures, and background tensions of tissue collectives in preparation for, and during, ground contact. To this end, the neuro-biological system gradually learns, through practice, to sensitively calibrate the tensional integrity of the runner’s body to the estimated mechanical and energetic challenges likely to emanate from the upcoming foot-strike.

This skillful calibration of global and local stiffness's subsequently provides an agile, adaptive and unified solution to the seemingly distinct coordinative challenges imposed during ground contact.

4.4.4 Stiffness is context-specific and task-dependent

Aside from the confusion caused by overlapping definitions, assessments and interpretations of stiffness, an additional confounding factor is that running stiffness is typically presented and communicated as a distinct measurable capacity. Yet, in relation to running and jumping activities, stiffness is demonstrably extremely sensitive to precise task and context-specific parameters. Prior research, for example, illustrates that stiffness evaluations vary in accordance with age; gender; foot architecture; running velocities; running surfaces; running surface predictability; footwear; foot-strike patterns; leg length and orientation; stride frequency, and external load carriage (Ferris et al, 1999; Williams et al, 2004; Silder et al, 2015; Lorimer and Hume, 2016; Pantoja et al, 2016; Wang et al, 2015; Murphy et al, 2013). Notably, stiffness also modulates with training status and performance level; between novice runners, experienced runners and non-runners; and between endurance runners and sprinters (Hobara et al, 2008 and 2010). Stiffness, accordingly, is modified by multiple variables, operating across multiple timescales, as dictated by both externally-imposed, and internally-generated, constraints. As such, the applicability and appropriateness of any given stiffness measure is extremely context-dependent.

4.4.5 Stiffness changes with fatigue

Multiple stiffness-dependent variables modulate as fatigue increases. The capacity to accurately tune running stiffness to the demands of the running task diminishes, for example, as fatigue progresses. Subsequently, the intensity of tissue vibrations increases; thereby exposing soft-tissue compartments to increased vibrational stress (Friesenbichler et al 2011; Khassetarash et al, 2015). Similarly, increases in tibial accelerations (sometimes by as much as 100%) have been noted towards the end of fatiguing high-intensity treadmill runs (Sheerin et al, 2018); thereby leading some researchers to conclude that exercise-induced fatigue plays a driving role in tibial stress fractures and multiple other manifestations of RRIs (Sheerin et al, 2018).

4.4.6 Section summary

If these debates highlight any one issue, it is that the running stiffness phenomenon is currently neither clearly delineated, nor one that encompasses current despite the apparent relevance of stiffness to both running performance and injury, our current understanding of the stiffness phenomenon remains incomplete.

4.5 Conclusion: The coordinated skill of running stiffness

4.5.1 Herbert Simon, satisficing and the myth of optimality

Evolution is commonly portrayed, in both popular and academic literatures, as an 'optimising' phenomenon –an iterative, gradually progressive, trial-and-error, problem-solving process through which 'best' answers eventually emerge. This presumption of *optimality* has, through cultural osmosis, gradually seeped into many sport science applications (for fuller review: Davids and Araújo, 2010). Yet, in the context of complex biological lifeforms –whose adaptive functionality depends on the integrated outcomes of multiple super-imposed, parallel, yet differently focussed, processes–, inevitably and by necessity, the optimization of any single output excludes the optimization of all others (Vilarroya, 2012; Hochberg, 2017). In the context of complex phenomena, accordingly, it is unclear how the concept of optimization can even be coherently defined –because, as evaluations of 'optimal' are always perspective-specific and context-dependent, there can be no single 'optimal' solution without first defining the specific aspect of the task to be optimised. Striving for optimization, in a multi-faceted task imposing multiple simultaneous challenges, consequently, seems a hopelessly impractical, inefficient and excessively expensive evolutionary search strategy (Hochberg, 2017).

In rationalising this observation, Nobelist and Turing award winning complexity theorist and artificial intelligence pioneer Herbert Simon (1955, 1956) suggested that, rather than committing resources to the complex (and ultimately futile) quest for ‘optimized’ solutions, evolution, instead, favours expediency. Expediency, in the context of a neurobiological search strategy, entails selecting the first available option that adequately accomplishes the objectives of all simultaneously imposed target tasks for an acceptable uptake of critical resources, for an acceptable level of risk, and within an acceptable timeframe. Simon introduced the term ‘*satisficing’* –a fusion of ‘satisfactory’ and ‘suffice’– to describe this process and argued that instead of dedicating precious resources to solving a hopelessly complex ‘optimisation’ problem, evolution instead chooses a ‘satisficing’ path. A path that, rather than seeking to find the single best solution to a single facet of a specific challenge, seeks to sufficiently and simultaneously satisfy a spectrum of multiple, concurrently imposed, overlapping tasks; a path that, in Simon’s words, *“will permit satisfaction at some specified level of all its [the organisms] needs”* (Simon, 1956).

The key insight of Simon’s rationalization is that evolution does not waste precious time and resources searching for optimal answers. Instead, evolution optimises the expediency of the search strategies from which satisfactory and sufficient answers naturally emerge (Simon, 1990; Berniker et al, 2013). Satisficing problem-solving approaches are subsequently eminently pragmatic under conditions of imperfect information, implicit uncertainty, pressurized time-constraints and when a near-limitless choice of feasible permutations –all capable of equitably providing viable, acceptable and resource-efficient outcomes– exist, *in potentia*, within the system.

4.5.2 Running stiffness: the dynamic resolution of negotiated trade-offs

When running, during the ground contact event, we must regulate dynamic stability; manage impacts of multiple times bodyweight; steer direction of travel; protect neural processing modules from the negative consequences of excessive momentums, accelerations and vibrations; generate sufficient propulsive forces to perpetuate momentum, and retain adequate maneuverability to escape suddenly imposed, unforeseen perturbations. Current tissue lengths; rates of tissue length change; relative joint angles; relative positioning's of the foot, ankle, knee, hip and torso; angle of postural inclination; speed of foot retraction prior to ground contact; quasi- and short-range muscle stiffness; uncertainty relating to predictions of surface integrity, and a near-limitless procession of other influencing factors, all interact and integrate to modulate the runners momentary stiffness during foot-strike. The skillful exploitation of this multiplicity of movement choices ensures that equivalent stiffnesses can be realized using an infinite array of subtly different kinematic and kinetic permutations.

The emergent regulation of running stiffness, accordingly, fulfills the descriptive criteria of a *satisficed* solution; in that the neurobiological system simultaneously resolves multiple partially overlapping, partially competing task-level demands, organisational constraints and *optimality* criteria to offset the host of threats, challenges and opportunities inherent in the ground contact event (Hodges and Tucker, 2011; Miller et al., 2012). Through this lens it no longer makes sense to assume that stiffness optimizes one specific running cost function. If running efficiency is maximised, for example, the risk of breakdown escalates. If stability is optimised, by increasing co-contraction around joints, metabolic efficiency is compromised. If energy depletes; if fatigue excessively accumulates; if tissue tolerances are exceeded; if vibration loads are excessive and/or prolonged; if neural processes are overloaded to the extent that injury risk increases due to eroding movement precision and/or a loss of local or global stability and/or if cognitive clarity declines due to increasing competition for limited neural resources; then, inevitably, survival probability is diminished. No single survival imperative necessarily predominates. Instead, there is always an evolutionarily-informed weighing of threats against benefits; a weighing of complex trade-offs between performance, stability, manoeuvrability and injury risk and between neural and energetic investments and rewards. An undue emphasis on a single constraint; inevitably detracts from the maximal realisation of another. There is no optimal; there is only the persistent, dynamically coordinated decision-making triangulation between accessibility, adequacy and immediacy.

4.6 Conclusion

Ultimately, the management of running stiffness is a learned, coordinative skill serving to simultaneously, sufficiently and satisfactorily resolve multiple impact-imposed challenges, opportunities and threats via the practiced manipulation of local and systemic mechanical rigidities and compliances. Stiffness emerges as the coordinated resolution of internal (neural, energetic and tissue capacities) and external (running surface, leg kinematics) constraints. When constraints change (such as when local sensitization and/or centrally mediated fatigue increases), running stiffness modulates in search of a new, more appropriate, solution to this changed set of circumstances. Stiffness, accordingly, is a highly sensitive, highly adaptive, closely calibrated and context-specific coordinated response serving to simultaneously solve a host of super-imposed running challenges; it is the evolutionary-enabled orchestration of the momentary rigidities and pliability's of tissue collectives to variously direct, divert, absorb, amplify, store and subsequently re-deploy mechanical loadings and energy flow to satisfy the control, safety and energetic demands imposed during running activity.

When we manage running stiffness poorly, because we are insufficiently practiced or because coordination is inhibited or compromised by fatigue or neuro-mechanical limitations, we are both energetically wasteful and vulnerable to tissue damage. When we manage stiffness well, however, we mitigate risk and productively exploit opportunities; thereby enabling us to run in a manner which is, simultaneously, responsive, resilient, evolutionarily economical and elegantly efficient.

4.7 In summary

1. Running stiffness is not a unidimensional vertically oriented challenge, but a complex coordinative solution serving to simultaneously manage, remediate and exploit the 3-dimensional perturbations imposed during ground contact
2. Running stiffness is the close calibration of the tensional integrity of the running body (and its component segments), to the anticipated challenges likely to be imposed during the impending foot-strike
3. Thus, the practiced regulation of running stiffness mitigates exposure to excess impact shock

Chapter 5:

Smoothness: An Unexplored Window into Coordinated Running Proficiency

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5.1 Introduction

Smoothness is a widely recognised feature of healthy, proficient movement. Nevertheless, although the term ‘smoothness’ is commonly used to describe skilled athletic movement within practical sporting contexts, it is rarely specifically defined; rarely quantified, and remains barely explored experimentally. Elsewhere, however, within various health-related and neuro-physiological domains, many manifestations of movement smoothness have been extensively investigated. Within this literature smoothness is considered a reflection of a healthy CNS and is implicitly associated with practiced coordinated proficiency; ‘non-smooth’ movement, in contrast, is considered a consequence of pathological, un-practiced, or otherwise inhibited motor control.

5.1.1 The progression and regression of movement smoothness

Across a diversity of literatures, and within practical coaching contexts, smoothness is generally recognized as a universal feature of skilled motor behavior (Zehr, Barss, Dragert et al, 2016). Smoothness increases progressively as we transition from infant, to developing child, to mature adult; and regresses as we move from adult maturity into old age (Einspieler, Peharz, and Marschik, 2016; Ketcham, Seidler, Van Gemmert, and Stelmach, 2002; Traynor, Galea, and Pierrynowski, 2012). Furthermore, smoothness—whether assessed in gross movements or fine motor skills—improves, in logarithmic fashion, in parallel with the number of practice trials performed (Hreljac, 2000; Bartolo, De Nunzio, Sebastiano, et al, 2014). This effect is such that practice-driven improvements are reflected in increased smoothness in movement tasks as diverse as walking (Bartolo et al, 2014); writing (Bisio, Pedullà, Bonzano, et al, 2017; rock climbing (Seifert, Orth, Boulanger, et al, 2014); driving a golf ball (Choi, Kim, Shin, et al, 2015); piano playing (Caramiaux, Bevilacqua, Wanderley, et al, 2018); wheelchair propulsion (Jayaraman, Beck, and Sosnoff, 2015); dancing (Bronner and Shippen, 2015); over-arm throwing (Yan, Hinrichs, Payne, and Thomas, 2000), and in the hand dexterity of surgeons (Ghasemloonia, Maddahi, Zareinia, et al, 2017).

Many dimensions of declining function are, conversely, reflected in the deterioration of movement smoothness. Most obviously, smoothness is compromised following neurological damage, such as stroke, and subsequent recovery is typified by the gradual restoration of smoother movement (Bartolo et al, 2014). This effect is such that even simple measures of smoothness—evaluated in sit-to-stand tests, for example—can distinguish between older adults at risk of falls, older adults who are not a falls risk, and younger adults (Pozaic, Lindemann, Grebe, and Stork, 2016; Dixon, Stirling, Xu, et al, 2018). Similarly, smoothness during lifting movements—assessed at hip and ankle—declines with advancing age (Sakata, Kogure, Hosoda, et al, 2010); and many disease states, such as Parkinson’s and Huntington's, are accompanied by deteriorating smoothness (Smith, Brandt, and Shadmehr, 2010). Furthermore, in children, developmental disorders such as Autism and Asperger's are typified by a lack of movement smoothness (Nayate, Bradshaw, and Rinehart, 2005), and smoothness measures accurately detect delayed motor skill acquisition (Sander, de Schipper, Brons et al, 2017). Additionally, smoothness measures can discern between those who have previously suffered cervical injury, and non-previously injured controls (Ali and Seáñez-González, 2017); between those feigning whiplash injury and sincere patients (Baydal-Bertomeu, Page, Belda-Lois et al 2011); and between wheelchair users with, or without, shoulder pain (Jayaraman, Beck, and Sosnoff, 2015). Smoothness assessments are also sufficiently sensitive to detect decrements in highly learned skills caused by, for example, the influence of distractions on the driving performance of experienced taxi drivers (Kim, Choi, Choi et al, 2013), and movement skill inhibition following emotional disturbances (Baddoura and Venture, 2014).

Smoothness, consequently, seems both intuitively and empirically recognized as a hallmark of skilled, coordinated movement (Gulde and Hermsdörfer, 2018). Nevertheless, in relation to sporting movements in general, and running specifically, although the term 'smoothness' is commonly used to describe performers movement ability, it is rarely defined; rarely empirically quantified; is barely explored academically and is typically not directly targeted in training. In short, running smoothness is a phenomenon that we instinctively 'feel' we recognise when watching elite performance. Yet beyond this intuitive recognition, exactly what smoothness is remains surprisingly vague. In attempting to enhance our appreciation of this potentially important, yet largely ignored phenomenon; here, the general evidence relating to movement smoothness is presented, before subsequent reflection on how these insights may contribute to a more robust understanding of running coordination.

5.2 Is movement smoothness important?

Within a number of academic literatures smoothness is acknowledged as a fundamental characteristic of goal-directed human movement (Gulde and Hermsdörfer, 2018; Balasubramanian, Melendez-Calderon, Roby-Brami, and Burdet, 2015). Although not well investigated within sporting contexts, preliminary evidence suggests smoothness measures are capable of discerning between different levels of expertise. The club-head trajectories of skilled golfers, for example, are smoother than those of unskilled golfers (Choi, Kim, Mun, et al, 2010). Recent research, furthermore, established that a lack of smoothness—in the postural sway adjustments of NCAA Division 1 College football players—, predicted the likelihood of subsequent injury (Wilkerson, Gupta, and Colston, 2018). Such findings suggest smoothness is a phenomenon reflecting both practice-related skill improvements, and the underpinning functional health of the neuro-muscular system. Specifically, in relation to running, empirical insights remain sparse. The limited existing evidence, however, suggests competitive runners run more smoothly than recreational runners and, interestingly, also move more smoothly during fast walking (Hjerlac, 2004).

5.2.1 Defining smoothness

Given the apparent relevance of smoothness to movement proficiency, current definitions remain surprisingly vague. Within the neuroscientific literature, smoothness has been previously described as any movement that is not ‘‘jerky’’ (Hogan and Sternad, 2007), and a recent definition suggests that a movement is perceived to be smooth, when it happens in a continual fashion without any interruptions (Balasubramanian et al, 2015). We can, however, for our purposes here, sensibly broaden this definition by proposing that smooth movements are those without abrupt, intermittent, discontinuous changes in accelerations, decelerations, relative joint positions and/or movement trajectories. Although movements may occur rapidly, unexpectedly, even violently, there is a sense of consistent flow, of finely regulated progression, of seamlessly continuous coordinative control. Key dimensions of movement—postural control; relative joint positions; the absorption of impacts—, all appear to rhythmically, incrementally and predictably rise and fall. Visually, accordingly, we register a sense of fluency as the athlete dynamically progresses through a given movement sequence. Non-smooth movements, in contrast, leave an impression of abruptness, erratic discordance and of disjointed, unpredictable control.

Most commonly, in the relevant literatures, smoothness has been discussed in the context of movement kinematics and, less frequently, in relation to force profiles. Multiple measures are typically used (one recent review suggests 8) to assess smoothness (Balasubramanian et al, 2015); most commonly, however, smoothness is quantified by measuring its opposite, kinematic jerk. Jerk is formally defined as the rate of change in acceleration, i.e. the time-derivative of acceleration (Flash and Hogan, 1985; Hogan and Sternad, 2009). The smoothest movements consequently have, by definition, the lowest jerk (Choi, Joo, Oh, and Mun, 2014).

5.2.2 Why is smoothness a universal feature of human movement?

Several theories of motor control hypothesise that the brain coordinates muscle activation patterns to minimise a single, task-relevant *cost function.* Historically, it was assumed that the most heavily prioritized cost function, shaping movement control, was energetic expenditure (Kistemaker, Wong, and Gribble, 2010). Recent investigations, however, clearly demonstrate that although energy conservation is unquestionably a consideration, it is neither the only, nor necessarily the dominant, cost function shaping motor behaviours (Kistemaker, Wong, and Gribble, 2010 and 2014). Modelling predictions, for example, illustrate that 'impulsive running'—running with infinitely stiff, straight legs and zero sweep angle—, minimizes the mechanical cost of transport (Srinivasan and Ruina, 2006). Nevertheless, we run with energetically costly, compliant legs in a manner deviating substantially from this hypothetical optimum (Daley and Usherwood, 2010). In fact, energy conservation appears to be only one of a growing list of proposed constraints, each capable of adequately predicting the common kinematics of human movement. Such considerations include, for example, preservation of stability, reduction in the neural 'effort' expended in controlling movement, the minimisation of changes in torque, the minimisation of discomfort, and the regulation of movement accuracy (Harris and Wolpert, 1998); Todorov and Jordan, 2002; Balasubramanian et al, 2015). Thus, although practiced movements are typically executed in a manner that reduces energy costs, energy expenditure is not the exclusively over-riding priority, and is certainly not minimised. Although, experimentally, it seems impossible to determine which cost function is most heavily prioritized by the brain; notably, models prioritising smoothness consistently produce high-performing predictions (Srinivasan and Ruina, 2006; Kistemaker et al, 2014; Balasubramanian et al, 2015).

Nevertheless, although various rationales have been proposed within the relevant literatures, the reasons why smoothness is such a fundamental feature of healthy movement remain unclear. Previous research suggested that smoothness, during ground contact events, is an indirect consequence of the CNS's preference to employ single activation signaling bursts to individual muscles (Bobbert and Casius, 2011). The authors speculated that this strategy enabled adequate outcomes, while greatly simplifying neural control complexity. Furthermore, the authors noted they could see no reason why smooth movements offered advantages over non-smooth ones. Their proposal, instead, was that smoothness evolves naturally from the interplay between a single-stimulation-burst-per-muscle activation pattern, the linear behavior of the leg spring, and the innate viscoelastic and geometric properties of the musculoskeletal system. In essence suggesting smoothness, in landing tasks, emerges as a by-product of an evolutionary preference for simplified neural control, rather than because smoothness, in and of itself, offers any additional benefits (Bobbert and Casius, 2011).

More recent work, however, has proposed that smooth movements are inherently more predictable than less smooth, more erratic ones (Schwartz, 2016). Enhanced prediction of likely upcoming demands is beneficial as it permits a more fine-grained alignment between forecasted demands, advance preparation to meet these demands, and actually imposed demands (Schwartz, 2016). Enhanced predictive accuracy, accordingly, facilitates a more precisely attuned —more timely, and more finely calibrated— preparation for impending challenge. Accordingly, it's suggested that smoothness, as it promotes predictability, minimises movement error (Schwartz, 2016; Buma, van Kordelaar, Raemaekers, et al, 2016; Salmond, Davidson, and Charles, 2016). Similarly, more sensitive detection of subtle deviations from predicted trajectories facilitates more sensitive remedial adjustments; thereby offsetting the need for periodic, larger, more disruptive and energetically-demanding corrective interventions (Kiely and Collins, 2016).

Non-smooth (by definition, more jerky) movements, in contrast, are inherently less predictable. This diminished predictability inevitably detracts from the accurate forecasting of the likely kinetic and kinematic consequences of upcoming ground contacts. Any loss of calibration between anticipated and actually imposed demands inevitably leads to larger deviations from expected trajectories; thereby requiring more drastic remedial interventions to ‘correct’ unwanted deviations (Wolpert and Ghahramani, 2000). Larger corrective interventions necessitate larger motor commands, which generate, as a natural by-product, more signal-dependent neural noise; thereby further diminishing movement proficiency (Wolpert and Ghahramani, 2000).

In summary: more precise predictability facilitates a finer calibration between current preparation for soon-to-be-imposed demands, and the likely extent of those challenges. Smooth movements, as they require smaller on-line course corrections, minimize the disruptive effects of signal-dependent noise emerging as a natural consequence of larger motor commands (Schwartz, 2016). Smoothness, accordingly, by facilitating improved prediction, minimizes the necessity of persistent remedial correction and thus serves to simultaneously (a) reduce the neuronal computational burden associated with complex movement, (b) reduces energetic expenditure and (c) minimizes exposure to the damaging consequences of elevated jerk (Balasubramanian et al, 2015; Schwartz, 2016; Buma et al, 2016). Consequently, in a mutually re-enforcing manner, smoothness enhances prediction and prediction enhances smoothness.

5.3. The foundations of movement smoothness

Locomotion is initiated by commands originating in the motor cortex (Kiely and Collins, 2016). These descending commands are mediated and modulated by control centres in mid-brain and brain stem; before subsequently activating spinally-located central pattern generating (CPG) networks responsible for controlling the rhythmic synchronisation of the arms and legs; thereby delegating much of the coordination burden to lower, less evolutionarily expensive, neural control centres (Kiely and Collins, 2016). As rhythmic locomotion progresses, streams of sensory feedback return to spinal centres and guide the on-going customization of CPG outputs to current contexts, and triggering stabilisation reflexes (Kiely and Collins, 2016). In this way, sensory feedback directly modulates on-going feedforward activation to the extent that both become irrevocably entwined in a mutually modulating sensorimotor loop (Wolpert and Ghahramani, 2000).

Inevitably, however, neural and reflex-activating feedback and feedforward loops take time and cannot instantaneously respond to imposed perturbation. Proficient execution of impact-dependent movements—walking, running, jumping—

thus requires that the earliest remedial compensations, upon ground contact, are mediated by the practiced manipulation of the intrinsic material and structural properties of biological tissue collectives (Turvey and Fonseca, 2014). When skilfully deployed, accordingly, the innate viscoelastic and geometrical properties of the running leg provide an instantaneous, non-neurological, yet skilled, response to impact perturbations; for little energetic and neurological investment (Kiely and Collins, 2016). Thus, informed by anticipation and actioned by feedforward instruction, the time-lag deficits implicit in top-down neurally-mediated motor control are offset by the skilled manipulation of biological tissue properties (Biewener and Daley, 2007). Rhythmic locomotion, accordingly, is regulated by the blended output of three distinct, but mutually and irrevocably entangled, levels of control:

1. Top-down, supra-spinal executive direction
2. Spinally-located CPGs and stabilisation reflexes
3. The bottom-up, self-stabilising capacities afforded by the innate perturbation-resilient characteristics of bio-composite tissue structures

When operating effectively, feedback and feedforward information is blended with the plastically-embedded legacy of prior experience, to facilitate the skilled deployment of robust, task-conditioned, bio-composite tissue capacities. The fusion of these multi-level control systems underpins the runner's ability to sensitively detect and respond to upcoming perturbations in ways that minimally disrupt rhythmical locomotion. Smoothness thus emerges as a natural outcome of this intimate integration (Zehr and Duysens, 2004; Zehr, 2005).

5.3.1 The drivers of deteriorating smoothness

The gradual degradation of movement smoothness is driven by the inevitable accumulation of the experience-dependent *wear and tear* associated with natural aging, declining health, and injury and illness, and is compounded by cycles of overuse, underuse, misuse and disuse (Laczko, Scheidt, Simo, and Piovesan, 2017; Kiely, 2017). Although the exact mechanisms underpinning this progressive deterioration remain unclear, two broad inter-related neuro-motor deficits have been implicated:

i. As the plastically embedded legacies of past cycles of injury, misuse, disuse and overuse accumulate within the CNS, the micro-structures underpinning neuronal connectivity progressively degrade (Kiely, 2017). Consequently, sensorimotor communication clarity erodes, and the interpretation of sensory feedback and the precision of feedforward activation gradually decay
ii. Declining muscular strength—driven by neural signaling decrements, decreasing muscle mass, and the degradation of tissue micro-structures—necessitates that, to adequately execute a task requiring a given movement force, weaker muscles require more relative activation than stronger muscles (Reid, Pasha, Doros, et al, 2014). Inevitably, greater relative voluntary activation results in increasing signal dependent noise; thereby resulting in more disorderly motor unit recruitment and more erratically variable force outputs

The progressive erosion of neural and biological capacities thus ensures that coordination inevitably deteriorates as the legacy of past insults relentlessly accumulate. As multiple aspects of sensorimotor control—sensory awareness, activation accuracy, and the load-management capacity of biological tissues—diminish, smoothness inevitably declines. Tissues, consequently, become ever more exposed to the damaging consequences of exacerbated kinematic jerk.

5.3.2 Sensitivity of smoothness to changing conditions

Most obviously, fatigue and aging diminish muscular force generating capacities, and thereby increase the likelihood of negative outcomes. Fatigue and age-related decline also inhibit coordinative capacity; thereby diminishing motor control precision and compromising the capacity to safely manage the mechanical shock associated with ground contact (Shmuelof, Krakauer, and Mazzoni, 2012; Kline and Williams, 2015).

Although investigations into the relationships between sports-related injury and movement smoothness remain sparse; nevertheless, prior injury has been observed to erode the proprioceptive capacities of elite performers, for example, in runners and ballet dancers (Switlick, Kernozek, and Meardon, 2015; Steinberg, Adams, Tirosh, et al, 2018). Any reduction in the fine-grained anticipation of upcoming loadings, inevitably impedes the sensitive calibration of running stiffness to the demands of the upcoming impact challenge, and can be expected to increase the magnitude of unexpected deviations from projected trajectories; subsequently suggesting that diminished proprioception diminishes, whereas enhanced proprioception enhances, movement smoothness (Iwańska, Karczewska, Madej, and Urbanik, 2015; Bellenger, Arnold, Buckley, et al, 2018). Smoother movements thus protect tissues from excessive loadings (Goetschius, Kuenze, and Hart, 2015). This perspective is supported by evidence illustrating that smoothness improves as an outcome of effective rehabilitation (Iwańska et al, 2015; Riva, Bianchi, Rocca, and Mamo, 2016; Bellenger et al, 2018).

5.4 Future (practical and research) directions?

Although smoothness is clearly an under-explored dimension of skilled athletic movement. We can, however, draw some speculative, but logical, initial conclusions:

1. Smoothness is modified by a range of factors, including:
    a. Underlying health status (including neuro-physiological, psycho-emotional and disease status)
    b. Training and injury history
    c. Current fatigue and/or psycho-emotional states

2. Smoothness progresses and regresses as a function of normal maturation and aging, injury and subsequent recovery; and declining smoothness exposes tissues to exacerbated mechanical stress
3. Smoothness is a product of proficient coordination, mediated by the CNS, and actioned via the skilled deployment of the innate perturbation-resilient capacities of robust biological tissues

More specifically, in relation to running activities, it seems sensible to suggest that smoothness is potentially promoted by three broad categories of training intervention:

1. Any running-related challenges promoting enhanced calibration between feedforward activation and feedback sensory information
2. Interventions promoting the sensitive activation of (long and short latency) stabilization reflexes
3. Any loading strategies upgrading the structural and material resilience of biological tissues habitually subjected to mechanical stress during running activities

Finally, the sparse existing evidence hints that while practice improves, excessively repetitive practice leads to deteriorating neural communications; and declining movement smoothness (Kiely, 2017). Accordingly, more volume is not necessarily better. Instead, as with other facets of training management, improving running smoothness likely requires the sensitive regulation of volumes, intensities, exercise variation, and the judicious balancing of work and recovery.

5.5 Conclusion

Smoothness is a product of the collaborative triangulation between accurately interpreted sensory feedback and sensitively adjusted feedforward activation, contextualized against plastically embedded prior learning. As physical capacities and movement experiences accumulate, we innately gravitate towards smoother movement solutions as we learn to more sensitively respond to small perturbations; thereby offsetting the need to periodically and 'jerkily' respond to the larger challenges that would emerge if minor errors were allowed to accumulate. Smoothness thus reflects sensorimotor coordination and provides a quantifiable window into movement proficiency (Hogan and Sternad, 2009).

The rapid evolution of wearable micro-technology provides us with opportunities to accurately, and non-invasively, evaluate running smoothness. Currently, however, although evidence strongly suggests smoothness metrics provide insights into coordination proficiency and can be used as markers of neuro-rehabilitation effectiveness; critically, the most appropriate means to measure, monitor and analyse smoothness remain unclear (Balasubramanian et al, 2015; Gulde and Hermsdörfer, 2018). Thus, although preliminary evidence demonstrates the informational value of smoothness assessments, such measures exist only on the periphery of our sporting cultural consciousness, and remain poorly articulated, poorly understood and poorly explored.

As suggested in this chapter, an evidence-led logic supports the potential worth of objective smoothness evaluations and, currently, there is ready access to technologies enabling such evaluations. While, unquestionably, much remains to be clarified, and further research is necessary, the background and rationale outlined here serves as a useful conceptual starting point from where to begin this exploration.

5.6 In summary

1. Smoothness is a universal feature of healthy skilled movement which, although infrequently considered and currently under-appreciated within sporting contexts, may provide a unique window into athletic coordinative proficiency
2. Existing evidence illustrates that smoothness changes as a consequence of natural aging, general health status, practice and injury history, and current fatigue and injury status. Preliminary research suggests that running proficiency is reflected in smoother running movement
3. Recent advances in contemporary technology provides the opportunity to sensitively detect changes in running smoothness and may, subsequently, potentially bestow unique insights into running coordination proficiency

Smoothness appears a highly prioritised coordinative output, and a fundamental facet of proficient movement. Existing evidence suggests that both injury and fatigue erode movement smoothness, potentially by compromising the coherent integration of feedback and feedforward sensorimotor information.

Smoothness is most commonly assessed via quantification of its opposite, kinematic jerk. Although running smoothness remains barely explored, the evidence reviewed suggests smoothness may provide an insightful lens through which to evaluate running coordination proficiency.

5.7 Synopsis of the Literature Review Chapters

A critical theme, emerging from chapters 2 through 5, is that the many undercurrents which flow together to create a runner's coordination capacity, all appear, at an individual level, highly malleable to personal traits, morphological idiosyncrasies, historical events and current conditions. Accordingly, and returning to a point emphasised throughout these chapters, although we all share a common neurobiological and morphological design template, our personal coordination habits seem likely to be customised by a constellation of individually specific shaping events, manifesting in different proclivities and compensations, operating across multiple overlapping timescales.

A further, perhaps obvious, but worth re-iterating, observation is that fatigue induces multi-faceted coordinative change. Chapters 2 and 3 discussed the neural origins of these modulations and explain how such changes initially emerge as progressively diminishing neuro-biological complexity, which subsequently drives multi-level compensations; which inevitably influences how the running task is actioned and executed. The effects of these systemic changes, inevitably, lead to modifications in emergent coordinated management strategies designed to handle the mechanical stress imposed during ground contact.

The evidence reviewed in Chapters 4 and 5 strongly suggest that the key events likely to induce perceptions of 'threat' −whether injury, pain sensitisation and/or fatigue-induced dysfunction−, occur during contact between the running foot, and the running surface. Chapters 4 and 5 also discussed some of the downstream mechanical consequences of these neuro-biological disruptions, highlighting how fatigue-induced coordinative change is influences the modulation of a number of ground contact-dependent parameters. In fact, a clear theme permeating the review chapters is that multiple dimensions of the running action, and multiple outcomes of running coordination, appear modulated by fatigue. The main implication of this rationalisation is to underline the suggestion that evaluating how coordination changes, in response to fatigue, may provide an informative window into the nature of the running coordination phenomenon.

Accordingly, although we can describe coordination in terms of changes in multiple related variables, we cannot necessarily make a definitive value judgement on whether any given coordination pattern is appropriate for any given individual. In essence, within a population of healthy and regular runners, we can’t directly evaluate ‘better or worse’ running coordination habits; or at least, given the extent of inter-individual modulating influences, it seems difficult to precisely judge the appropriateness of one individual’s coordination habits against a generalised coordination template. We can, however, via the application of fatigue, potentially perturb habitual patterns and subsequently quantify the nature, direction and magnitude of subsequent change in critical coordinative outputs.

Chapter 6:

Methods and Methodological Decision-Making

6.1 Introduction

Running coordination is the study of the changing relationships between individual sub-components of the all-encompassing running system. The ‘outputs’ of these coordinating component sub-systems may take the form of any measurable variables whose relationships modulate, in an associated manner, over time. In human running contexts, multiple kinetic, kinematic and energetic variables continually change across micro- (within a single stride cycle) and macro-timescales (over the course of a run, a season, or a lifetime). This multiplicity of dynamically changing relationships entails that it is feasible to examine the running coordination phenomenon through a diversity of investigative lens, in isolation or combination, and via the use of a wide variety of technologies and measures (Bianchi, Angelini, Orani, and Lacquaniti,1998). As example, segmental joint angles, measured using 2- or 3-dimensional kinematics, continually change relative to each other during running.

Accordingly, their changing relationships provide a valid means of investigating coordination (Chapman, Vicenzino, Blanch, et al, 2008). Similarly, the changing recruitment patterns of specific muscle groups, measured using EMG (Cappellini, Ivanenko, Poppele, et al, 2006), provide a valid, albeit very different window into the running coordination phenomenon (Li, van den Bogert, Caldwell, et al, 1999). Subsequently, there are many potentially viable and interesting ways of empirically describing running coordination. Nevertheless, it currently remains unclear which set of quantifiable coordinating variables provides the most prescient insights into running 'health' and/or fatigue status.

The objective of this chapter is to consider the key themes emerging in the previous chapters, and to consider these as conceptual guides informing the experimental decision-making process. Subsequently, in the second portion of this chapter, the methods and protocol selections, judged to best reflect the stated thesis objectives, are described.

6.2 Methodological decision-making

The purpose of this section is to weigh core experimental considerations against the thesis objectives. In effect, the prime directive of this methodological decision-making process is to align both the insights and empirical deficits apparent in the literature, with the potential opportunities presented by contemporary technologies, and subsequently to evolve experimental protocols capable of illuminating the central questions directing this thesis.

6.2.1 Considerations guiding experimental design

Although many dimensions of running movement can be readily quantified, the evidence and logic contained in the preceding chapters strongly suggest that the most critical mechanical events, over the course of the stride cycle, arise consequent to the sudden, dramatic changes in acceleration which occur subsequent to collision between the running foot, and the running surface. Many of these facets of running coordination, however, remain unamenable to practical assessment without access to cumbersome and/or expensive laboratory equipment, such as instrumented treadmills, force platforms, and/or multi-camera motion capture systems.

The past decade of technological innovation has driven an explosion in the development of progressively smaller, yet more technologically dense, sensors capable of accurately measuring kinematic information. Specifically, these advances have enabled the manufacture of small, lightweight inertial measurement units (IMUs) capable of providing information-rich data, without impeding the runners comfort and without compromising coordinated running patterns. Such IMUs typically contain accelerometers, goniometers and magnetometers, with some also having the capability to capture muscle activity, via EMG.

In particular the recent evolution of small, low mass, wireless accelerometers offers the potential to investigate the changing nature of acceleration related metrics, without interfering with running mechanics (McMaster, Gill, Cronin, and McGuigan, 2013; Norris et al, 2014).

6.2.1.1 The potential practical utility of low-mass accelerometers

Accelerometers provide a means of directly assessing the accelerations experienced at the specific location to which that measurement unit is attached. Importantly, other relevant metrics, discussed within the preceding chapters, which can also be directly calculated from acquired acceleration data. Metrics, such as, for example, impact shock, kinematic jerk and movement smoothness, system and/or leg stiffness, and the timing of key ground contact events.

The potential utility of sensitive, low-mass accelerometry to running applications has not gone unnoticed (Healy et al, 2015). Subsequently, over the course of recent years, a number of differently focussed studies have appeared in the peer-reviewed literature. Accelerometer sensors have, for example, been used to explore CoM excursions in a cohort of regular runners during treadmill running (Clermont et al, 2019); detect deviations in centre of mass motion caused by treadmill running-induced fatigue (Schütte et al, 2015); estimate stride frequency during a very short overland run (Healy et al, 2015), and to evaluate the effects of static stretching on uphill running performance (Lowery et al, 2014). Whilst the majority of previous studies employed uniaxial accelerometers, the number of investigations using triaxial sensors is increasing (Schütte et al, 2015).

Within the recent literature, the key recommendations advocated when considering the use of accelerometers, in running based research, are that the sensors are:

a) Low mass, ideally weighing less than 3 grams
b) Have wireless data capture capability
c) Are triaxial, thereby having the capacity to record simultaneous streams of acceleration data for 3 orthogonally aligned axes

(adapted from Sheerin et al, 2018)

Understandably, however, given the rapid and recent evolution of this technology, many aspects of accelerometer use remain controversial, and current best practice remains unclear. Two relevant questions, which have not been definitively answered, relate to firstly, how best to attach units to anatomical locations; and secondly, to which anatomical locations accelerometers should be attached to best deliver the most pertinent data relevant to ambulatory applications.

6.2.1.2 Reliability and sensitivity of accelerometer assessments

As previously noted, there is an evident lack of published studies assessing the reliability of triaxial local acceleration measurements during running activities (Besier et al, 2018). Nevertheless, until recently, very few investigations had attempted to establish the reliability of acceleration measures assessed using isoinertial sensors. However, a handful of contemporary investigations have begun to address this deficit.

6.2.1.2.1 Between session reliability

Sheerin and colleagues (2018), in the first running study to report reliability and variability for triaxial accelerometers housed within commercially available isoinertial units, placed sensors on the tibias and the L4-L5 region of the lumbar spines of 14 regular, healthy runners. The participants subsequently ran, on a treadmill, at a moderate pace (13.5 km/h), until exhaustion. The same runners were subsequently re-tested, under the same conditions, at 1 week and again at 6 months. Mean resultant peak tibial acceleration values ranged from 7.8 to 12.0g for the baseline to one-week comparison, and 8.6 to 12.9 g for the baseline to six-month comparison. The mean differences in resultant peak tibial acceleration, between baseline and one-week sessions, ranged from 0.0 to 0.2 g (0–3.5%), and from 0.0 to 0.5 g (0–5.3%) between baseline and six-month sessions. Notably, no measures of resultant peak tibial acceleration exceeded an ES of 0.14. Subsequently, resultant peak tibial accelerations demonstrated 'good' to 'moderate' reliability, and 'small' to 'moderate' measurement variability, at all speeds, with baseline comparisons at one week and 6 months timepoints. These results were interpreted as indicating that triaxial accelerometers, attached to the tibias, can be reliably employed to assess running-imposed impact shocks across distinct data collection contexts (Sheerin et al, 2018).

6.2.1.2.2 Within-session reliability

Until recently no studies had investigated the within-session reliability of local acceleration metrics obtained from research-grade, commercially available, integrated measurement units (IMUs). In the first study attempting to establish the within-session reliability of acceleration measures obtained from research-grade triaxial accelerometers, Van den Berghe and colleagues (2019) attached IMUs to the distal third of the anteromedial portion of the tibias of 13 healthy, regular runners. The runners subsequently ran 3 differently paced overground trials –at speeds of 2.55, 3.20 and 5.10 ms^{-1}– on a 20meter instrumented runway. Running efforts were interspersed with intervals for detachment and re-attachment of the system. A reasonable inter-trial reliability of vertical and resultant peak tibial accelerations was confirmed for the within-session measurements. No statistically significant difference in peak tibial accelerations was detected between sessions. Notably, however, the repeatability of resultant peak tibial accelerations was superior to vertical peak tibial accelerations. The investigators acknowledged that reliability may be improved if the running interval was extended, thereby adding more ground contacts to the data collection. Nevertheless, the investigators did conclude that contemporary IMU technologies provide reliably repeatable within-session acceleration measurements across a variety of over-ground running paces (Van den Berghe et al, 2019).

In another recent study Aubol and colleagues (2020) fitted 19 healthy, regular runners with research-grade IMUs attached to the anteromedial aspect of the tibia, before running participants through 10 brief, multi-foot contact trials, at a 10.8 km/h pace, interspersed with short rest periods. Notably, these trials were run both overground and on a treadmill. Peak vertical and peak resultant tibial accelerations were determined for each trial. Subsequent analysis suggested that within-session reliability was excellent for both treadmill and overground conditions (intraclass correlation coefficients (ICCs) =.95–.99). Minimal detectable differences (MDDs) were also calculated and ranged from 0.6 to 1.4 g for peak axial acceleration and from 1.6 to 2.0 g for peak resultant acceleration. This study appears to be the first to report the within-session reliability of peak axial or peak resultant tibial accelerations, measured specifically during treadmill running. The authors concluded that, when using research-grade IMUs mounted on the distal tibia during treadmill running, investigative protocols are likely to reliably quantify peak vertical and peak resultant tibial accelerations.

Yet another recent, and novel, treadmill running study compared the reliabilities of 2 IMUs –specifically comparing a research-orientated system, the Delsys Trigno IM (Delsys Inc, Natick, Massachusetts, USA) and an IMU designed for clinical use, the ViPerform (Dorsavi, Melbourne, Australia). Sensors were fitted to the anteromedial portion of the tibias. Runners ran multiple 60 second trials at 2 paces, 14 kph and 18 kph. The peak tibial acceleration metric demonstrated good to excellent

ICCs (range = 0.83–0.91) for the ViPerform system and excellent (0.91–0.96) ICCs, with the exception of the non-preferred leg at 18kmph on Day 2 which had good reliability (ICC = 0.86), for the Delsys Trigno system. Similarly, the Delsys system provided consistently lower standard errors of measurement (SEM) and minimal detectable changes (MDC). SEMs ranged between 4.98 and 8.05% for ViPerform and 2.50–3.32% for the Delsys IMU. MDCs ranged between 13.80 and 22.32% for ViPerform and 6.92–9.21% for the Delsys IMU (Hughes et al, 2019). Subsequently, the investigators concluded that although both systems demonstrated a ‘good’ to ‘excellent’ intra-session reliability. The Delsys Trigno system demonstrated less measurement error and higher reliability. The authors suggested this enhanced precision was due to the higher sampling frequency of 148.1 Hertz (Hz), in comparison to the lower (100/20/20 Hz) frequency of the ViPerform units (Hughes et al, 2019).

These recent investigations illustrate the reliability of acceleration data obtained from contemporary research-grade IMUs, specifically fixed to the distal portion of the tibiae. The reliability of local acceleration data obtained from IMUs located on thoracic and/or sacral sites, during treadmill running, however, remains unexplored.

6.2.1.3 Selecting an attachment method

Previous work, focussing on the assessment of bone strain, attached accelerometers directly to the tibial bone matrix. These methods typically used Hoffman pins to secure the sensor directly to the bony surface, and were subsequently highly invasive (Nigg, Cole and Brüggemann, 1995). Although the relationships between bone strain and tibial accelerations are still unclear, this method has produced reasonable correlations with key force related parameters associated with ground contact, for example vertical impact peaks (r=0.7-0.85) and loading rates (r=0.87-0.99) (Hennig and Lafortune, 1991). However, given the invasiveness of this method, and the very plausible consideration that subsequent discomfort likely leads to modifications to running coordination patterns, this method is clearly impractical for regular runners.

Thus, more commonly, accelerometers have been mounted on the skin covering the anteriomedial aspect of the tibia. This method has yielded weaker correlations in relation to GRF parameters; nevertheless, subsequently derived average vertical loading rate (r=0.274 to 0.439) and instantaneous loading rate (r=0.469) have been shown to be significantly correlated with peak tibial accelerations (Greenhalgh, Sinclair, Protheroe, and Chockalingam, 2012).

There are three points especially worth noting here. Firstly, previous studies, using direct attachments between sensors and the skin, have used various means of securing the sensors to different anatomical locations. Subsequently, as these studies did not typically report on the quality of attachment, it is not clear which investigations most securely fixed the sensors to the body. Secondly, even recent investigations have employed inertial measurement units of different sizes and masses, ranging in weight from as much as 25g (Clark et al, 2017) to less than 3g (Norris et al, 2014). Both of these factors seem likely to either increase or diminish subsequent measurement accuracy. In this context, it is worth noting that one previous study did examine the effects of different means of attaching units to the tibia, and established that when sensors were more securely attached to the bodily surface, subsequent results were judged to be more accurate (Forner-Cordero, Mateu-Arce, Forner-Cordero, Alcántara, Moreno, and Pons, 2008). However, currently, as recently noted, there is no consensus, or indeed reliable way to measure, what constitutes an adequately secure attachment method (Sheerin et al, 2017). Thirdly, previous investigations have used tibial acceleration as a proxy for bone loading, in the belief that the forces generated at ground contact are critical regulators of bony injury risk further up the kinematic chain. As discussed in Chapter 4, however, this may not be a sustainable presumption.

Currently, the weight of recent evidence suggests that tibial accelerations provide a more informative window into running-related injury risk, and possibly running performance, than force related variables. Crucially, however, it is worth re-emphasising an observation noted by Sheerin and colleagues (2018), who pointed out that despite previous research efforts, it is still not clear which measurable dimension of the ground contact event exerts the greatest influence on injury likelihood and/or running proficiency. Similarly, it is not clear whether various metrics touted as being indicative of fatigue produce similar estimations of fatigue status.

6.2.1.4 Attachment locations

Despite the growing popularity of wearable accelerometry, there is little agreement on which anatomical locations provide the most informative running-related data. Most frequently, in relation to running, accelerometers have been attached to either the distal or proximal portions of the anteromedial aspect of the tibia. Many of these studies have focussed primarily on knee kinematics, in relation to previous injuries or injury risk factors, and subsequently positioned sensors high on the tibia, proximal to the knee. In contrast, other investigations, designed to explore tibial fatigue fractures (which are more commonly located distally), have chosen to attach sensors to the distal portion of the shank. Unsurprisingly, the selected tibial attachment site influences the magnitudes of measured accelerations, and broadly suggests that distally derived accelerations are typically substantially higher than those measured closer to the proximal end of the tibia (Lafortune and Hennig, 1991; Lucas-Cuevas et al, 2017).

A smaller fraction of the relevant research attached uniaxial or triaxial accelerometer units on the low back, at a site approximating Centre of Mass (CoM), and provided evidence that such placement provides a meaningful reflection of the kinematic challenges imposed on the whole-body during running activities (Le Bris et al, 2006; MacGregor et al, 2009). Recently, triaxial accelerometers have been placed on the low backs of patients recovering from stroke, to assess the extent of stroke-induced gait asymmetries (Zhang, Smuck, Legault, Ith, Muaremi, and Aminian, 2018). More specifically to running, recent studies have positioned tri-axial accelerometers on the low backs of non-athletic humans to determine shock attenuation during walking and running, and to assess changes to CoM motion driven by running-induced fatigue (Schütte et al, 2015; Provot, Chiementin, Bolaers, and Munera, 2019).

A much smaller sub-set of the running-related research used accelerometer units attached higher on the torso, or to the head (for example: Edwards et al, 2019). As noted earlier, in Chapter 2, the capacity to dampen the accelerative shockwaves, emanating from ground contact, before those shockwaves reach the brain, is a critical coordinative imperative. Supporting this conjecture, prior investigations have noted a significant dampening of accelerations, measured via head-mounted accelerometers, prior to those accelerations reaching the brain (Gruber et al, 2014). However, such studies necessitated the mounting of accelerometers on the forehead, typically requiring the manufacture of specialist strapping equipment that, as noted in these studies, was a discomforting limitation to the runner; thereby potentially moderating habitual running patterns (Gruber at al, 2014). More pragmatically, a small number of recent studies have attached sensors to the bony prominence of the 7th cervical vertebrae (C7). These studies have concluded that this location provides valuable insights into dimensions of performance, such as injury risk, stability and kinematic variability in walking (Zoffoli, Ditroilo, Federici, and Lucertini, 2017), dance (Brogden, Armstrong, Page, Milner, Norris, and Greig, 2018), and sub-maximal running activities (De Brabandere, De Beéck, Schütte, Meert, Vanwanseele, and Davis, 2018).

6.2.1.5 The influence of fatigue on running coordination

A persistent theme, emerging from the evidence reviewed in Chapters 2 through 5, is that running coordination evolves to resolve multiple simultaneously applied, and overlapping, movement problems. Solving these complex running-imposed coordination tasks demands the proficient deployment of innate, perturbation-resilient structural and tissue properties, actioned via the entwined collaboration of multiple sensorimotor processing modules dispersed throughout the central nervous system. Over-time, running coordination gradually improves in parallel with normal maturation and accumulating running practice, as multiple neural and tissue collectives become more accustomed and better equipped to cope with the challenges posed during running activity. Conversely, running coordination deteriorates in parallel with any decline in sensorimotor processing capacity and/or any detrimental erosion of tissue properties. Thus, inevitably, coordination seems likely to decline in tandem with aging, accumulating injury and increasing fatigue.

Most notably, as discussed within the preceding chapters, fatigue induces change across multiple mechanical coordinated outcomes, such as, locally experienced acceleration shocks, movement smoothness and the modulation of shock attenuation via the manipulation of system stiffness. However, it should be noted that 'how much', or in 'which direction,' change occurs remains unclear and controversial across all these metrics. Similarly, it is worth noting that a common problem with investigations claiming to have induced fatigue, is that fatigue is ultimately a subjective experience that does not necessarily lend itself to direct quantification and/or standardisation across protocols. Nevertheless, regardless of these uncertainties, perhaps the single most consistent finding emanating from the literature review chapters is that fatiguing activity drives multi-faceted change, and that this change is evident across multiple measurable running-related parameters.

6.2.2 Assessing running coordination: Resolving challenges and opportunities

As noted at the start of this chapter, selecting a variable that reflects some dimension of running coordination is not necessarily a challenge; the challenge instead is selecting variables that provide optimally pertinent and insightful information. To this end, the recent evolution of low mass, triaxial accelerometery provides a flexible and adaptive assessment tool capable of, non-invasively, collecting very large datasets pertinent to accelerations and accelerative change. Furthermore, within the literature reviewed in the preceding chapters, sufficient causal links have been hypothesised between acceleration and/or acceleration-derived metrics and performance, injury risk, injury recovery, general health factors (such as disease symptoms and normal aging) and running-related fatigue, to suggest that further investigation is warranted. Furthermore, as noted by Norris and colleagues (2016), the majority of previous running-related studies have relied on information provided by uniaxial accelerometers. Consequently, the use of triaxial sensors potentially provides an opportunity to provide new and novel insights into the running coordination phenomenon.

Subsequently, within the investigations contained in this thesis, 4 lightweight triaxial accelerometer sensors were attached to 4 anatomical locations, each of which have previously been used in running investigations, specifically: the proximal portion of left and right tibias, the sacrum, at a position approximating CoM, and the thoracic spine at the bony prominence of C7.

Subsequently 4 metrics, each discussed in literature review chapters 4 and 5, and each separately advocated as being sensitive to fatigue-driven declines in coordination proficiency, were selected as the core experimental metrics. Specifically, these 4 measures were:

- Local accelerations
- Local jerk
- Shock attenuation assessed between tibia and pelvis, and tibia and thorax
- Resultant accelerations

Notably, each of these metrics, although measuring different dimensions of the ground contact event, are directly calculated from acceleration timeseries. Subsequently, reflecting the pragmatic objectives of this thesis, and the decision to focus on the set of acceleration and acceleration-derived metrics, the specific anatomical locations selected for accelerometer attachments were:

- The distal third of the anteromedial aspect of left and right tibias, with the axis of the accelerometer aligned with the longitudinal axis of the tibia, following the protocol of Lucas-Cuevas and colleagues (2016)
- On the low back at a site approximating Centre of Mass (CoM), with the axis of the accelerometer aligned vertically, as per the protocols of Le Bris and colleagues (2006), and MacGregor and colleagues (2009)

- On the bony prominence of the 7th cervical vertebrae, with the axis of the accelerometer aligned vertically, for example, Edwards and colleagues (2019).

While it has been suggested that any misalignment may result in measurement errors, these errors have previously been estimated to be trivial, accounting for only 1-2% of total peak acceleration magnitude (Derrick et al, 1998). Furthermore, although accepting that some error is inevitable, providing measurement errors were reliably consistent, in terms of magnitude and direction, then these inaccuracies seemed unlikely to detract from the core experimental objectives. In other words, if sensors remain on the runner, rather than being detached and then reattached, the likelihood of errors is reduced.

6.2.2.1 The utility of fatigue as an experimental perturbation

As noted at the start of this chapter, coordination can be objectively assessed using a potential multitude of measures. Nevertheless, while it's possible to describe how specific variables co-modulate during running activity, determining what constitutes 'good' coordination remains problematic. In short: there are many potential investigative lenses through which running coordination can be evaluated, but it remains unclear which lens provides the most relevant and practically amenable insights.

The twin premises underpinning this experimental design are that: firstly, accumulating fatigue drives coordinative changes and, secondly, that changes to coordination are reflected in a changing ability to manage different facets of the mechanical challenge imposed during ground contact. This approach specifically investigates the potential erosion of the coordinated regulation of the mechanical stressors that present the most direct ‘threats’ to both running proficiency and tissue health. Subsequently, focussing directly on the ‘outputs’ of running coordination, and examining how these outputs may change over the course of a fatiguing run, potentially provides a window into how running coordination may, or may not, deteriorate subsequent to running-induced fatigue. Such an approach enables comparison, at both a whole cohort and individual level, between what could be considered ‘better’ coordination, i.e., coordination while fresh, and ‘worse’ coordination, i.e. coordination that is compromised by fatigued conditions. Accordingly, as running fatigue accumulates, deteriorations to running coordination may be reflected and detected by monitoring changes in these kinematic parameters.

6.2.2.2 The fatigue quantification problem

A central theme permeating the literature review chapters is that fatigue induces change across multiple dimensions and facets of running coordination. From this perspective fatigue potentially provides a lens through which to investigate the broader nature of running coordination.

Fatigue, however, is a complex, multidimensional, task-dependent phenomenon conventionally divided into central and peripheral components (Fernandez, Firdous, Jehangir et al, 2020). Central fatigue is perceived as the diminishing capability of the central nervous system (CNS) to generate and transmit appropriate activation signals to the musculature during exercise. Peripheral fatigue, in contrast, refers to a diminishing ability to generate appropriate muscular tensions at, or distal to, the neuromuscular junction driven by excessively intense or prolonged physical activity (Fernandez et al, 2020).

As noted in Chapter 2, when running increases in severity and/or duration we are made consciously aware of mounting 'threat' through increasingly discomforting interpretations of emerging sensory information (Marcora et al, 2009; Smirmaul, 2012). At the whole-body level effortful physical activity drives changes to cognitive, psycho-emotional and physiological states (Kuppuswamy, 2017). These pervasive changes drive growing perceptions of anxiety, physical discomfort and diminished attention, which in turn intensify the inner conflict between motivational drive and perceived exertion –what we, collectively, interpret as mounting 'fatigue' (Marcora et al, 2009; Smirmaul, 2012; Seay et al, 2011).

6.2.2.3 Defining Fatigue

The multi-faceted nature of fatigue ensures it is a nebulous and variously defined concept (DeLuca, 2005). Definitions of fatigue differ across academic domains and across specific sporting contexts. One recent and notable definition, specifically within sports science contexts, provided by Enoka and Duchateau (2016), describes fatigue as a disabling psychophysiological symptom underpinned by the interactions between (i) performance fatigability—the decline in an objective measure of performance over a discrete period of time and (ii) perceived fatigability—changes in sensory experience driven by the performers subjective interpretation of the imposed challenge. This definition captures how performance fatigability, assessed via objectively quantified decrements in some discrete measurable facet of performance, and perceived fatigability, assessed via subjective evaluation of the performers perception of their capacity to cope with imposed challenges, are linked through mutually interactive and mutually modulating non-linear feedback and feedforward communication loops (Enoka and Duchateau, 2016). The multi-faceted nature of fatigue and the varied inputs that collectively shape the fatigue experience, combine to ensure that direct quantification of fatigue is problematic (Kuppuswamy, 2017; Fontes et al, 2020). Crewe, Tucker and Noakes, 2008; St Clair Gibson et al, 2003).

6.2.2.4 Perceived exertion

In his seminal work, in the early 1980’s, Gunner Borg described perceived exertion as a Gestalt phenomenon arising from the integration of informational content flowing from both peripheral and central nervous systems, and the environment. Notably, Borgs primary objective was to create a global perceptual scale of exertion. Subsequently, he did not differentiate between exertion, effort, exhaustion and fatigue, nor between psycho-emotional or physical dimensions of exertion. Borg believed that subjective ratings of perceived exertion were the single best indicators of activity-driven physical strain and, in order to evaluate perceived exertion created a 20-point Category Ratio (CR-20) scale, later followed by the CR-10 and CR-100 scales (Borg, 1980, 1982, 1985 and 1998). Borg scales for rating perceived exertion have subsequently been extensively employed for estimating effort, discomfort, and fatigue during exercise and/or exercise testing and provide simple alternatives to more invasive, and technology-dependent, markers of physiological stress, such as blood lactate (HLa) concentration, oxygen consumption and heart rate (HR) (American College of Sports Medicine, 2010; Crewe, tucker et al, 2008; Fusco et al, 2020).

Although ratings of perceived exertion are frequently used as surrogate measures of fatigue Borg specifically noted that while fatigue is related to perceived exertion, both phenomena are distinct, writing that "the concept of fatigue should be distinguished from the concept of perceived exertion even though these two concepts have very much in common" (Borg, 1986, pp. 3-12). Reflecting this conceptual overlap, definitions of perceived exertion frequently include reference to fatigue. As illustration, Noble and Robertson (1996), in an influential text on the topic, defined perceived exertion as the subjective intensity of effort, strain, discomfort and/or fatigue that is felt during exercise. This relationship is such that higher ratings of perceived exertion (RPE) are associated with higher physiological stress and fatigue. Subsequently, numerous works demonstrate that physiological adjustments –such as increases in heart rate, ventilation, oxygen consumption and metabolic acidosis, for example—, exhibited by endurance athletes under fatiguing exercise conditions, are strongly reflected in increasing ratings of perceived exertion (Crewe, Tucker and Noakes, 2008; Fusco et al, 2020; Halperin and Emanuel, 2020).

Accordingly, although fatigue is clearly a complex and multidimensional phenomenon, and although there is a clear definitional distinction between fatigue and PE, ratings of PE are considered to offer imperfect, yet reflective and readily amenable, surrogate indicators of fatigue (Ten Haaf, van Staveren, Oudenhoven et al, 2017; Fusco et al, 2020). Importantly, despite the inaccuracies of assuming PE provides a direct measure of fatigue, CR10 scales are commonly used to estimate the perceptual effects of fatiguing activity during running investigations (for example: Arney, Glover, Fusco, et al, 2019).

Subsequently, and in full acknowledgement that there are inherent limitations in such an approach, multiple distinct definitions of fatigue and multiple different perspectives on whether PE is an appropriate surrogate for fatigue, within the context of these investigations the Borg Category-Ratio-10 scale (CR-10) was selected as the primary means of assessing the participants response to fatiguing running activity (Borg, 1998). The CR-10 (See Appendix A) was also used to help communicate the desired sense of effort and discomfort deemed as signifying an acceptable fatigue threshold, using practical running training related terms and reference points. In the absence of definitive objective metrics this subjective, surrogate means of estimating fatigue status was chosen as the most relevant, most relatable, and importantly most familiar assessment method for this cohort of recreational runners.

Notably, as mentioned in Chapter 1, although it seems clear that running coordination deteriorates under conditions of severe acute and/or chronic fatigue. Less clear, however, is the extent to which running coordination may change in response to regularly conducted, moderately fatiguing, training runs, such as those commonly encountered within recreational and competitive running training programmes. Accordingly, in the context of the investigation conducted within this thesis, participants were requested to continue running until reaching a CR-10 rating of 8.5. This exertion rating broadly equates to a moderately hard to hard training run, and follows the example of prior running investigations which successfully observed biomechanical adjustments, over comparable running durations, using similar target ratings of exertion (for example: Brown, Zifchock and Hillstrom, 2014).

6.2.2.5 Gathering relevant information

Previously, in 2011, a study by Professor Hamill's group at the University of Massachusetts at Amherst (USA), used a uni-axial accelerometer fixed to a single shin to quantify the local tibial accelerations experienced over the course of a sustained run (Meardon et al 2011). In searching for a suitable questionnaire capable of adequately capturing relevant training and injury data, the lead investigator of this study was contacted, and a copy of the questionnaire was requested. Given that the experimental questions, study designs and participant profiles were broadly similar in both this published study, and these planned investigations, this questionnaire seemed highly applicable to this research project. In essence, this questionnaire interrogated the 3 main topics of key relevance, namely: performance levels, training history and injury history.

6.2.3 Familiarisation with the experimental protocol

A series of single subject pilot studies were conducted, under University ethics, prior to final consolidation of the experimental procedure. Over the course of these pilots three assessment technologies were trialled: Qualisys^{TM} Motion capture system, Optogait^{TM} gait analysis system, and Delsys Trigno^{TM} accelerometer sensors. Following these trials, the Delsys Trigno^{TM} system was chosen as the most appropriate assessment technology; primarily based on the efficiency of data collection, the broad applicability of the derived data, and the capacity of the Trigno system to capture direct, unfiltered acceleration data.

These pilot studies were designed to trial, and subsequently, refine individual elements of the intended protocol, and also to consolidate timings, and organisational and logistical considerations. A series of single subject pilot trials also provided the opportunity to practice the location and mark up of anatomical locations, and to consolidate the securing and wrapping of sensors to these anatomical locations.

Subsequent pilots helped familiarise to the different elements of the experimental protocol and to trial, and subsequently, refine individual elements of the intended protocol, including the location and mark up of anatomical locations, and provided practice opportunities to consolidate the securing and wrapping of sensors to the anatomical locations. And the distinct steps necessary within the Delsys software to prepare, record and subsequently export the experimental data. These practice opportunities ensured that timings and logistics (the organisation of forms, writing materials, changing facilities, and so on), were previously practiced prior to the start of the first experimental data collection.

6.2.3.1 In summary

A number of implications, emerging from Chapters 2 through 5, served as conceptual guides and constraints influencing experimental design. Key amongst these considerations were:

1. The multi-layered evidence, and rationale, suggesting that running coordination proclivities are likely to be highly personalised and to vary extensively, on an inter-individual basis, across multiple levels of observation
2. Although the evidence base remains incomplete, conflicting and frequently vague, existing evidence does suggest that the critical events most likely to drive injury risk and impede running performance occur during contact between the foot and the running surface
3. Increasing fatigue drives changes to the magnitudes of acceleration-derived metrics experienced during the ground contact event

Reflecting the pragmatic objectives described in Chapter 1, the experimental approach designed for this thesis reflected the following key ambitions:

- To utilise low-mass acceleration sensor technology to provide novel, and practically translatable, insights capable of illuminating the running coordination phenomenon
- To design a protocol that was simultaneously conceptually robust; yet remained sufficiently 'light-weight' as to be readily replicated
- To test the perspective, emerging from the conceptual review chapters, that fatigue moderates multiple measurable running coordination outputs

In the subsequent section the experimental methods designed to action these various considerations and ambitions, within the context of a practically implementable experimental protocol, are described.

6.3 Methods

This section outlines the protocols and procedures designed and adhered to relating to the recruitment of appropriate participants for the empirical investigation, the technologies employed, and the subsequent content, organisation and delivery of the experimental protocol.

6.3.1 Ethical approval

The studies included in this thesis were approved by the Science, Technology, Engineering, Medicine and Health (STEMH) ethics committee, at the University of Central Lancashire (See Appendix B). Prior to data collection, all participants provided written informed consent (See Appendix C).

6.3.2 Participants

6.3.2.1 Participant Recruitment

Participants were recruited (a) via direct contact with the Press Officers of various running and triathlon clubs within the greater Preston area, who then circulated the relevant introduction and Participant Information Sheets (PIS) to any interested club members and (b) via a research advertisement circulated via social media (see Appendix D). Interested potential participants were advised to contact the researcher via email or phone.

6.3.2.2 Participant profiles

Potential participants were informed they should be regular runners and/or field sports players; be experienced with running on a motorised treadmill; have no current injuries or disorders and have no medical conditions which contra-indicated effortful physical activity and/or would potentially expose participants to undue risk through physical exertion.

Participants were also informed they must be over the age of 18, with a history of > 3 running sessions per week, for at least 40 weeks a year, for at least the past 12 months. Prior to beginning data collection, all participants were informed that they could volitionally terminate the test, without explanation, at any juncture. Fifteen participants who volunteered met the inclusion criteria. They were aged between 18 and 66, with a history of >3 running sessions per week, for at least 40 weeks of the preceding 12 months. The majority of participants were previous, or current, regional, national and/or international age-grade competitors in running, mountain running, duathlon and/or triathlon events. All participants were free from injury and any running-related discomfort at the time of data collection and were partaking in regular running training.

Two participants were female and 13 male. Two of the participants were left foot dominant and 13 were right foot dominant. All participants reported that they were experienced treadmill runners.

6.3.2.3 Sample size estimation

Previously published research established significant differences between injured and non-injured runners, assessed using a uni-axial accelerometer attached to a single shin, using a cohort of 9 runners in each group (Meardon et al 2011). A sample size calculation shows that 8 subjects are needed in each group to detect a difference between non-injured and injured groups with a mean difference of 0.17g with a standard deviation of 0.1g at the 5% significance level with 90% Power. However, the mean differences between fatigue states were significantly greater with a mean difference of 0.33g with a standard deviation of 0.1g; thereby requiring a smaller sample size. However, given that the studies within this thesis would be employing 4 triaxial accelerometers; thereby necessitating the capture and subsequent analysis of substantially more data, it was estimated that a cohort of 10 or more runners would be adequate to explore the experimental questions posed. Subsequently, and to allow for the potential for dropouts and missing data, a cohort of 15 participants were initially recruited.

6.3.3 Personal data collection

Upon arriving at the laboratory, the participants were greeted, and asked to re-read the PIS and sign the consent form. Next, all participants were asked to fill out the experimental questionnaire. Following completion of the questionnaire, participants were asked to change into their running kit. Leg dominance was also recorded for each participant. Leg dominance was established by the participant’s verbal response to the question “which foot would you use to kick a ball”; a means of determining leg dominance previously found to have 97.7% agreement with task performance, and a 96% test-retest agreement (Coren and Porac, 1978).

6.3.1.1 Questionnaire

In order to collate relevant background information on injury and training histories and current performance data, an adapted form of the questionnaire previously used by Meardon and colleagues (2011) was issued and completed by all participants (Appendix E). This questionnaire was completed in the experimental laboratory on the day of data collection, so that any subsequent queries, or necessary clarifications, required by the participants could be answered immediately and directly. Pertinent results are presented in Appendix F).

6.3.1.2 Anthropometric data

Prior to commencing warm-up, participants were weighed on a calibrated scale and had height and leg length assessed using the laboratory stadiometer and anatomical tape (See Appendix F).

6.3.4 Sensors and sensor placement

The Delsys Trigno™, Type 3, Wireless sensors (Delsys Inc., Boston, MA, USA), are capable of measuring tri-axial impact accelerations of up to 9g. All accelerations were expressed in gravitational units (g), with one gravitational unit = 9.81 m/s^2. During testing the system streamed data to EMGworks Acquisition where the data was recorded and stored for later export and processing in EMGworks Analysis (Delsys Inc., Boston, MA, USA). The Delsys Trigno Lab Wireless system was supported on a 64-bit Windows 10 laptop. Each Trigno sensor had a pre-existing factory calibration for the triaxial accelerometery. When data is collected in EMGworks, this calibration information is used to compute and display selected variables. As per the only previous directly comparable investigation which used the Trigno sensors in a running application (Whelan et al 2015), a manufacture set sampling frequency of 148Hz was used.

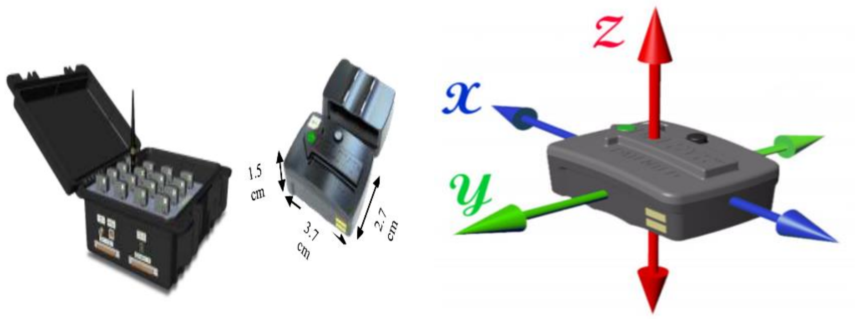


Figure 6.1: Delsys Trigno™, Type 3, Wireless sensor and charging unit

As much care as possible was taken to securely attach the accelerometer sensors to the designated locations. Sensors were first fixed to the skin using hypoallergenic adhesive and subsequently further secured by circumferentially wrapping an elasticated strap around the body segment. After each runner’s initial warm-up was completed, sensors were checked to ensure all remained secure, and the participant was questioned as to whether each sensor felt comfortably fixed in place. Any insecure sensors were subsequently reattached, and the runner again encouraged to warm-up on the treadmill, before once more reporting on both sensor security and comfort.

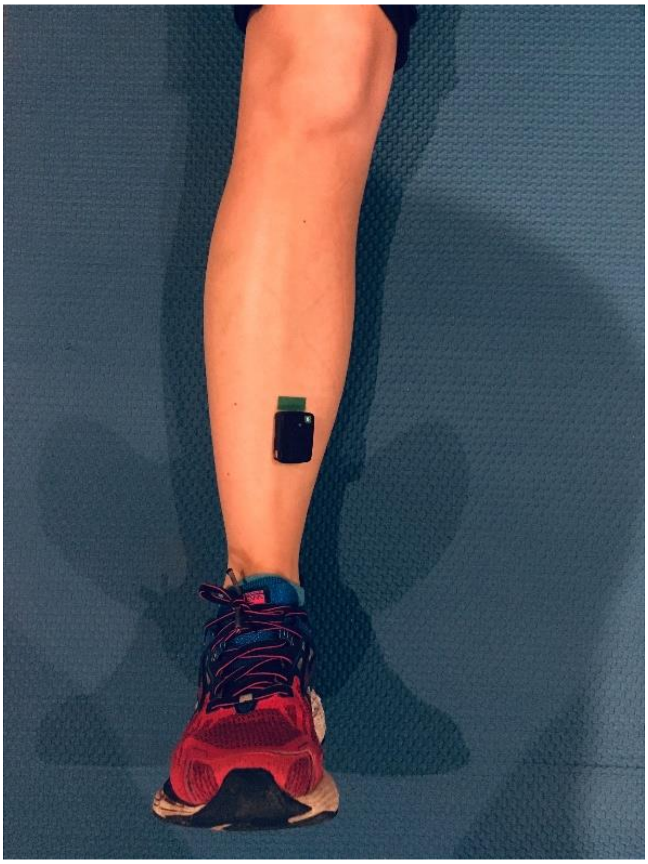


Figure 6.2: Tibial Mounted sensor before wrapping

Following height, weight and leg length measurements, and before each participant warmed-up 4, Type 3, Trigno low-mass sensors (Delsys Inc., Boston, MA), containing triaxial accelerometers, were secured to specific anatomical sites on the left and right tibia, sacrum and cervical vertebrae. Using anthropometric tape, the distance between the bony prominences of the anterior aspect of the tibial tuberosity and the medial malleolus was measured, and the position of the distal third of the tibia was subsequently calculated, measured and marked. The remaining attachment sites, on the bony prominences of the spinal processes at L2/L3 and C7, were identified via palpation and subsequently marked, as per previous investigations

(Rabuffetti, Scalera and Ferrarin, 2019). The sensors were positioned with the arrow pointing vertically upwards in parallel with the longitudinal axis of the tibia. In the case of the sensors attached to L2/L3 and C7, sensors were fixed on the spine with the arrow pointing upwards.

The sensors were initially attached to the skin using a hypoallergenic double-sided adhesive tape, approved for dermatological applications. Once fixed in this manner the sensors were firstly, further secured using medical adhesive tape applied in multiple directions and, secondly, additionally secured by wrapping an elasticated medical bandage around the sensor and body segment. This additional strapping was to ensure each sensor was securely stabilized against the bony prominences at each site; thereby minimising any relative movement between the sensors and the runner’s body. If the runner reported any undue discomfort, the strapping was removed and subsequently re-applied.

6.4 The running-specific intervention

6.4.1 Determining an appropriate ‘Run to Fatigue’ pace

Based upon personal best times and current training and running fitness status, a 'Run to fatigue' (RtF) pace (km/h) was pre-agreed with each participant. The RtF was gauged to be initially readily sustainable, but to result in quickly mounting perceptions of fatigue as the trial continued. The agreed pace was estimated to bring the participant to an RPE of 8.5 on the Borg CR-10 scale (1982) in between 8 to 18 minutes. Participants were asked to gauge this level of running-induced fatigue against the peak subjective rating of fatigue habitually encountered towards the end of a 'hard' run-training session.

6.4.2 Familiarisation process

Prior to beginning the running component of the experimental protocol, participants were asked to complete their personal pre-running warm-ups, and were allowed time to familiarize with the experimental treadmill. Once the participants indicated they felt ready to commence the running-specific element of the experimental protocol they were encouraged to rest for 2 minutes and take a drink or a toilet break if necessary.

6.4.3 Pre-Run to fatigue process

Following briefing and warm up, the participant ran, for 60 seconds at a pace that was 1 km per hour higher than the pre-agreed pace for the prolonged 'run to fatigue' (RTF). Firstly, to begin the Pre-RtF interval the participant initially walked, then jogged, then ran on the treadmill as the experimenter gradually increased the treadmill speed. When the treadmill reached the required the experimenter counted to 3 before starting the 60 second timer. Participants then ran for 60 seconds at this pace. At the end of this pre-RtF interval, participants rested for 2 minutes, before then commencing the RtF intervention.

6.4.4 Run to fatigue process

After 2 minutes rest the participant once more mounted the treadmill and the treadmill speed was again gradually increased until reaching the agreed RtF pace, i.e. a pace that was 1 km per hour slower than the pre- and post-running interval pace. Upon commencing the RtF, participants ran continuously at the agreed pace such that throughout the duration of the RtF the treadmill speed remained unchanged. When participants no longer wished to continue, they signalled volitional termination. More specifically, the participants signalled that they wished to stop running at the end of the next minute. Although this RtF protocol is novel, broadly similar means of imposing moderate fatigue have been employed in at least one previous treadmill running study (Koblbauer et al 2013).

As per previous studies, ratings of perceived exertion were obtained every 2 minutes by asking participants to indicate RPE scores (Brown, Zifchock, and Hillstrom, 2014), either verbally or by pointing to a Borg CR-10 chart (Borg 1998). Heart rate was also monitored and recorded every 2 minutes.

6.4.5 Post-Run to fatigue process

Upon cessation of the RtF participants rested for 2 minutes, before again mounting the treadmill to complete the second 60 second running interval, the post-RtF interval. The pace, used throughout this 60 seconds of running, was identical to the pace used in the first running interval, the pre-RtF interval.

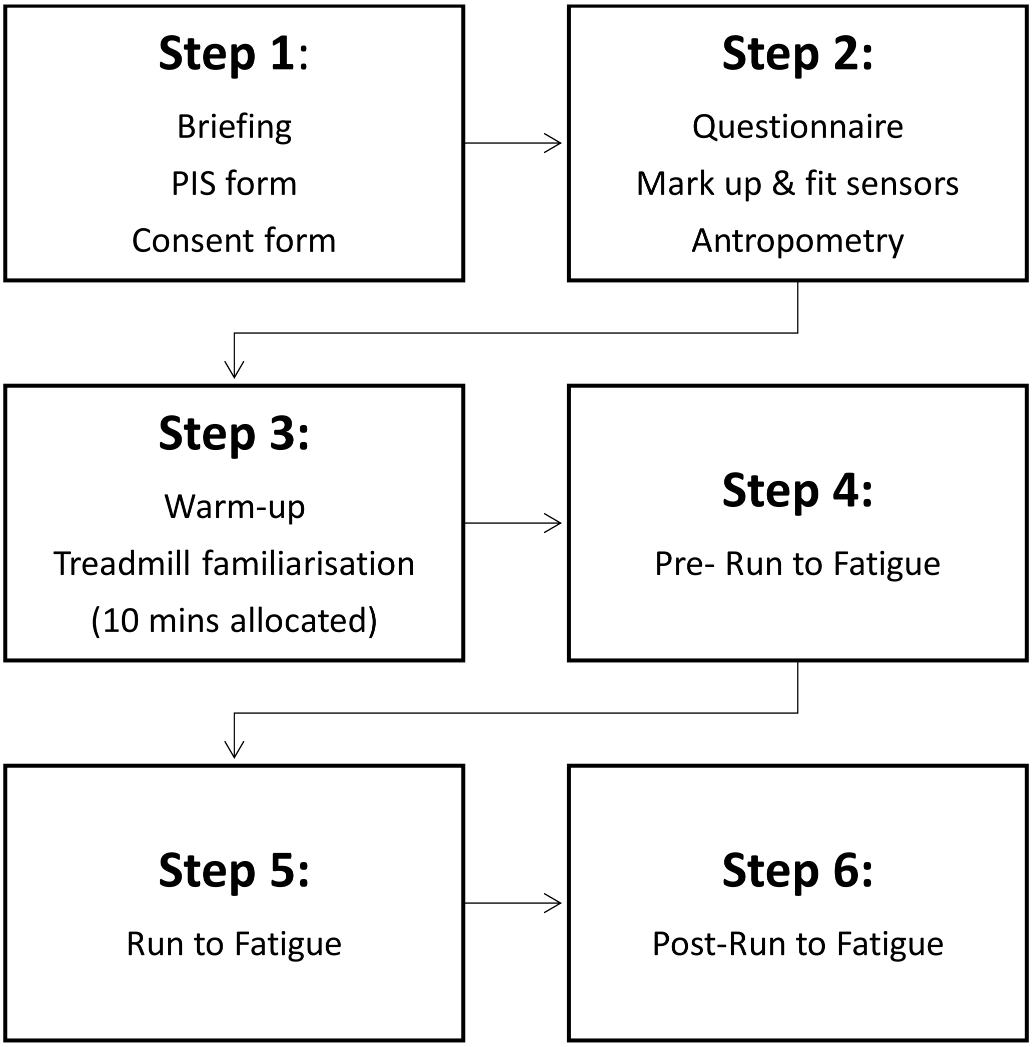


Figure 6.3: Experimental workflow

6.4.6 Results inclusion criteria

Criteria for results inclusion were, following the protocol of Koblbauer and colleagues (2013), that the participant's subjective rating of perceived exertion on termination was rated as 8.5, or above, on the CR-10 scale, and that the runners heart rate at termination was within 20% of their age-predicted maximum. This level of run-induced fatigue was agreed, with each individual participant, to be within the spectrum of training-induced fatigue habitually encountered during 'moderately hard' running training sessions. Consequently, adhering to these conditions did not demand any excessive or unusually strenuous effort from participants.

6.5 Data processing

Within EMGWorks Acquisition (Delsys Inc., Boston, MA), the data was recorded and stored for later export and processing in EMGworks Analysis (Delsys Inc., Boston, MA),. Following initial data treatment in *EMGworks* Analysis, all data was exported directly from the Delsys software to Visual 3D software (Version 4.87.0. C-Motion Inc., USA), a Microsoft Windows compatible data management tool designed specifically for biomechanical research.

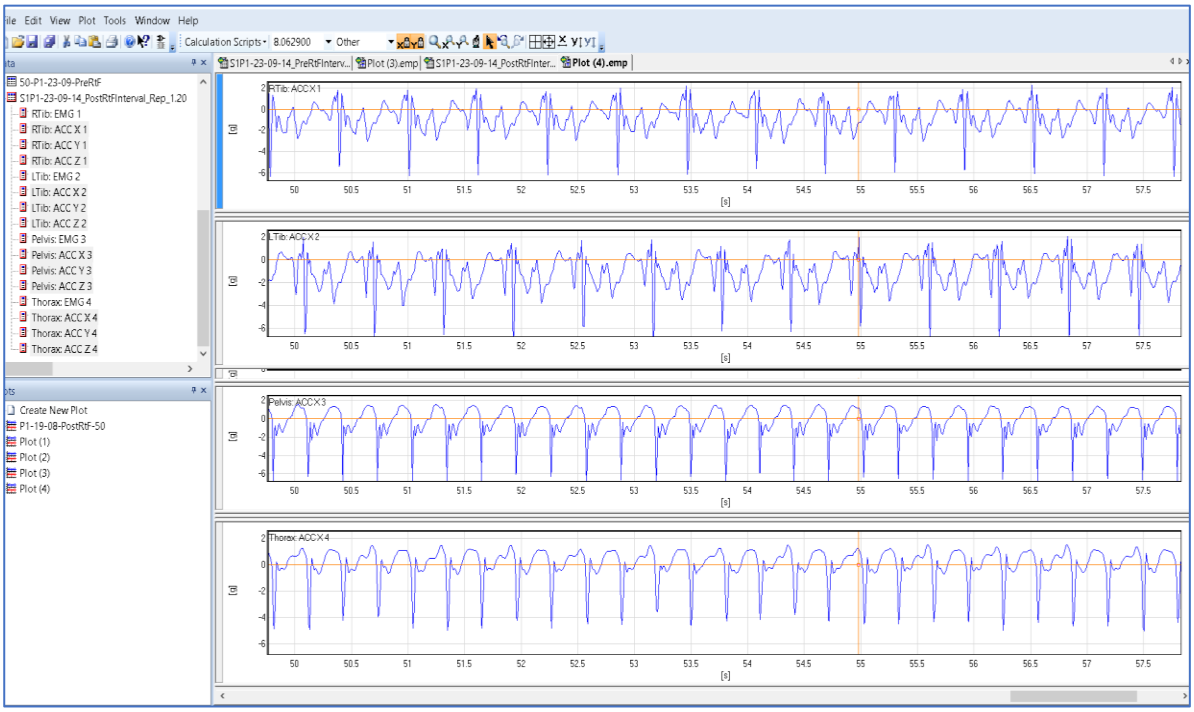


Figure 6.4: Sample acceleration data window in EMGworks Analysis

6.5.1 *Treating data in Visual 3D*

The acceleration peaks associated with each ground contact event were identified and marked within Visual 3D. The Cartesian coordinate system sign conventions, for the Delsys Trigno Type 3 sensor, are defined as: x-axis, superior-inferior; y-axis, mediolateral; z-axis, anteroposterior.

The ground contact event was marked using the mediolateral acceleration trace and was initiated from the base of the first medial deceleration of the ground contact event. The mediolateral acceleration trace was selected as the datum as it is the first of the triaxial acceleration traces to peak subsequent to ground contact. Using these criteria, 42 consecutive left and right strides were manually identified within the Visual 3D software. 42 strides were marked as 2 strides, one at each end of the timeseries, are lost during processing (see figure 6.5, for example).

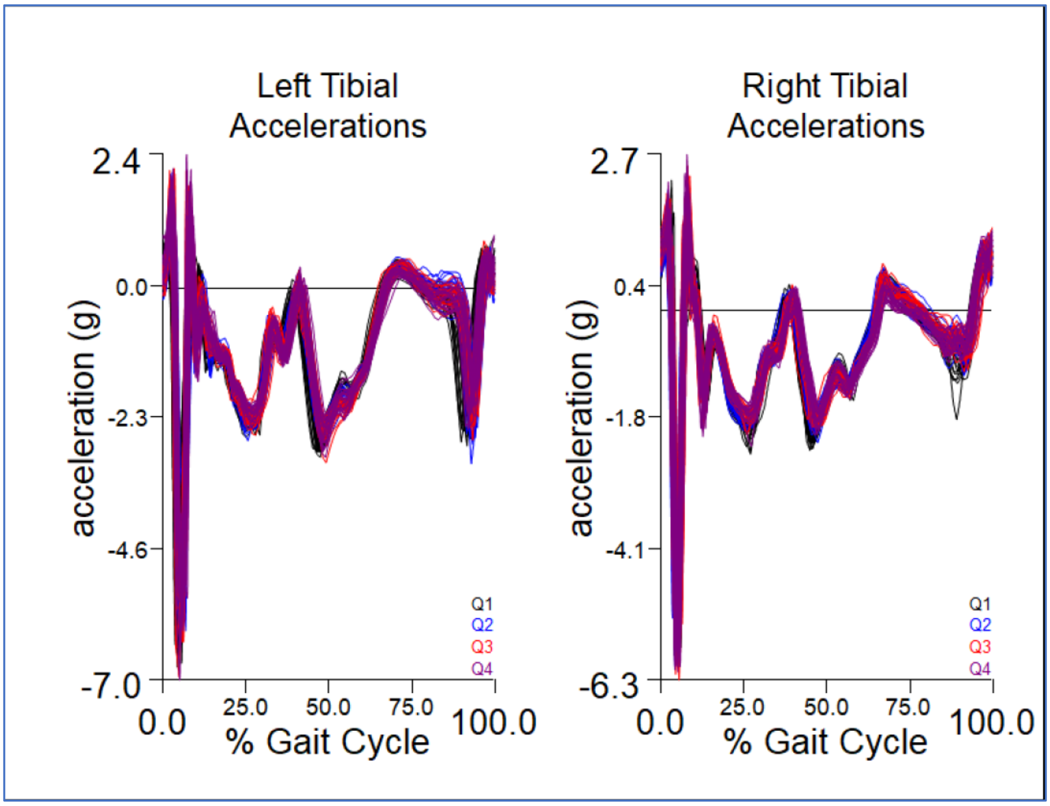


Figure 6.5: Sample tibial vertical acceleration time-series comprising superimposed left and right ground contacts

Once each individual acceleration peak had been identified, graphs for each timeseries along each axis were generated in Visual3D, thereby enabling quality control checks for processing and marking errors (see figure 6.6, for example).

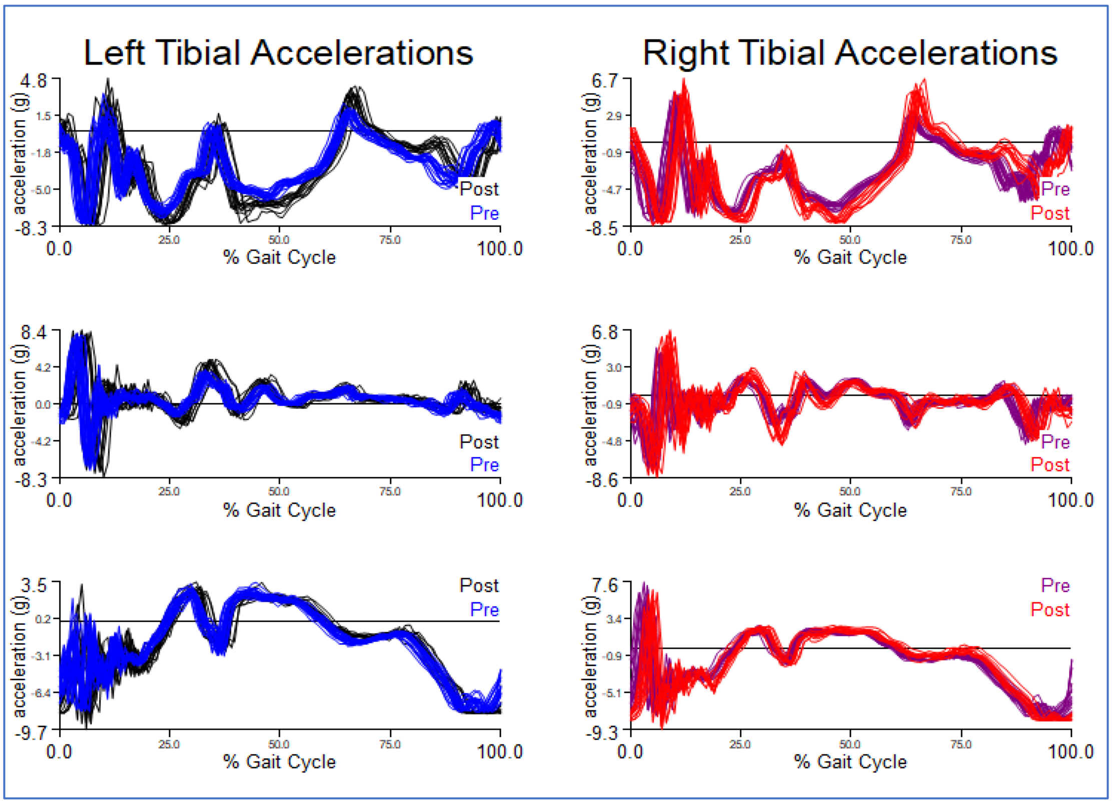


Figure 6.6: Left and right tibial time-series comprised of 20 overlaid consecutive strides

Once processing was completed, in Visual3D, the data timeseries was exported as an ASCII file, and subsequently imported into Excel^{TM}. Maximum and minimum values were extracted for each parameter, under each experimental condition. Subsequent statistical analysis was conducted in SPSS (20.0), excepting t tests for the quartile data of the fatiguing run, which were conducted in Excel^{TM}.

6.5.2 Data preparation

6.5.2.1 Pre- and post-fatigue interval runs

42 consecutive strides (20 for each leg) were marked within the time-series of each of the pre- and post-fatigue 60 second interval runs, subsequently exported to .csv and imported into SPSS for analysis. Once processing was completed, in Visual3D the data timeseries was exported as an ASCII file and subsequently imported into Excel^{TM}.

6.5.2.2 Continuous Run-to-Fatigue

Across the duration of the fatiguing run, the acceleration time-series was recorded for 30 seconds of every minute. Upon completion of the run, and after initial data treatment, each run was divided into quartiles. For each quartile, 42 consecutive left to right stride cycles recorded during the last minute of the quartile were marked and treated in Visual3D^{TM}, and subsequently exported as an ASCII file before being imported into Excel^{TM}.

6.5.3 Deriving experimental metrics

6.5.3.1 Local accelerations

9 acceleration metrics, for each sensor attachment location, were extracted from the acceleration timeseries in Excel. The specific acceleration metrics extracted were:

- Vertical positive (+) acceleration
- Vertical negative (-) acceleration
- Vertical range, extending between positive and negative vertical peaks
- Lateral acceleration
- Medial acceleration
- Mediolateral range, extending between medial and lateral acceleration peaks
- Anterior acceleration
- Posterior acceleration
- Anteroposterior range, extending between anterior and posterior acceleration peaks

6.5.3.2 Kinematic jerk

9 jerk metrics, for each sensor attachment location, were calculated within the Visual3D software directly from the acceleration timeseries using a specially customised

pipeline. Jerk was calculated as the rate of change of acceleration, i.e., the 1st differentiation of acceleration with respect to time.

The specific jerk metrics extracted were:

- Vertical positive (+) jerk
- Vertical negative (-) jerk
- Vertical range, extending between positive and negative vertical peaks
- Lateral jerk
- Medial jerk
- Mediolateral range, extending between medial and lateral jerk peaks
- Anterior jerk
- Posterior jerk
- Anteroposterior range, extending between anterior and posterior jerk peaks

6.5.3.3 Shock Attenuation

Shock attenuation was calculated, following the suggestions of both MacDermid and colleagues (2017) and Sinclair (2017), as the ratios between peak pelvic and peak tibial deceleration (SA – P) and peak thoracic and peak tibial deceleration (SA – T), using the formulas:

SA - P = 1 – (Peak Vertical Pelvic Shock/Peak Vertical Tibial Shock) *100

SA - T = 1 – (Peak Vertical Thoracic Shock/Peak Vertical Tibial Shock) *100

6.5.3.4 Resultant accelerations

The resultant accelerations experienced at tibial, pelvic and thoracic sites were calculated using the vertically, medially- and posteriorly-directed deceleration peaks experienced during ground contact, following the formula suggested by Sheerin and colleagues (2018):

$$\textbf{Resultant Acceleration (RA)} = \sqrt{(\text{Vertical deceleration peak})^2 + (\text{Medial deceleration peak})^2 + (\text{Posterior deceleration peak})^2}$$

All metrics were calculated for both pre- and post-fatigue intervals and for all the quartile data of the fatiguing run.

6.5.4 Test-retest reliability of the experimental protocol

To determine the test-retest reliability of the experimental procedures 3 participant runners were recruited under ethics adapted for COVID-19 precautions (see Appendix G). Each runner followed the experimental protocol, in terms of completing the questionnaire, warming up and running the 1st 60 second interval. After this pre-fatigue interval, however, the participant was instructed to rest for 10 minutes before running the 2nd 60 second interval. All data were collected, treated and analysed as per the experimental protocol. The means, for each of the selected investigative metrics, were calculated for both 60 second runs using the pre- and post-run acceleration time-series. Subsequently, the smallest worthwhile changes (SWCs) for each metric, at each location, were calculated.

Hopkins (2000) argued that an observed change equal in magnitude to the limits of agreement was excessively stringent, in most real-World cases. Hopkins suggests, instead, that 1.5 times the typical error (a little over half the limits of agreement) represents a realistic threshold for change as the corresponding odds of change are between 6 and 12 to 1 (Hopkins, 2000). SWCs were calculated following this recommendation. The specific SWCs, for each metric, were subsequently derived and applied to all experimental data. The derived SWCs, for all metrics, are available in Appendix H.

6.5.5 Analysis

Analysis, in the form of Repeated measures (RM) ANOVA and t tests were primarily conducted using SPSS for Windows (IBM-SPSS, Armonk, NY, USA, version 20.0), with some t tests also being conducted in Excel^{TM}. Alpha values were set at 0.05 for ANOVA and all subsequent post hoc tests. Effect sizes were classified as large, medium or small. The smallest worthwhile changes (SWCs) for each metric, at each location, were calculated from the acceleration time-series data collected in a test-retest reliability investigation. These SWCs were subsequently applied to all experimental data.

6.6 Conclusion

This chapter began by detailing the methodological decision-making underpinning key decisions relating to experimental design. Paramount amongst these considerations were the choice of technology; the consideration of best practice, as it exists, for both location and attachment of acceleration sensors; the utility of fatigue as an informative intervention capable of revealing, how and when, running coordination degrades during a fatiguing run.

The first part of this chapter discussed key considerations influencing experimental design decisions. Throughout the literature review chapters, a persistently emerging theme was that fatigue changes multiple facets of coordination, and multiple measurable coordination outputs. Accordingly, the first key experimental decision related to the selection of an assessment technology that could provide insightful data, but without unduly interfering with the runner's comfort and/or capacity to run in a natural, normal and unimpeded manner. Consequently, reflecting the pragmatic ambitions of this thesis, low mass accelerometer sensors were selected as the most applicable measurement technology. These sensors are capable of providing large quantities of pertinent data and enable the collection of time-series data without any reports of any modifications to the participants natural running style. Furthermore, the use of acceleration sensors enabled the collection of running data in a training context that was, albeit in a laboratory, familiar to all runners; a moderately hard, moderately fatiguing treadmill run. Based on literature review, 4 related, yet distinct, acceleration-derived metrics, each suggested within the literature as indicative of changes driven by running-related fatigue and dysfunction, were selected as the core measures for the experimental analysis. The second part of this chapter detailed the experimental protocol.

Chapter 7:

Evaluating changes to accelerations and acceleration-derived metrics following a fatiguing run

7.1 Introduction

Previous research, reviewed in chapters 2 through 5, suggest that the consequences of impaired coordination, driven by fatigue, diminish the ability to optimally manage shock loadings and, ultimately, increases the likelihood of running-related injuries. This increased risk of negative outcomes is propelled by a reduced capacity to appropriately absorb, dissipate, and distribute the shockwave of impact emanating from the ground contact event.

Within the relevant literature, 4 metrics (local accelerations, jerk, shock attenuation and resultant accelerations) are proposed as capable of differentiating between non-fatigued and fatigued running states. Although these 4 measures have each been advocated as sensitive indicators of fatigue-induced change, no prior work has applied these measures, simultaneously, to the same experimental dataset.

Subsequently, it remains unclear whether these measures typically change synchronously providing concurring, or change differentially providing conflicting, estimations of running-induced fatigue.

A further feature of the existing literature is the assumption that the placement of measuring devices, at different anatomical locations, provides comparable information and insights pertinent to running behaviours (for example: Nedergaard et al., 2017). These assumptions are reflected, in practical running contexts, by the range of commercial products capturing local movement measures at different anatomical locations, typically using different attachment methods, yet citing, as validation of their efficacy, analysis of data collected in other locations, often using different devices and different means of securing devices to the runners body (for example: Edwards et al, 2019).

Finally, the literature reviewed in chapters 2 through 5 suggests that individual runners, driven by personal injury and training histories, evolve deeply personalised strategies to mitigate the mechanical stressors associated with the repetitive impact shocks emanating from ground contact. However, whether this personal customisation is evident across the various acceleration-derived metrics used in this investigation remains unclear and is, in fact, obscured by conventional grouped analysis methodologies. Accordingly, whether inter-individual, runner-specific differences conform to a generalised template of change, or respond in widely disparate manners, remains under-investigated and unresolved.

A clearer understanding of whether, and how, key acceleration-derived measures may differ between individual runners, and whether fatigue influences accelerations and acceleration-derived metrics in a manner which is consistent or differs substantially across individuals, appears highly relevant to both the design and subsequent interpretation of future studies investigating fatigue-induced changes to coordinated running behaviours.

To examine these questions the analysis detailed in this chapter tracked acceleration time-series data across pre- and post-fatigue 60 second interval runs, separated by a continuous run-to-fatigue protocol. The Smallest Worthwhile Changes (SWCs), calculated as per the recommendations of Hopkins (2000) and presented in Chapter 6, were applied to all metrics.

7.1.1 Objectives

Accordingly, the specific objectives of this investigation were three-fold:

i. To determine whether acceleration-derived metrics, advocated in the literature as sensitive indicators of coordinated running behaviours, change subsequent to moderate running-induced fatigue
ii. To assess whether running-induced fatigue changes local accelerations and acceleration-derived metrics in a uniform manner across measures, and across attachment locations

iii. To examine whether fatigue-induced changes, to these measures, are experienced similarly across a cohort of runners or are experienced in an individually idiosyncratic, runner-specific manner.

7.2 Results: Group analysis of local accelerations and acceleration-derived metrics

The acceleration peaks experienced by each runner during ground contact, at each of the 4 anatomical locations and along each of the vertical, mediolateral and anteroposterior axes, were identified and extracted from the pre- and post-fatigue time-series of each runner.

Using this acceleration data, local jerk, shock attenuations and resultant accelerations, for each of the 4 anatomical attachment locations, and for each of 9 locational and directional-specific metrics, were calculated. These metrics captured peak positive and peak negative magnitudes, and the magnitude of the range experienced between positive and negative peaks. Pre- and post-fatigue datasets were compared using repeated measures analysis of variance. The Smallest Worthwhile Changes (SWCs), for each metric, as per the recommendations of Hopkins (2000) and presented in Chapter 6, were applied to all metrics. The results are presented below.

7.2.1 Changes to local accelerations

7.2.1.1 Changes to tibial accelerations

Analysis first checked for interaction effects. No interactions between either pre- and post-fatigue, or left and right, tibial accelerations, were detected. Subsequently, repeated measures ANOVAs were conducted to explore main effects of side. Where significant main effects were detected, post-hoc pairwise comparisons were conducted between tibias. Significant main effects were observed for the magnitude of peak vertical tibial decelerations, between pre- and post-fatigue conditions (F(1,14)=7.45, p=0.016 ES=0.35 – Large), and for the range of vertical tibial accelerations experienced between pre- and post-fatigue conditions (F(1,14)=12.07, p=0.004 ES=0.46 – Large). The magnitudes of vertical decelerations, experienced by left and right tibias, also differed significantly (F(1,14)=34.43, p=0.001 ES=0.71 – Large), with vertical tibial accelerations experienced by the left tibia being significantly higher than those experienced by the right tibia (See table 7.1).

The magnitudes of medially-directed decelerations, experienced by left and right tibias, increased significantly (F(1,14)=13.76, p=0.003 ES=0.49 – Large), as did the magnitudes of laterally-directed accelerations experienced by left and right tibias (F(1,14)=13.43, p=0.002 ES=0.49 – Large). Similarly, the magnitudes of posteriorly-directed decelerations experienced by left and right tibias, differed significantly (F(1,14)=11.01, p=0.005 ES=0.44 – Large). Results are presented in table 7.1.

The relevant metric-specific Smallest Worthwhile Changes (SWCs), as shown in Appendix H, were compared to pre- and post-fatigue and left and right tibial data. The difference between pre- and post-fatigue means (0.438 ms^{-2}) exceeded the SWC for vertical tibial deceleration (0.256 ms^{-2}). Similarly, the SWC for vertical tibial range (0.220 ms^{-2}) was less than the difference between pre- and post-fatigue vertical tibial range means (0.635 ms^{-2}). The SWC for vertical tibial deceleration (0.256 ms^{-2}) was smaller than the difference between the means of left and right tibial decelerations (0.919 ms^{-2}). The SWC for lateral acceleration (0.154 ms^{-2}) was less than the difference between left and right lateral tibial means (1.377 ms^{-2}), and the SWC for medial tibial accelerations (0.203 ms^{-2}) was less than the difference between left and right medial tibial means (1.342 ms^{-2}).

Table 7.1: Tibial Accelerations and main effects from RM ANOVA

Measure	Left tibia Pre-fatigue Mean (SD)	Left tibia Post-fatigue Mean (SD)	Right tibia Pre-fatigue Mean (SD)	Right tibia Post-fatigue Mean (SD)	Pre v Post p value (np^2)	Left v Right p value (np^2)	Interaction p value (np^2)
Vertical							
Vertical Accel	1.69 (0.80)	1.88 (1.12)	2.02 (0.99)	2.22 (1.65)	0.313 (0.07)	0.165 (0.13)	0.998 (0.01)
Vertical Decel	6.17 (1.40)	6.72 (0.85)	5.36 (1.30)	5.68 (1.27)	**0.016*(0.35)**	**0.001*(0.71)**	0.077 (0.21)
Vertical Range	7.85 (1.40)	8.60 (1.64)	7.38 (1.90)	7.90 (2.52)	**0.004*(0.46)**	0.103 (0.18)	0.313 (0.07)
Mediolateral							
Lateral Accel	2.90 (2.16)	2.93 (2.51)	4.29 (1.92)	4.23 (2.15)	0.586 (0.01)	**0.002*(0.49)**	0.483 (0.04)
Medial Decel	5.56 (3.13)	5.83 (2.80)	4.29 (2.77)	4.34 (2.63)	0.951 (0.01)	**0.003*(0.01)**	0.729 (0.01)
ML Range	8.46 (4.85)	8.76 (5.04)	8.58 (4.40)	8.57 (4.55)	0.720 (0.09)	0.953 (0.01)	0.720 (0.01)
Anteroposterior							
Anterior Accel	1.13 (1.58)	0.80 (1.50)	1.49 (1.52)	1.22 (1.84)	0.265 (0.09)	0.364 (0.06)	0.090 (0.19)
Posterior Decel	4.86 (1.45)	5.52 (1.49)	4.57 (1.23)	5.13 (1.42)	0.061 (0.23)	**0.005*(0.44)**	0.660 (0.01)
AP Range	5.99 (2.01)	6.32 (2.12)	6.06 (2.53)	6.35 (2.71)	0.137 (0.15)	0.901 (0.01)	0.920 (0.01)

**Denotes that the mean difference is significant at the .05 level*

Table 7.2: Post hoc comparisons using Least Significant Differences between Pre- and Post-fatigue conditions

Post hoc comparisons using Least Significant Differences between Pre- and Post-				
	Mean diff.	p value	Upper bound	Lower bound
Lateral Accel	1.377	0.002	2.17	0.58
Medial Decel	-0.84	0.004	1.01	-0.71
Posterior Decel	0.34	0.005	1.25	-0.03

Table 7.3: Post hoc comparisons using Least Significant Differences between Left and Right sides

Post hoc comparisons using Least Significant Differences between Left and Right Sides				
	Mean diff.	p value	Upper bound	Lower bound
Vertical Decel	0.92	p>.001	-0.58	-1.25
Vertical Range	-0.84	0.004	0.05	-1.15

7.2.1.2 Changes to pelvic accelerations

Following comparisons between pre- and post-fatigue pelvic accelerations, significant changes were detected in 2 pelvic acceleration measures. Specifically: Peak medial accelerations increased between pre- and post-fatigue conditions, *t*(15)=2.77 p

= 0.015, and the range of mediolateral accelerations extending between peak positive maximum to peak negative minimum also increased significantly between pre- and post-fatigue conditions *t*(15)=2.77 p=0.016 (see Table 7.4). Notably, the magnitudes of each of these changes also exceeded the magnitude of the calculated SWC metrics for both pelvic medial acceleration (0.158 ms^{-2}) and mediolateral range (0.276 ms^{-2}).

Table 7.4: Changes to pelvic accelerations between pre- and post-fatigue conditions

Measure	**Pre-Fatigue Mean (SD)**	**Post-Fatigue Mean (SD)**	**p value**
Vertical			
Vertical Accel	0.89 (0.53)	0.99 (0.62)	0.469
Vertical Decel	-4.45 (1.65)	-4.95 (1.94)	0.318
Vertical Range	5.34 (1.74)	5.94 (1.94)	0.218
Mediolateral			
Lateral Accel	0.84 (0.66)	1.10 (0.74)	0.069
Medial Decel	-2.05 (0.57)	-2.29 (0.77)	**0.015***
ML Range	2.90 (0.96)	3.40 (1.30)	**0.016***
Anteroposterior			
Anterior Accel	1.98 (0.49)	1.99 (0.47)	0.857
Posterior Decel	-0.50 (1.18)	-0.29 (0.51)	0.414
AP Range	2.48 (1.16)	2.28 (0.70)	0.467

**Denotes that the mean difference is significant at the .05 level*

7.2.1.3 Changes to thoracic accelerations

When differences between pre- and post-fatigue thoracic acceleration means were compared, no differences between conditions either reached significance or surpassed the relevant SWC (see Table 7.5).

Table 7.5: Group changes to thoracic accelerations between pre- and post-fatigue conditions

Measure	Pre-Fatigue Mean	Post-Fatigue Mean	p value
Vertical			
Vertical Accel	1.05 (0.43)	1.14 (0.31)	0.283
Vertical Decel	3.31 (1.56)	3.63 (1.87)	0.305
Vertical Range	4.36 (1.78)	4.77 (1.93)	0.238
Mediolateral			
Lateral Accel	0.73 (0.15)	0.76 (0.15)	0.579
Medial Decel	0.56 (0.37)	0.84 (0.63)	0.085
ML Range	1.29 (0.47)	1.59 (0.65)	0.087
Anteroposterior			
Anterior Accel	2.45 (0.49)	2.47 (0.94)	0.923
Posterior Decel	0.17 (0.09)	0.21 (0.17)	0.375
AP Range	2.28 (0.51)	2.26 (0.98)	0.93

**Denotes that the mean difference is significant at the .05 level*

7.2.2 Changes to local kinematic jerk

The peak local jerk experienced by each runner, at each of the 4 anatomical locations and along each of the vertical, mediolateral and anteroposterior axes, were extracted from the pre- and post-fatigue time-series of each runner. Subsequently, group means for each of 9 metrics, for each of the 4 anatomical attachment locations, were calculated. These metrics captured the magnitudes of the peak positive and peak negative jerk, and the magnitude of the jerk range extending between positive and negative peaks, for each of the vertical, mediolateral and anteroposterior axes. All pre- to post-fatigue jerk metrics were compared using repeated measures analysis of variance. The Smallest Worthwhile Changes (SWCs) for each metric were applied to the experimental data. Analysis first checked for interaction effects. The results, for all jerk metrics, are presented below.

7.2.2.1 Changes to tibial jerk

No interactions between either pre- and post-fatigue jerk metrics, or left and right tibial jerk, were detected. Repeated measures ANOVAs were conducted to explore main effects of side and pre- and post-fatigue conditions. However, no significant main effects were detected (see Table 7.6). Similarly, none of the differences between pre- and post-fatigue means, for any tibial jerk metrics, exceeded the calculated SWCs.

Table 7.6: Group changes to tibial jerk between pre- and post-fatigue conditions

Measure	Left tibia Pre Mean	Left tibia Post Mean	Right tibia Pre Mean	Right tibia Post Mean	Pre v Post p value (np^2)	Left v Right p value (np2)	Interaction p value (np2)
Vertical							
Upward Jerk	537.9 (230.3)	574.9 (219.1)	487.6 (162.3)	536.7 (188.2)	0.084 (0.19)	0.262 (0.08)	0.787 (0.01)
Downward Jerk	523.8 (201.4)	540.6 (201.9)	487.3 (165.3)	490.9 (189.2)	0.589 (0.02)	0.355 (0.06)	0.643 (0.02)
Vertical Range	1061.7 (380.1)	1115.6 (350.9)	974.9 (295.9)	1027.6 (350.9)	0.143 (0.14)	0.243 (0.09)	0.986 (0.01)
Mediolateral							
Lateral Jerk	511.7 (252.9)	522.9 (245.2)	801.5 (484.7)	751.1 (470.4)	0.734 (0.08)	0.063 (0.47)	0.055 (0.24)
Medial Jerk	773.1 (510.9)	767.6 (465.9)	549.9 (465.9)	542.1 (269.6)	0.879 (0.02)	0.074 (0.32)	0.932 (0.01)
ML Range	1284.7 (746.4)	1290.4 (700.4)	1351.5 (713.8)	1293.2 (689.0)	0.792 (0.01)	0.677 (0.01)	0.141 (0.15)
Anteroposterior							
Anterior Jerk	576.5 (271.8)	543.9 (276.90	531.2 (234.9)	519.1 (251.2)	0.484 (0.04)	0.231 (0.10)	0.541 (0.03)
Posterior Jerk	545.6 (277.2)	494.6 (245.2)	515.3 (231.1)	511.3 (261.8)	0.436 (0.04)	0.878 (0.01)	0.262 (0.09)
AP Range	1122.2 (459.9)	1038.5 (470.1)	1046.6 (408.6)	1030.4 (471.7)	0.419 (0.05)	0.526 (0.03)	0.303 (0.08)

**Denotes that the mean difference is significant at the .05 level*

7.2.2.2 Changes to pelvic jerk

When differences between pre- and post-fatigue pelvic jerk were compared, no significant changes were detected between conditions. Results are presented in Table. No mean differences, between pre- and post-conditions, for any jerk metric, exceeded the relevant SWCs.

Table 7.7. Group changes to pelvic jerk between pre- and post-fatigue conditions

Measure	Pre-Fatigue Mean (SD)	Post-Fatigue Mean (SD)	p value
Vertical			
Upward Jerk	291.3 (155.3)	310.6 (175.3)	0.251
Downward Jerk	326.1 (182.5)	347.8 (190.6)	0.203
Vertical Range	617.5 (335.1)	658.5 (362.3)	0.213
Mediolateral			
Lateral Jerk	206.7 (97.4)	221.5 (94.1)	0.441
Medial Jerk	157.6 (79.9)	165.2 (75.5)	0.321
ML Range	364.3 (157.2)	386.7 (155.4)	0.34
Anteroposterior			
Anterior Jerk	128.6 (41.1)	145.6 (65.6)	0.111
Posterior Jerk	151.0 (104.7)	161.4 (27.0)	0.329
AP Range	279.6 (143.7)	306.9 (164.2)	0.143

**Denotes that the mean difference is significant at the .05 level*

7.2.2.3 Changes to thoracic jerk

When the differences between pre- and post-fatigue thoracic jerk were compared, lateral thoracic jerk increased significantly (p=0.047) between pre- and post-fatigue conditions. No other significant changes were detected. Notably, the magnitude of change, noted for lateral thoracic jerk, between pre- and post-fatigue conditions, did not exceed the relevant SWC. Results are presented in Table 7.8.

Table 7.8. Group changes to thoracic jerk between pre- and post-fatigue conditions

Measure	Pre-Fatigue Mean (SD)	Post-Fatigue Mean (SD)	p value
Vertical			
Upward Jerk	257.1 (104.6)	315.4 (164.5)	0.161
Downward Jerk	247.6 (116.3)	246.6 (110.0)	0.961
Vertical Range	470.9 (196.3)	471.1 (205.6)	0.995
Mediolateral			
Lateral Jerk	57.9 (28.4)	75.7 (45.3)	**0.047**
Medial Jerk	93.3 (48.9)	108.9 (43.6)	0.276
ML Range	151.3 (72.8)	184.6 (84.9)	0.133
Anteroposterior			
Anterior Jerk	106.9 (45.1)	111.1 (52.3)	0.74
Posterior Jerk	90.8 (75.5)	111.6 (83.2)	0.241
AP Range	197.6 (117.6)	222.7 (129.5)	0.401

**Denotes that the mean difference is significant at the .05 level*

7.2.3 Changes to Shock Attenuation between pre- and post-fatigue conditions

Shock attenuation was calculated, following the suggestions of both MacDermid and colleagues (2017) and Sinclair (2017), as the ratios between peak pelvic, or peak thoracic, acceleration and peak tibial acceleration, using the formulas:

Shock Attenuation (Pelvis) = 1 – (Peak Vertical Pelvic Shock/Peak Vertical Tibial Shock) *100

Shock Attenuation (Thorax) = 1 – (Peak Vertical Thoracic Shock/Peak Vertical Tibial Shock) *1

The shock attenuation experienced between left tibia and pelvis (p<0.001) and between right tibia and pelvis (p<0.001) both changed significantly. In both instances, shock attenuation increased between conditions, as illustrated in Table 7.9. However, upon application of the relevant SWCs, the magnitudes of neither of these changes were greater than the SWC calculated for these metrics (See Appendix H).

Table 7.9. Changes to Shock Attenuation between pre- and post-fatigue conditions

Shock Attenuation	Pre-Fatigue Mean (SD)	Post-Fatigue Mean (SD)	p value
Left Tibia to Pelvis	-70.57 (19.84)	-74.91 (22.69)	**<0.001***
Left Tibia to Thorax	-52.81 (20.23)	-51.98 (24.89)	0.299
Right Tibia to Pelvis	-82.55 (27.89)	-90.41 (29.35)	**<0.001***
Right Tibia to Thorax	-60.43 (22.71)	-61.37 (27.77)	0.881

**Denotes that the mean difference is significant at the .05 level*

7.2.4 Changes to Resultant Accelerations between pre- and post-fatigue conditions

The resultant accelerations experienced at tibial, pelvic, and thoracic sites were calculated using the vertically, medially- and anteriorly-directed peak decelerations experienced during ground contact, following the formula suggested by Sheerin and colleagues (2018):

$$\textbf{Resultant Acceleration (RA)} = \sqrt{(\text{Vertical deceleration peak})^2 + (\text{Medial deceleration peak})^2 + (\text{Posterior deceleration peak})^2}$$

Pre- and post-fatigue comparisons revealed that left tibial resultant accelerations changed significantly (p=0.023). No other resultant metrics changed between conditions. The difference between pre- and post-fatigue left tibial resultant means (0.770 ms^{-2}) was greater than the relevant SWC (0.149 ms^{-2}) derived for tibial resultant accelerations (See Appendix H).

Table 7.10: Descriptive statistics: Resultant Accelerations

Resultant Accelerations	Pre-Fatigue Mean (SD)	Post-Fatigue Mean (SD)	p value
Left Tibial	9.94 (2.92)	10.71 (2.47)	**0.023***
Right Tibial	8.42 (2.41)	8.92 (2.32)	0.581
Pelvic	5.47 (1.40)	5.96 (1.82)	0.231
Thoracic	4.43 (1.40)	4.76 (2.03)	0.291

**Denotes that the mean difference is significant at the .05 level*

7.3 Results: Individual analysis of local accelerations and acceleration-derived metrics

7.3.1 Individual changes to local accelerations between pre- and post-fatigue conditions

7.3.1.1 Changes to tibial acceleration metrics

The differences between pre- and post-fatigue conditions, for each individual runner, were compared using the relevant metric-specific SWC (See Appendix H). All pre- to post-fatigue acceleration metrics, for each individual runner, were also compared using paired sample t tests, these results are presented in Appendix I.

At the left tibia the difference between pre- and post-fatigue acceleration means exceeded the relevant SWC for 100, of the 135, local acceleration metrics. The highest incidence of meaningful changes (14) occurred mediolaterally, where 14 of the 15 runners exhibited meaningful increases or decreases to the range of tibial accelerations experienced between medial to lateral acceleration peaks. Notably, every runner experienced pre- to post-fatigue changes. The number of changes experienced per runner ranged from a minimum of 2, to a maximum of 9. Of further note, all metrics, excepting vertical deceleration, alternatively demonstrated both increases and decreases to peak magnitudes between conditions. However, on all 6 occasions where vertical decelerations exceeded the relevant SWC, vertical deceleration magnitudes increased. In no case did left tibial vertical deceleration decrease.

At the right tibia the difference between pre- and post-fatigue acceleration means exceeded the relevant SWC for 89, of the 135, metrics. Again, as with the left tibia, the highest incidence of meaningful changes (14) occurred mediolaterally, where 14 of the 15 runners exhibited meaningful increases or decreases to the range of tibial accelerations experienced between medial to lateral acceleration peaks. Notably, every runner exhibited changes, to pre- and post-fatigue means, that exceeded the relevant SWC. The number of changes experienced per runner ranged from a minimum of 2, to a maximum of 8. Again, as was the case at the left tibia, all metrics, excepting vertical deceleration, demonstrated both increases and decreases to acceleration peaks between conditions. All changes to vertical deceleration at the right tibia, which exceeded the relevant SWC, of which there 4, resulted in increases, but not decreases, to vertical deceleration magnitudes.

7.3.1.2 Changes to pelvic acceleration metrics

At the pelvis, the difference between pre- and post-fatigue acceleration means exceeded the relevant SWC for 69, of the 135, metrics. The highest incidence of change occurred vertically, where 10 of the 15 runners exhibited meaningful increases or decreases to the range experienced between vertical acceleration to deceleration peaks. Every runner experienced pre- to post-fatigue changes. The number of pelvic acceleration changes experienced per runner ranged from a minimum of 1, to a maximum of 8. Furthermore, at the pelvis, all 9 acceleration metrics variously exhibited meaningful increases and/or decreases to acceleration magnitudes.

7.3.1.3 Changes to thoracic acceleration metrics

At the thorax, the difference between pre- and post-fatigue acceleration means exceeded the relevant SWC for 89 of the 135 metrics. The highest incidence of change occurred vertically, where 14 of the 15 runners exhibited meaningful increases or decreases to the range experienced between vertical acceleration to deceleration peaks. Every runner experienced pre- to post-fatigue changes. The number of pelvic acceleration changes experienced per runner ranged from a minimum of 1, to a maximum of 8. Furthermore, at the thorax, all 9 acceleration metrics variously exhibited meaningful increases and decreases to acceleration magnitudes. A summary of the results, for all acceleration metrics, are presented in the tables below.

Table Legend

🟩	Significant increases to the magnitude of positive (+) accelerations
🟧	No significant change
🟥	Significant increases to the magnitude of negative (-) accelerations/decelerations

Table 7.11: Individual changes to local accelerations between pre- and post-fatigue conditions

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Left Tibia										
Positive increase	9	0	11	7	1	10	6	4	7	55
No change	4	9	3	4	6	1	4	2	2	35
Negative increase	2	6	1	4	8	4	5	9	6	45
Right Tibia										
Positive increase	6	0	9	5	6	8	5	3	8	50
No change	7	11	2	6	4	1	5	6	4	46
Negative increase	2	4	4	4	5	6	5	6	3	39
Pelvis										
Positive increase	5	4	6	6	2	5	1	3	5	37
No change	4	5	5	8	10	9	12	5	8	66
Negative increase	6	6	4	1	3	1	2	7	2	32
Thorax										
Positive increase	9	5	9	4	1	9	4	5	3	49
No change	4	2	1	8	6	5	6	7	7	46
Negative increase	2	8	5	3	8	1	5	3	5	40

7.3.1.4 Local acceleration differences exceeding tibial SWCs

Local acceleration changes were expressed as a percentage of maximum possible changes. These percentages of change ranged from 20% (for right tibial anterior acceleration), to 93% (for tibial medio-lateral acceleration range). Percentage changes for each metric are illustrated in Table 7.10, below.

Table 7.12 Percentage of acceleration metrics exceeding the Smallest Worthwhile Change

Sensor Location	Vertical Accel	Vertical Decel	Vertical Range	Lateral Accel	Medial Accel	ML Range	Anterior Accel	Posterior Accel	AP Range
Left tibia	73%	40%	80%	73%	60%	93%	73%	87%	87%
Right tibia	53%	27%	87%	60%	73%	93%	67%	60%	73%
Pelvis	73%	67%	60%	47%	33%	40%	20%	67%	47%
Thorax	73%	67%	93%	47%	60%	67%	60%	53%	53%

7.3.2 Individual changes to local kinematic jerk

The differences between pre- and post-fatigue conditions were compared using the relevant jerk specific SWCs (see Appendix H). Subsequent results are presented in this section. Furthermore, all pre- to post-fatigue jerk metrics, for each individual runner, were compared using paired sample t tests. These results are presented in Appendix J.

7.3.2.1 Changes to local jerk metrics

Across the 4 sensor locations, on each of the 15 runners only 70, of a potential maximum of 540, individual local jerk metrics exhibited meaningful change between pre- and post-fatigue conditions. The local jerk metric exhibiting the greatest frequency of meaningful change was vertical thoracic acceleration (5). Notably, unlike the corresponding local acceleration measures, the majority of jerk metrics failed to exhibit any meaningful changes between pre- and post-fatigue conditions. Importantly, the number of changes exceeding the relevant jerk SWCs was substantially less than the number of changes observed for local accelerations. Similarly, the number of changes exceeding the jerk-specific SWCs was substantially less than the number of incidences of significance established, between conditions, using inferential statistics (See Appendix J).

Table 7.13. Individual changes to local jerk metrics between pre- and post-fatigue conditions

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Left Tibia										
Positive increase	**3**	**0**	**1**	**0**	**2**	**0**	**0**	**1**	**0**	7
No change	9	13	12	14	11	14	14	14	13	114
Negative increase	3	2	2	1	2	1	1	0	2	14
Right Tibia										
Positive increase	**0**	**1**	**1**	**0**	**0**	**0**	**5**	**0**	**0**	7
No change	15	14	14	14	14	13	3	15	15	117
Negative increase	0	0	0	1	1	2	7	0	0	11
Pelvis										
Positive increase	0	0	1	0	1	1	3	1	2	9
No change	15	15	14	15	13	13	10	10	10	115
Negative increase	0	0	0	0	1	1	2	4	3	11
Thorax										
Positive increase	5	0	0	0	1	0	0	0	0	6
No change	9	14	14	14	14	14	15	15	15	124
Negative increase	1	1	1	1	0	1	0	0	0	5

7.3.3 Individual changes to Shock Attenuation between pre- and post-fatigue conditions

The SWCs, for both SA – P and SA – T metrics, were calculated and applied to the experimental data. Pre- and post-conditions were also compared using paired sample t tests. Statistical analysis established significant changes, between the magnitudes of pre- and post-fatigue SA - P metrics, for 7 of the 15 runners. Of these 7 runners, 4 decreased, whilst 3 runners increased, the magnitude of shock attenuated between conditions. Similarly, individual analysis found significant changes between the magnitudes of pre- and post-fatigue SA - T metrics, for 8 of the 15 runners. Of these 8, 5 runners decreased, whilst 3 runners increased the magnitude of shock attenuated. Notably, however, upon application of the relevant SWC values, the differences in means between pre- and post-fatigue conditions only exceeded the relevant SWC on 5 occasions. Twice for SA- P and three times for SA- T. These results, for both SA – P and SA – T, are presented in Table 7.14.

Table 7.14. Shock attenuation between tibia and pelvis (SA -P) and between tibia and thorax (SA- T)

Runner	Shock Attenuation – Tibia to Pelvis			Shock Attenuation – Tibia to Thorax		
	Pre-fatigue Mean (SD)	Post-fatigue Mean (SD)	p value	Pre-fatigue Mean (SD)	Post-fatigue Mean (SD)	p value
1	140.9 (14.5)	132.0 (20.4)	**<0.001**	98.0 (17.2)	69.4 (10.4)	**<0.001***
2	80.3 (20.1)	90.7 (4.8)	**0.020**	50.8 (18.8)	61.3 (7.6)	0.051
3	41.4 (8.5)	102.2 (3.6)	**0.004***	55.1 (6.0)	65.7 (8.1)	**<0.001**
4	101.4 (10.6)	100.9 (10.5)	0.913	78 (5.8)	70 (9.8)	**<0.001**
5	65.1 (6.9)	69 (10.6)	0.224	78 (5.8)	50.3 (9.0)	**0.017***
6	63.4 (5.7)	35.3 (2.8)	**0.042***	64.8 (8.6)	35.5 (2.9)	**<0.001***
7	73.8 (6.6)	70.4 (10.7)	0.237	80.8 (6.5)	94.7 (3.4	**<0.001**
8	91.8 (8.5)	94.3 (8.3)	**0.041**	22 (2.4)	27.2 (2.8)	**<0.001**
9	64.9 (6.4)	63.8 (6.9)	0.59	75.1 (8.2)	74.8 (5.8)	0.924
10	44.4 (5.1)	38.6 (3.6)	**0.006**	44.6 (6.1)	48.1 (4.3)	0.063
11	89.3 (5.2)	79.1 (8.4)	**0.032**	28.9 (3.4)	25.5 (4.6)	**0.012**
12	55.5 (9.2)	55.4 (10.6)	0.965	50.2 (8.9)	52.8 (6.2)	0.474
13	46.2 (8.3)	44.1 (5.5)	0.378	39.8 (4.2)	42.8 (7.0)	0.053
14	87.0 (12.4)	86.5 (16.8)	0.925	80.2 (11.4)	82.3 (9.5)	0.624
15	98.5 (6.8)	95.9 (6.7)	0.329	19.9 (3.5)	22.0 (8.5)	0.212

**Denotes that difference between pre- and post-fatigue means exceeded the Smallest Worthwhile Change*

7.3.4 Individual changes to Resultant Accelerations between pre- and post-fatigue conditions

The SWCs for RA metrics were applied to the experimental data. Individual analysis established that 10 of the 15 runners experienced significant change to the magnitude of left tibial RAs between conditions. Of these 10 runners, 9 increased the magnitude of RAs, while one significantly reduced the magnitude of left tibial RA. Similarly, 10 of the 15 runners experienced significant change to the magnitude of right tibial RAs between conditions. Of these 10 runners, 9 significantly increased the magnitude of RAs, while one reduced the magnitude of right tibial RA. Notably, in each of these 20 incidences of significant change, the magnitude of the differences, between pre- and post-fatigue tibial RA means, exceeded the SWC (0.211 ms^{-2}) calculated for tibial RA.

At the pelvis, 6 runners exhibited significant change to pelvic RAs between pre- and post-fatigue conditions. 4 runners increased pelvic RA, whilst 2 decreased pelvic RA. However, the magnitude of the differences, between pre- and post-fatigue pelvic RA means, exceeded the relevant SWC (0.206 ms^{-2}) for only 5 of these runners (see Table 7.15).

At the thorax, 13 runners exhibited significant change to thoracic RAs between pre- and post-fatigue conditions. Of these, 8 runners increased, and 5 runners decreased, thoracic RAs. For each of these instances of change the differences between pre- and post-fatigue pelvic RA means exceeded the relevant SWC (0.176 ms^{-2}) for this measure. RA results, for all 4 sensor locations, are presented in Tables 7.15 and 7.16.

Table 7.15: Individual changes to left and right tibial resultant accelerations between pre-and post-fatigue conditions

Runner	**Left tibial Resultant Acceleration**			**Right tibial Resultant Acceleration**		
	Pre-fatigue Mean (SD)	**Post-fatigue Mean (SD)**	**p value**	**Pre-fatigue Mean (SD)**	**Post-fatigue Mean (SD)**	**p value**
1	5.81 (0.41)	7.97 (0.51)	**0.002***	5.02 (0.41)	5.73 (0.61)	**P<0.001***
2	8.62 (0.81)	8.56 (0.64)	0.713	7.52 (0.64)	6.82 (1.11)	**0.010***
3	7.92 (0.43)	10.55 (0.51)	**0.002***	10.73 (0.57)	11.51 (0.45)	**0.002***
4	8.73 (0.87)	9.54 (0.78)	**0.003***	8.82 (0.56)	9.91 (0.54)	**P<0.001***
5	7.71 (0.36)	7.86 (0.54)	0.232	6.52 (0.45)	6.97 (0.62)	**0.010***
6	5.94 (0.63)	7.74 (0.24)	**0.002***	5.47 (0.34)	6.36 (0.21)	**0.003***
7	12.85 (0.61)	14.23 (0.42)	**0.004***	11.54 (0.33)	13.24 (0.6)	**0.002***
8	7.96 (0.54)	8.13 (0.56)	0.413	8.25 (0.24)	8.31 (0.4)	0.100
9	10.61 (0.32)	10.62 (0.34)	0.921	10.56 (0.43)	10.54 (0.48)	0.442
10	10.64 (0.42)	11.37 (0.79)	**0.003***	11.44 (0.91)	11.52 (0.76)	0.644
11	10.91 (0.48)	10.56 (0.54)	**0.020***	8.65 (0.94)	9.91 (0.6)5	**P<0.001***
12	7.65 (0.95)	8.54 (0.52)	**0.012***	9.55 (0.85)	11.82 (0.44)	**P<0.001***
13	7.51 (0.33)	7.74 (0.42)	0.223	6.44 (0.44)	6.32 (0.44)	0.568
14	9.93 (0.51)	10.81 (0.56)	**0.002***	9.82 (0.95)	9.97 (0.77)	0.186
15	7.62 (0.67)	8.64 (0.17)	**0.012***	6.21 (0.44)	6.94 (0.41)	**0.003***

**RA metrics achieved significance and the difference between pre- and post-fatigue means exceeded the Smallest Worthwhile Change*

Table 7.16. Individual changes to pelvic and thoracic Resultant Accelerations between pre- and post-fatigue conditions

Runner	Pelvic Resultant Acceleration			Thoracic Resultant Acceleration		
	Pre-fatigue Mean (SD)	Post-fatigue Mean (SD)	p value	Pre-fatigue Mean (SD)	Post-fatigue Mean (SD)	p value
1	4.94 (0.57)	5.16 (0.47)	0.144	4.53 (0.56)	3.74 (0.38)	**0.003***
2	6.50 (0.71)	7.02 (0.20)	**0.005***	4.21 (0.38)	4.77 (0.94)	**0.028***
3	2.79 (0.53)	8.50 (0.28)	**P<.001**	2.84 (0.35)	5.83 (0.55)	**0.003***
4	6.48 (0.45)	6.48 (0.37)	0.947	4.74 (0.34)	4.38 (0.41)	**0.009***
5	4.81 (0.54)	5.01 (0.63)	0.083	4.26 (0.50)	3.56 (0.44)	**0.002***
6	2.80 (0.29)	1.05 (0.25)	**P<.001***	2.75 (0.33)	0.32 (0.13)	**0.003***
7	6.40 (0.83)	6.36 (1.09)	0.814	6.66 (0.54)	7.99 (0.31)	**0.006***
8	6.02 (0.48)	6.15 (0.35)	0.369	1.56 (0.18)	1.87 (0.13)	**0.002***
9	4.81 (0.7)	5.00 (0.47)	0.053	5.36 (0.51)	5.35 (0.44)	0.981
10	4.91 (0.56)	4.74 (0.47)	0.171	4.99 (0.47)	5.54 (0.46)	**0.003***
11	6.45 (0.40)	5.70 (0.52)	**P<.001***	2.17 (0.25(	2.08 (0.28)	0.349
12	5.51 (0.43)	5.99 (0.86)	**0.045***	4.71 (0.48)	3.57 (0.37)	**0.014***
13	3.25 (0.61)	3.28 (0.62)	0.857	4.84 (0.19)	3.01 (0.31)	**0.002***
14	5.49 (0.81)	5.61 (0.95)	0.588	1.30 (0.60)	5.29 (0.63)	**0.043***
15	6.31 (0.41)	6.73 (0.37)	**0.002***	1.30 (0.18)	1.62 (0.52)	**0.013***

**The difference between pre- and post-fatigue means exceeded the Smallest Worthwhile Change*

7.4. Collective observations

Certain observations detailed in this chapter appear novel and relevant to the thesis objectives. Comparison of runner-specific pre- and post-condition means, against the relevant SWCs, revealed unexpectedly pervasive changes across local acceleration, shock attenuation and resultant acceleration metrics. In contrast, however, local jerk metrics did not exhibit extensive change between pre- and post-fatigue conditions.

Both the magnitude of peak vertical tibial decelerations and the range of vertical tibial accelerations, experienced between pre- and post-fatigue conditions, exceeded the specific SWCs for those metrics. In an uncommonly reported observation, meaningful differences between left and right tibial means, exceeding the relevant SWCs, were noted for vertical decelerations and medial and lateral accelerations. At the pelvis, the differences between pre- and post-fatigue means exceeded the magnitude of the calculated SWC metrics for both medial decelerations and the acceleration range extending between medial and lateral peaks. None of the, rarely assessed, thoracic acceleration measures, however, changed between pre- and post-conditions.

Runner-specific analysis illustrated the extent of intra- and inter- changes between pre- and post-conditions. Application of metric specific SWCs to individual data highlighted a high frequency of meaningful change, across metrics and sensor locations. In contrast, analysis of the magnitudes of mean jerk measures did not detect changes to any grouped jerk metric at any of the sensor locations.

The application of the relevant SWCs to individual runner’s jerk metrics revealed few incidences of change between conditions. Although many changes to jerk metrics (approximately 50%) achieved statistical significance, when relevant smallest worthwhile changes were applied, the magnitude of the SWCs predominantly exceeded the differences between pre- and post-fatigue jerk means.

Shock attenuation metrics did not change, as determined by application of the relevant SWCs. Upon individual analyses, the differences in runner means between pre- and post-fatigue conditions exceeded the relevant SWC for 5 of the 15 runners. Although the difference in left tibial resultant acceleration means, between conditions, exceeded the relevant SWC, the difference between right tibial resultant means did not.

The findings of the individual data analyses emphasise the observation that different anatomical locations are commonly subjected to very different mechanical experiences., as kinematic experiences varied extensively between both tibias, between pelvis and thorax, and between upper- and lower-body sensor locations. An additional noteworthy observation is that, just as individual runners experienced differential patterns of change between anatomical attachment sites, they also experienced different frequencies of change between upper- and lower-body sites. To illustrate: 6 runners experienced more lower body, than upper-body, local acceleration changes; 1 experienced an equal distribution of change between upper- and lower-body; 7 experienced more changes to upper-, than lower-, body locations. Crucially, these local changes were distributed across the 4 anatomical locations in individually customised signatures of change which were unique to each runner.

Notably, in an additional uncommonly reported finding, the experiences of left and right tibias differed across a number of metrics. Specifically, the magnitudes of the vertical decelerations, lateral accelerations, medial decelerations, and posterior decelerations were differently experienced at each tibia, as were both resultant accelerations.

7.5 Conclusions

The objectives of this chapter were threefold:

i. To determine whether acceleration and acceleration-derived metrics, advocated in the literature as indicative of running-induced fatigue, change following a moderately fatiguing run
ii. To perform both whole-cohort and individual analysis to identify discrepancies between group and runner-specific observations
iii. To establish whether running-induced fatigue changes local accelerations and acceleration-derived metrics in a uniform manner across runners, or whether such changes vary in a metric-specific manner

The core findings of the analysis presented in this chapter suggest that:

i. A number of local acceleration measures, and one resultant acceleration measure, meaningfully changed subsequent to fatigue
ii. The shock loading experiences of left and right tibias meaningful differed between pre- and post-fatigue conditions
iii. Runner-specific analysis revealed substantial variation in the pattern of meaningful changes exhibited by each runner
iv. Excepting vertical tibial decelerations, which always increased between conditions, all other metrics variously demonstrated instances of both increases and decreases in peak magnitudes between conditions
v. Each runner exhibited a unique pattern of meaningful changes, across metrics, suggesting that individual responses to fatigue may be highly personalised

The implications of these findings will be explored, in more detail, in chapter 9.

Chapter 8:

Evaluating changes to acceleration and acceleration-derived metrics during a moderately fatiguing run

8.1 Introduction

The balance of evidence, reviewed in chapters 2 through 5, suggests that accumulating fatigue inhibits running coordination leading to a reduced capacity to optimally manage the perturbations imposed during ground contact. The analysis presented in Chapter 7 investigated how local accelerations and acceleration-derived metrics, each suggested within the literature as indicative of running-induced fatigue, changed before and after the fatiguing protocol. Yet, how accumulating fatigue, as reflected by the experimental metrics, progresses over the course of a fatiguing run, remains undocumented. Similarly, whether fatigue-induced changes to key running behaviours occur suddenly and simultaneously, at some discernible transition point, or gradually and progressively accrue across time, remains largely unexplored.

To examine these questions the analysis detailed in this chapter tracked acceleration time-series data, for 30 seconds of every minute, across a continuous run-to-fatigue protocol. As noted in section 6.2.2, a Borg (1982) Category Ratio-10 scale (CR10), was administered every 2 minutes and after each runner had signalled their intent to stop running. The Smallest Worthwhile Changes (SWCs), calculated as per the recommendations of Hopkins (2000) and presented in Chapter 6, were applied to all metrics.

8.1.1 Objectives

Accordingly, the specific objectives of the analysis contained in this chapter are:

1. To evaluate whether fatigue-induced change progresses uniformly across acceleration-derived metrics, each suggested as indicative of fatigue, over the course of a moderately fatiguing run
2. To assess whether local accelerations and acceleration-derived metrics change in a consistent manner across measures, and across attachment locations, over the course of a fatiguing run
3. To examine whether the progression of fatigue, over the course of the fatiguing run, is experienced similarly across a cohort of runners, or is experienced in an individually idiosyncratic, runner-specific manner

8.2 Results: Group Analysis of local accelerations and acceleration-derived metrics

The acceleration peaks experienced by each runner during ground contact, at each of the 4 anatomical locations and along each of the vertical, mediolateral and anteroposterior axes, were identified and extracted, for each runner, from the quartile time-series of the fatiguing run. Using this acceleration data, local jerk, shock attenuations and resultant accelerations, for each of the 4 anatomical attachment locations, and for each of 9 locational and directional-specific metrics, were calculated. These metrics captured peak positive and peak negative magnitudes, and the magnitude of the range experienced between positive and negative peaks. All quartile datasets were compared using repeated measures analysis of variance. The Smallest Worthwhile Changes (SWCs), for each metric, were also calculated and applied to the experimental dataset. The results, for all metrics, are presented below.

Across the 60 quartiles of the experimental data collection, time-series data transferred wirelessly from the integrated measurement sensors to the experimental laptop. During this process the experimental information collected for 59 of the 60 distinct quartiles transferred successfully. The data for one runner (Participant 11), however, did not record during the 4th quartile of the fatiguing run. The reasons for this failure are not clear and the sensors did successfully record the subsequent 60 second post-fatigue interval data. Accordingly, due to this unexplained collection error, no 4th quartile metrics are presented for this runner.

8.2.1 Changes to local accelerations

A two factor repeated measures analysis of variance examined differences between left and right tibial accelerations and between quartiles and for tibial accelerations in all three directions. Analysis checked for interaction effects between left and right sides and between quartiles. A repeated measures analysis of variance also examined differences between accelerations, across quartiles, at both pelvic and thoracic sensor locations. Post hoc pairwise comparisons were conducted for all significant main effects.

8.2.1.1 Changes to tibial accelerations

No significant interactions between left and right tibias and quartiles were detected. Significant main effects were seen for the magnitude of peak vertical accelerations (F(1,14)=6.41, p<.001 ES=0.33 - Large) and for the acceleration range experienced between maximum and minimum vertical peaks experienced between quartiles (F(1,14)=11.46, p<.001 ES=0.46 – Large).

The magnitudes of the peak vertical accelerations experienced by left and right tibias showed a significant main effect (F(1,14)=15.09, p=0.002 ES=0.54 – Large). The magnitudes of the peak vertical decelerations experienced by left and right tibias also differed significantly (F(1,14)= 32.91, p<.001 ES=0.71 – Large) (See Table 8.1). Similarly, the magnitudes of medially-directed accelerations experienced by left and right tibias differed significantly (F(1,14)=13.76, p<0.001 ES =0.49 – Large), as did the magnitudes of laterally-directed accelerations experienced by the left and right tibias (F(1,14)=12.12, p=0.004 ES = 0.48 – Large), as shown in Table 8.1.

Post-hoc comparisons revealed that vertical accelerations increased between quartiles 1 and 3 (p = 0.041); quartiles 1 and 4 (p<.001), and quartiles 2 and 4 (p = 0.028). The magnitude of acceleration range experienced at the tibias increased between quartiles 1 and 2 (p = 0.018); quartiles 1 and 3 (p = 0.002); quartiles 1 and 4 (p = 0.003); quartiles 2 and 4 (p = 0.018) and quartiles 3 and 4 (p = 0.009), as shown in Table 8.2.

The magnitude of vertical acceleration change observed between quartiles 1 and 3, and quartiles 1 and 4, both exceeded the SWC for that metric. The SWC for vertical acceleration (0.282 ms^{-2}), however, was greater than the magnitude of measured change between quartiles 2 and 4 (0.243 ms^{-2}). Each of the 5 observed changes to vertical acceleration range, between quartiles, exceeded the magnitude of the SWC (0.220 ms^{-2}) for that metric.

Post-hoc comparisons of the accelerations experienced by left and right tibias revealed that vertical (p=0.002) and lateral accelerations (p = 0.004) were higher at the right tibia, whereas vertical decelerations (p=0.003) and medial accelerations were higher at the left tibia (p=0.004), as shown in Table 8.3. In each of these cases the magnitude of the observed differences between left and right tibias exceeded the relevant SWC for those metrics (see Appendix H).

Table 8.1: Changes to tibial accelerations

Measure	Left tibia Mean (SD)				Right tibia Mean (SD)				Left v Right	Quartiles	Interaction
	Q.1	Q.2	Q.3	Q.4	Q.1	Q.2	Q.3	Q.4	p value (np^2)	p value (np^2)	p value (np^2)
Vertical											
Accel	1.39 (0.83)	1.71 (1.09)	1.71 (1.13)	1.94 (1.01)	2.06 (1.12)	2.31 (1.45)	2.39 (1.53)	2.57 (1.60)	**0.002* (0.53)**	**0.001* (0.33)**	0.738 (0.10)
Decel	6.03 (1.21)	6.20 (1.11)	6.30 (1.04)	6.40 (1.01)	5.21 (1.44)	5.40 (1.36)	5.49 (1.25)	5.48 (1.30)	**p<.001* (0.72)**	0.105 (0.17)	0.588 (0.05)
Range	7.42 (1.55)	7.91 (1.88)	8.01 (1.80)	8.34 (1.61)	7.28 (2.18)	7.71 (2.47)	7.89 (2.39)	8.05 (2.52)	0.477 (0.40)	**p<.001* (0.47)**	0.790 (0.09)
Mediolateral											
Lateral	2.24 (2.28)	2.22 (2.40)	2.32 (2.33)	2.32 (2.46)	3.45 (1.96)	3.92 (2.43)	3.83 (2.20)	4.00 (2.55)	**0.004* (0.48)**	0.609 (0.04)	0.635 (0.14)
Medial	5.18 (2.85)	4.82 (2.70)	4.90 (2.75)	5.09 (2.60)	3.79 (2.17)	3.83 (2.30)	3.80 (2.24)	3.83 (2.27)	**0.003* (0.49)**	0.971 (0.01)	0.375 (0.24)
Range	7.42 (4.93)	7.04 (4.80)	7.22 (4.79)	7.41 (4.85)	7.24 (4.03)	7.75 (4.54)	7.64 (4.24)	7.84 (4.62)	0.528 (0.03)	0.731 (0.02)	0.564 (0.05)
Anteroposterior											
Anterior	0.60 (0.60)	0.81 (1.37)	0.77 (0.94)	0.71 (0.76)	1.16 (1.37)	1.02 (0.62)	1.12 (0.77)	0.95 (0.69)	0.180 (0.13)	0.920 (0.02)	0.177 (0.12)
Posterior	4.59 (1.46)	4.63 (1.63)	4.60 (1.60)	4.56 (1.64)	4.24 (1.56)	4.22 (1.32)	4.22 (1.62)	4.27 (1.74)	0.105 (0.19)	1.000 (0.01)	0.415 (0.07)
Range	5.20 (1.57)	5.44 (2.00)	5.37 (2.02)	5.30 (1.94)	5.41 (2.50)	5.24 (1.82)	5.34 (1.93)	5.22 (1.94)	0.950 (0.01)	0.986 (0.04)	0.105 (0.19)

**Denotes that the mean difference is significant at the .05 level*

Table 8.2: Post hoc comparisons using Least Significant Differences between Quartiles

Post hoc comparisons using Least Significant Differences between Quartiles							
		Q1 vs Q2	**Q1 vs Q3**	**Q1 vs Q4**	**Q2 v Q3**	**Q2 v Q4**	**Q3 v Q4**
Vertical Accel	**Mean diff.**	-0.28	-0.33	-0.52	-0.47	-0.24	-0.2
	p value	0.132	**0.041**	**P<.001**	0.556	**0.028**	0.079
	Upper bound	-0.02	-0.02	-0.22	-0.62	-0.12	-0.64
	Lower bound	-0.64	-0.64	-0.83	-0.06	-0.22	-0.37
Accel Range	**Mean diff.**	-0.46	-0.6	-0.84	-0.14	-0.39	-0.24
	p value	**0.018**	**0.002**	**0.003**	0.092	**0.018**	**0.009**
	Upper bound	-0.08	-0.25	-0.41	0.05	-0.08	-0.07
	Lower bound	-0.84	-0.96	-0.28	-0.33	-0.69	0.41

Table 8.3: Post hoc comparisons using Least Significant Differences between Left and Right tibias

Post hoc comparisons using Least Significant Differences between Left and Right tibias				
	Mean diff.	**p value**	**Upper bound**	**Lower bound**
Vertical Accel	0.65	**0.002***	1.00	-0.29
Vertical Decel	0.84	**0.003***	0.05	-1.15
Lateral Accel	0.42	**0.004***	-0.01	-0.35
Medial Accel	1.19	**0.004***	1.94	0.45

**Denotes that the mean difference is significant at the .05 level*

8.2.1.2 Changes to pelvic acceleration

No significant interactions between quartiles were detected. Subsequently, repeated measures ANOVAs were conducted to explore main effects of quartiles. As illustrated in Table 8.3, there were no significant effects. No differences between quartile means exceeded the relevant SWCs.

Table 8.4: Changes to pelvic accelerations

Measure	Q1	Q2	Q3	Q4	Quartiles p value (np^2)
			Vertical		
Accel	0.79 (0.47)	0.77 (0.43)	0.82 (0.43)	0.74 (0.49)	0.683 (0.37)
Decel	4.94 (1.47)	4.86 (1.41)	5.04 (1.52)	4.83 (1.68)	0.409 (0.71)
Vert Range	5.73 (1.58)	5.63 (1.54)	5.85 (1.71)	5.57 (1.88)	0.443 (0.07)
			Mediolateral		
Lateral	0.83 (0.55)	0.97 (0.68)	1.06 (0.57)	0.97 (0.75)	0.287 (0.28)
Medial	2.01 (0.49)	2.19 (0.65)	2.29 (0.76)	2.30 (0.82)	0.292 (0.09)
ML Range	2.84 (0.84)	3.16 (1.10)	3.35 (1.10)	3.28 (1.36)	0.082 (0.15)
			Anteroposterior		
Anterior	1.83 (0.43)	1.89 (0.46)	1.84 (0.43)	1.90 (0.45)	0.676 (0.04)
Posterior	0.16 (0.43)	0.18 (0.41)	0.21 (0.39)	0.17 (0.41)	0.241 (0.05)
AP Range	1.99 (0.62)	2.06 (0.60)	2.05 (0.51)	2.07 (0.58)	0.930 (0.01)

**Denotes that the mean difference is significant at the .05 level*

8.2.1.3 Changes to thoracic accelerations

Significant main effects were detected for thoracic posterior accelerations and for the range of accelerations experienced between posterior and anterior acceleration peaks. Post-hoc comparisons revealed that posterior thoracic deceleration increased significantly (p=0.016) between quartile 1 and quartile 4. Similarly, the range of anteroposterior accelerations experienced at the thorax decreased significantly between quartiles 2 and 3 (p=0.018). In the case of posterior thoracic deceleration, the change in means between quartiles (0.153 ms^{-2}) exceeded the SWC (0.047 ms^{-2}) for that metric. Similarly, the change in means between anterior-posterior acceleration ranges, measured between quartiles 2 and 3 (0.157 ms^{-2}), exceeded the SWC (0.150 ms^{-2}) for that metric (see Appendix H).

Table 8.5 Changes to thoracic accelerations

Measure	Q.1	Q.2	Q.3	Q.4	p value (np^2)
Vertical					
Accel.	0.97 (0.27)	1.01 (0.20)	1.02 (0.23)	0.96 (0.23)	0.884 (0.02)
Decel.	3.32 (1.27)	3.54 (1.50)	3.41 (1.47)	3.46 (1.48)	0.485 (0.06)
Range	4.30 (1.24)	4.54 (1.50)	4.43 (1.45)	4.42 (1.50)	0.591 (0.05)
Mediolateral					
Lateral	0.69 (0.13)	0.71 (0.13)	0.73 (0.16)	0.77 (0.16)	0.063 (0.17)
Medial	0.58 (0.44)	0.67 (0.54)	0.73 (0.57)	0.63 (0.49)	0.519 (0.06)
Range	1.27 (0.46)	1.38 (0.56)	1.44 (0.62)	1.40 (0.47)	0.434 (0.07)
Anteroposterior					
Anterior	2.43 (0.63)	2.47 (0.77)	2.35 (0.79)	2.41 (0.72)	0.213 (0.11)
Posterior	0.24 (0.10)	0.28 (0.08)	0.31 (0.11)	0.32 (0.10)	**0.019*(0.22)**
Range	2.20 (0.69)	2.20 (0.80)	2.05 (0.78)	2.09 (0.78)	**0.043*(0.20)**

**Denotes that the mean difference is significant at the .05 level*

Table 8.6: Post hoc comparisons using Least Significant Differences

Post hoc comparisons using Least Significant Differences							
		Q1 vs Q2	**Q1 vs Q3**	**Q1 vs Q4**	**Q2 v Q3**	**Q2 v Q4**	**Q3 v Q4**
Posterior Accel	**Mean diff.**	-0.38	-0.59	-0.7	-0.22	-0.32	-0.11
	p value	0.053	0.057	**0.016**	0.338	0.07	0.662
	Upper bound	-0.01	0.01	0.02	0.03	0.03	0.06
	Lower bound	-0.08	0.12	-0.13	-0.68	-0.69	-0.42
AP Range	**Mean diff.**	-0.007	0.15	0.114	0.157	0.121	-0.36
	p value	0.918	0.059	0.187	**0.018**	0.096	0.36
	Upper bound	0.13	0.31	0.29	0.28	0.27	0.05
	Lower bound	-0.14	-0.01	-0.63	0.03	-0.24	-0.12

8.2.2 Changes to local jerk

A two factor repeated measures analysis of variance examined differences between left and right tibial jerk and between quartiles and for tibial jerk in all three directions. Analysis also checked for interaction effects between left and right sides and between quartiles. A repeated measures analysis of variance examined differences between jerk, across quartiles assessed, at both pelvic and thoracic sensor locations. For all significant main effects post hoc pairwise comparisons were performed. The results, for all jerk metrics, are presented below.

8.2.2.1 Changes to tibial jerk

No significant interactions were detected. Repeated measures ANOVAs were conducted to explore main effects of quartiles and side. However, as detailed in Table 8.5, there were no significant main effects. No mean differences exceeded the relevant SWC for any tibial jerk metric. Results are presented in Table 8.5.

Table 8.7: Changes to tibial jerk

Measure	Left Tibia Mean (SD)				Right Tibia Mean (SD)				Left v Right	Quartiles	Interaction
	Q.1	Q.2	Q.3	Q.4	Q.1	Q.2	Q.3	Q.4	p value (np^2)	p value (np^2)	p value (np^2)
Vertical											
Positive	492.1 (152.9)	509.8 (147.3)	500.5 (172.7)	506.8 (135.9)	467.9 (149.0)	484.9 (147.1)	511.8 (151.1)	492.41 (174.2)	0.593 (0.02)	0.617 (0.58)	0.706 (0.04)
Negative	491.1 (152.7)	483.9 (137.7)	499.6 (169.5)	516.3 (123.9)	476.6 (168.7)	478.8 (152.2)	508.3 (156.6)	497.1 (180.0)	0.825 (0.01)	0.364 (0.26)	0.601 (0.05)
Vertical Range	983.2 (253.8)	993.7 (246.1)	1000.2 (316.2)	1025.24 (243.2)	943.9 (290.2)	963.7 (274.2)	1020.1 (282.8)	989.5 (335.1)	0.697 (0.01)	0.403 (0.34)	0.61 (0.05)
Mediolateral											
Lateral	446.2 (238.4)	434.4 (255.9)	451.0 (234.4)	466.6 (252.2)	452.1 (201.0)	475.2 (224.7)	458.5 (222.7)	500.6 (275.4)	0.685 (0.03)	0.588 (0.05)	0.51 (0.06)
Medial	665.78 (479.3)	624.8 (402.6)	633.4 (399.8)	656.7 (409.5)	651.7 (399.6)	654.4 (414.6)	649.41 (418.8)	690.3 (464.9)	0.706 (0.01)	0.790 (0.03)	0.802 (0.25)
ML Range	1112.0 (707.9)	1059.2 (646.1)	1084.5 (623.4)	1123.3 (652.5)	1103.8 (577.8)	1129.6 (618.7)	1107.9 (709.5)	1193.0 (709.5)	0.652 (0.02)	0.716 (0.03)	0.688 (0.04)
Anteroposterior											
Anterior	451.7 (119.8)	508.3 (241.9)	477.6 (191.3)	461.5 (185.3)	464.9 (160.5)	459.5 (143.4)	455.5 (160.8)	458.7 (149.9)	0.581 (0.02)	0.874 (0.06)	0.263 (0.09)
Posterior	-433.4 (117.3)	-442.4 (126.4)	-430.9 (148.8)	-421.5 (99.6)	-449.4 (170.2)	439.1 (127.0)	435.3 (157.1)	430.6 (145.3)	0.855 (0.01)	0.920 (0.02)	0.645 (0.04)
AP Range	885.2 (196.1)	950.7 (327.2)	908.5 (278.4)	883.0 (236.2)	914.4 (295.3)	898.7 (236.5)	890.9 (273.1)	849.3 (339.3)	0.736 (0.01)	0.836 (0.02)	0.354 (0.08)

**Denotes that the mean difference is significant at the .05 level*

8.2.2.2 Changes to pelvic jerk

Analysis checked for interaction effects between quartiles. As no significant interactions were detected, repeated measures ANOVAs were conducted to explore main effects of quartiles. However, as detailed in Table 8.6, there were no significant main effects. No mean differences exceeded the relevant SWC for any pelvic jerk metrics. Results are presented in Table 8.6.

Table 8.8: Changes to pelvic jerk

Measure	Q.1	Q.2	Q.3	Q.4	p value (np^2)
Vertical					
Positive	289.4 (134.7)	292.6 (139.6)	309.6 (139.0)	309.7 (136.6)	0.608 (0.18)
Negative	332.0 (156.5)	328.2 (152.9)	335.0 (141.6)	301.1 (158.3)	0.232 (0.11)
Vertical Range	621.4 (287.7)	620.8 (288.6)	644.7 (271.2)	580.9 (325.0)	0.416 (0.60)
Mediolateral					
Lateral	187.3 (80.7)	196.5 (79.0)	208.5 (86.2)	197.9 (98.7)	0.480 (0.05)
Medial	138.2 (68.0)	151.0 (75.1)	159.4 (71.7)	145.3 (85.8)	0.326 (0.08)
ML Range	325.6 (130.7)	347.5 (140.7)	368.0 (138.0)	343.2 (165.8)	0.350 (0.08)
Anteroposterior					
Anterior	118.6 (36.8)	129.2 (48.9)	122.1 (45.4)	132.1 (52.5)	0.376 (0.07)
Posterior	135.6 (81.5)	137.6 (79.5)	128.7 (70.1)	124.3 (79.9)	0.401 (0.07)
AP Range	254.2 (111.3)	266.2 (121.1)	254.1 (108.2)	256.5 (124.0)	0.580 (0.04)

**Denotes that the mean difference is significant at the .05 level*

8.2.2.3 Changes to thoracic jerk

Analysis checked for interaction effects between quartiles. No significant interactions between quartiles were detected. Repeated measures ANOVAs were conducted to explore main effects of quartiles. No mean differences exceeded the relevant SWC for any thoracic jerk metrics. Results are presented in Table 8.7.

Table 8.9: Changes to thoracic jerk

Measure	Q.1	Q.2	Q.3	Q.4	p value (np^2)
Vertical					
Positive	199.9 (58.4)	226.3 (74.2)	226.3 (74.2)	217.3 (75.4)	0.148 (0.13)
Negative	228.0 (72.3)	224.9 (85.1)	230.3 (86.8)	232.1 (79.8)	0.710 (0.03)
Vertical Range	428.0 (124.1)	471.2 (153.6)	450.6 (161.9)	449.5 (150.0)	0.473 (0.06)
Mediolateral					
Lateral	58.7 (21.6)	66.7 (27.9)	70.8 (22.0)	62.6 (20.4)	0.390 (0.07)
Medial	94.7 (36.1)	100.8 (39.7)	107.0 (35.4)	95.5 (37.1)	0.449 (0.06)
ML Range	153.4 (52.2)	167.6 (65.2)	178.5 (52.7)	158.2 (54.1)	0.419 (0.07)
Anteroposterior					
Anterior	107.2 (47.4)	105.0 (47.4)	102.9 (43.1)	99.9 (47.7)	0.621 (0.03)
Posterior	89.1 (57.5)	94.8 (71.5)	89.1 (53.0)	87.2 (70.6)	0.757 (0.02)
AP Range	196.4 (102.6)	199.8 (118.7)	192.1 (92.9)	187.2 (115.6)	0.721 (0.02)

**Denotes that the mean difference is significant at the .05 level*

8.2.3 Changes to Shock Attenuation during the fatiguing run

Shock attenuation metrics, between tibias and pelvis (SA – P) and tibias and thorax (SA – T) were calculated, from the acceleration data obtained over 20 consecutive strides, using the formulas noted in chapter 7, section 7.2.3. Subsequent analysis revealed that SA - P decreased significantly between Quartiles 1 and 4 (p=0.021). Importantly, however, the difference in magnitudes between the SA – P for quartile 1 and quartile 4 did not exceed the SWC for this metric. Shock attenuation, assessed between tibia and thorax, did not exceed the relevant SWC across any quartile transitions (see Tables 8.10 and 8.11).

Table 8.10: Descriptive statistics

Measure	Quartile 1	Quartile 2	Quartile 3	Quartile 4
	SA Mean (SD)	SA Mean (SD)	SA Mean (SD)	SA Mean (SD)
Tibia to Pelvis	-81.95 (20.46)	-77.98 (17.70)	-79.22 (20.21)	-76.08 (21.04)
Tibia to Thorax	-55.23 (18.11)	-56.40(17.61)	-52.81 (17.63)	-52.50 (17.19)

Table 8.11: Results

Measure	Q.1 to 2 p value	Q.2 to 3 p value	Q.3 to 4 p value	Q.1 to 3 p value	Q.1 to 4 p value	Q.2 to 4 p value
Tibia to Pelvis	0.085	0.489	0.095	0.1	**0.021***	0.13
Tibia to Thorax	0.495	0.14	0.986	0.452	0.506	0.197

**Denotes that the mean difference is significant at the .05 level*

8.2.4 Changes to Resultant Accelerations during the Run to Fatigue

Resultant acceleration (RA) metrics were calculated using the formulas noted in chapter 7, section 7.2.4. All RA means were calculated using the acceleration data obtained from 20 consecutive strides. The change in RA experienced at the left tibia achieved significance between quartiles 3 and 4 (p=0.017), quartiles 1 and 4 (p=0.003), and quartiles 2 and 4 (p=0.017). Upon application of the relevant SWCs, each of these tibial resultant changes exceeded the relevant SWC (0.149 ms^{-2}) calculated for that metric (see Appendix H). There were no changes detected for the RAs experienced at the right tibia or at the pelvis. The change in RA experienced at the thorax achieved significance between quartiles 1 and 2 only (p=0.023). Please see Tables 8.12 and 8.13, below. Notably, this increase also exceeded the relevant SWC (0.176 ms^{-2}) for the thoracic RA metric.

Table 8.12: Descriptive statistics

Measure	Quartile 1 Mean (SD)	Quartile 2 Mean (SD)	Quartile 3 Mean (SD)	Quartile 4 Mean (SD)
Left tibia	9.32 (2.56)	9.287(2.39)	9.40 (2.41)	11.21 (1.89)
Right tibia	7.67 (2.65)	8.25 (2.48)	8.24 (2.40)	8.23 (1.89)
Pelvis	5.77 (1.43)	5.73 (1.45)	5.90 (1.54)	5.33 (4.32)
Thoracic	4.50 (1.34)	4.74 (1.60)	4.58 (1.57)	4.32 (1.87)

Table 8.13: Results

Measure	Q.1 to 2 p value	Q. 2 to 3 p value	Q.3 to 4 p value	Q.1 to 3 p value	Q1 to 4 p value	Q.2 to 4 p value
Left tibial Resultant	0.62	0.355	**0.017***	0.11	**0.003***	**0.017***
Right tibial Resultant	0.327	0.428	0.219	0.354	0.765	0.987
Pelvic Resultant	0.406	0.066	0.166	0.226	0.255	0.302
Thoracic Resultant	**0.023***	0.271	0.417	0.57	0.576	0.221

**Denotes that the mean difference is significant at the .05 level*

8.3 Results: Individual analysis of local accelerations and acceleration-derived metrics

Changes to all metrics, for each individual runner, across quartiles, were compared against the smallest worthwhile changes calculated for each sensor location and direction of movement. Subsequent results, for all metrics, are summarised in this section. The quartile means for each metric were also compared using paired sample t tests, these results are presented in Appendix K.

8.3.1 Individual changes to local accelerations during the Run to Fatigue

8.3.1.1 Individual acceleration changes at the left tibia

Left tibial accelerations exceeded the relevant SWCs, calculated for each direction, 195 times (48%) across the 3 quartile transitions. The greatest incidence of left tibial changes, 70 (51%), occurred medio-laterally and the least, 57 (42%), occurred vertically. Left tibial changes occurred most frequently (71 times) between quartiles 1 and 2, and least frequently between quartiles 3 and 4 (53 times). The left tibial metric demonstrating the highest incidence of meaningful change, as evaluated by comparison against the relevant SWC, was mediolateral acceleration range. This metric changed a total of 28 times across quartile transitions. Increasing on 11, and decreasing on 17, occasions. A summary of the results, for all acceleration metrics, are presented in the tables below.

Table Legend

🟩	Significant increases to the magnitude of positive (+) accelerations
🟧	No significant change
🟥	Significant increases to the magnitude of negative (-) accelerations/decelerations

Table 8.14 Left tibial changes exceeding the Smallest Worthwhile Change between Quartiles

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	5	1	7	3	2	0	6	5	5	34
No change	9	12	7	6	10	6	5	3	6	64
Negative increase	1	2	1	6	3	9	4	7	4	37
Quartile 2 to 3										
Positive increase	1	4	8	8	2	7	2	3	3	38
No change	11	7	3	5	9	5	9	5	10	64
Negative increase	3	4	4	2	4	3	4	7	2	33
Quartile 3 to 4										
Positive increase	4	1	6	4	2	4	3	3	3	30
No change	9	12	8	8	12	6	8	10	9	82
Negative increase	2	2	1	3	1	5	4	2	3	23
Left tibial changes: summary										
Positive increase	10	6	21	15	6	11	11	11	11	102
No change	29	31	18	19	31	17	22	18	25	210
Negative increase	6	8	6	11	8	17	12	16	9	93

8.3.1.2 Individual acceleration changes at the right tibia

Right tibial accelerations exceeded the relevant SWC 170 times (42%) across quartile transitions. The highest incidence of right tibial changes 66 (49%) occurred antero-posteriorly, and the least, 50 (37%), occurred medio-laterally. Right tibial changes occurred most frequently between quartiles 1 and 2 (60 times), and least frequently between quartiles 2 and 3 (53 times). The right tibial metric demonstrating the highest incidence of meaningful change, as evaluated by comparison against the relevant SWC, was posterior tibial deceleration. This metric changed a total of 25 times across quartile transitions. Increasing on 13, and decreasing on 12, occasions.

Table 8.15 Right tibial changes exceeding the Smallest Worthwhile Change between Quartiles

Direction of change	**Vertical**			**Mediolateral**			**Anteroposterior**			**Incidence of change**
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	4	2	8	9	1	0	7	3	4	38
No change	11	11	7	6	9	15	4	5	7	75
Negative increase	0	2	0	0	5	0	4	7	4	22
Quartile 2 to 3										
Positive increase	3	7	3	4	1	1	0	6	0	25
No change	11	5	5	8	13	13	11	8	8	82
Negative increase	1	3	7	3	1	1	4	1	7	28
Quartile 3 to 4										
Positive increase	4	2	4	4	3	6	1	4	1	29
No change	10	12	9	9	9	3	8	7	11	78
Negative increase	1	1	2	2	3	6	6	4	3	28
Right tibial changes: summary										
Positive increase	11	11	15	17	5	7	8	13	5	92
No change	32	28	21	23	31	31	23	20	26	235
Negative increase	2	6	9	5	9	7	14	12	14	78

8.3.1.4 Individual acceleration changes at the pelvis

Pelvic accelerations exceeded the relevant SWCs, on 151 (37%) occasions across quartile transitions. The highest incidence of pelvic changes, 71 (52.5%), occurred antero-posteriorly and the least, 50 (37%), occurred medio-laterally. Pelvic changes occurred most frequently between quartiles 3 and 4, 53 times, and least frequently, 47 times, between quartiles 1 and 2. The individual metric demonstrating the highest incidence of change, as evaluated by comparison against the relevant SWC, was decreasing posterior pelvic acceleration. This metric changed a total of 24 times across quartile transitions. All pelvic metrics variously increased and decreased across the quartiles of the fatiguing run

Table 8.16: Pelvic Accelerations changes greater than the SWC

Runner	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Accel	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	4	2	2	3	0	2	4	3	3	23
No change	7	12	13	10	14	12	7	10	8	92
Negative increase	4	1	0	2	1	1	4	2	4	19
Quartile 2 to 3										
Positive increase	5	0	6	1	0	3	4	1	5	25
No change	10	10	9	14	14	12	7	12	7	95
Negative increase	0	5	0	0	1	0	4	2	3	15
Quartile 3 to 4										
Positive increase	2	2	2	1	1	2	4	2	2	18
No change	11	11	11	13	13	11	8	11	11	100
Negative increase	1	1	1	0	0	1	2	1	1	8
Pelvic Changes: Summary										
Positive increase	11	4	10	5	1	7	12	6	10	66
No change	28	33	33	37	41	35	22	33	26	287
Negative increase	5	7	1	2	2	2	10	5	8	42

8.3.1.3 Individual acceleration changes at the thorax

Thoracic accelerations exceeded the relevant SWCs, for each direction, 143 times (35%) across quartile transitions. Most changes, 56 (41%), occurred vertically and the least, 39 (29%), occurred antero-posteriorly. The highest incidence of change occurred between quartiles 2 and 3 (53 times), and the least, 44 times, between quartiles 1 and 2. The metric demonstrating the highest incidence of meaningful change was lateral acceleration. This metric changed a total of 29 times across quartile transitions. Increasing on 12, and decreasing on 9, occasions. All thoracic metrics increased and decreased across the quartiles of the fatiguing run.

Table 8.17: Thoracic Accelerations changes greater than the SWC

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	1	1	6	4	3	3	1	2	1	22
No change	11	8	7	8	10	10	14	13	10	91
Negative increase	3	6	2	3	2	2	0	0	4	22
Quartile 2 to 3										
Positive increase	2	0	4	5	3	6	4	2	3	29
No change	12	14	10	6	8	5	8	8	11	82
Negative increase	1	1	1	4	4	4	3	5	1	24
Quartile 3 to 4										
Positive increase	0	7	3	3	4	1	1	3	2	24
No change	12	5	8	10	11	11	11	9	12	89
Negative increase	3	3	4	2	0	3	3	3	1	22
Thoracic changes: summary										
Positive increase	3	8	13	12	10	10	6	7	6	75
No change	35	27	25	24	29	26	33	30	33	262
Negative increase	7	10	7	9	6	9	6	8	6	68

8.3.2 Individual changes to local jerk during the Run to Fatigue

The changes to all jerk metrics for each individual runner were compared against the smallest worthwhile changes calculated for each sensor location and plane of movement. Subsequent results, for all metrics, are summarised in this section. The quartile means for each metric were also compared using paired sample t tests, these results are presented in Appendix L.

8.3.2.1 Individual change to tibial jerk metrics

Left tibial jerk exceeded the relevant SWCs, calculated for each tibial direction, on 21 occasions across the 3 quartile transitions. This number represents only 4% of the total number of quartile transitions evaluated across runners. Similarly, at the right tibia, jerk measures exceeded the relevant SWCs, calculated for each tibial direction, on 4 occasions across the 3 quartile transitions. This number represents less than 1% of the total number of quartile transitions evaluated across runners. At the pelvis, jerk measures exceeded the relevant SWCs, calculated for each pelvic direction, on 67 occasions across the 3 quartile transitions, representing 12% of all possible opportunities for meaningful change. At the thorax, jerk measures exceeded the relevant SWCs, calculated for each pelvic direction, on 53 occasions across the 3 quartile transitions, representing less than 10% of all possible opportunities for meaningful change. The individual jerk metric demonstrating the highest incidence of change, as evaluated by comparison against the relevant SWC, was an increasing vertical thoracic range. This metric changed a total of 13 times across quartile transitions.

In total, only 7% of all local jerk metrics exceeded the relevant SWCs across the 4 sensor locations. This low incidence of change was unexpected. As comparison, when changes to local jerk metrics across quartiles were tested for significance, approximately 25% of all inter-quartile mean differences were considered significant.

8.3.3 Individual changes to Shock Attenuation during the Run to Fatigue

Smallest worthwhile changes were calculated for the SA metrics between tibias and pelvis and between tibias and thorax. The results of these calculations are presented in Appendix H.

8.3.3.1 Shock Attenuation between tibia and pelvis (SA – P) and tibia and thorax (SA – T)

The runner-specific differences between SA – Pelvis means, across quartile transitions, exceeded the relevant SWC on only 3 occasions. The differences between SA – Thoracic means, however, exceeded the relevant SWC on 10 occasions across the quartile transitions (for illustrative chart, see Appendix M). Accordingly, determining change using SWCs elicited only 13 incidences of change. In contrast, when significant change was evaluated using paired sample t tests, SA – P and SA - T changed 44 times each across the quartile transitions (See Appendix N).

8.3.4 Individual changes to Resultant Accelerations during the Run to Fatigue

Smallest worthwhile changes were calculated for the RA metrics at the tibias, pelvis and thorax. The results of these calculations are presented in Appendix H.

8.3.4.1 Tibial Resultant Accelerations

Changes to left tibial RAs exceeded the relevant SWC on 30 occasions across 44 quartile transitions (68%). The highest incidence of meaningful change occurred between quartiles 3 and 4 (11), followed by between quartiles 2 and 3 (10), and quartiles 1 and 2 (8). Of these changes, 22 resulted in increases, and 8 resulted in decreases, to the magnitudes of left tibial resultant accelerations.

Changes in right tibial RAs exceeded the SWC on 31 occasions across 44 quartile transitions (70%). The highest incidence of meaningful change occurred between quartiles 1 and 2 (11), followed by both quartiles 2 and 3 (10) and quartiles 3 and 4 (10). Of these changes, 23 resulted in increases, and 8 resulted in decreases, to the magnitudes of right tibial resultant accelerations. An overview of these results are presented in Table 8.18.

Table 8.18: Incidence of inter-quartile changes greater than the Smallest worthwhile Change to tibial resultant accelerations

	Left Tibial Resultant			Right Tibial Resultant		
Runner	**Q.1 To 2**	**Q.2 To 3**	**Q.3 To 4**	**Q.1 To 2**	**Q.2 To 3**	**Q.3 To 4**
1	🟩	🟥	🟩	🟩	🟥	🟧
2	🟥	🟧	🟥	🟩	🟩	🟥
3	🟥	🟩	🟧	🟩	🟧	🟩
4	🟩	🟩	🟩	🟧	🟩	🟧
5	🟧	🟩	🟩	🟩	🟩	🟧
6	🟧	🟩	🟩	🟩	🟧	🟩
7	🟧	🟥	🟩	🟩	🟥	🟩
8	🟧	🟩	🟧	🟧	🟧	🟥
9	🟥	🟧	🟩	🟥	🟧	🟩
10	🟧	🟩	🟩	🟥	🟧	🟩
11	🟩	🟩	⬛	🟩	🟧	⬛
12	🟥	🟩	🟧	🟧	🟩	🟩
13	🟧	🟧	🟥	🟩	🟥	🟩
14	🟩	🟧	🟩	🟩	🟥	🟩
15	🟧	🟩	🟩	🟩	🟩	🟧
Positive Increase	4	9	9	10	5	8
No change	7	4	3	3	6	4
Negative Increase	4	2	2	2	4	2

8.3.4.2 Pelvic and thoracic Resultant Accelerations

Changes in pelvic RAs exceeded the SWC on 19 occasions across 44 quartile transitions (42%). The highest incidence of meaningful change occurred between quartiles 3 and 4 (10), followed by quartiles 2 to 3 (6) and quartiles 1 to 2 (4). Of these changes, 4 resulted in increases, and 15 resulted in decreases, to the magnitudes of pelvic resultant accelerations.

At the thorax, changes in RAs exceeded the SWC on 24 occasions across 44 quartile transitions (58%). The highest incidence of meaningful change occurred between quartiles 1 and 2 (10), followed by quartiles 2 to 3 (9) and quartiles 2 and 4 (6). Of these changes, 15 resulted in increases, and 9 resulted in decreases, to the magnitudes of thoracic resultant accelerations.

Table 8.19: Incidence of inter-quartile changes greater than the Smallest worthwhile Change to pelvic and thoracic resultant accelerations

	Pelvic Resultant			**Thoracic Resultant**		
Runner	**1 To 2**	**2 To 3**	**3 To 4**	**1 To 2**	**2 To 3**	**3 To 4**
1	🟩	🟩	🟥	🟩	🟥	🟥
2	🟨	🟨	🟥	🟩	🟥	🟥
3	🟨	🟨	🟨	🟩	🟥	🟨
4	🟨	🟥	🟨	🟨	🟥	🟥
5	🟨	🟥	🟥	🟥	🟥	🟨
6	🟨	🟨	🟨	🟨	🟨	🟩
7	🟨	🟨	🟨	🟩	🟥	🟨
8	🟥	🟩	🟥	🟩	🟨	🟨
9	🟨	🟥	🟥	🟨	🟨	🟨
10	🟨	🟨	🟥	🟩	🟨	🟨
11	🟨	🟩	⬛	🟩	🟨	⬛
12	🟨	🟨	🟥	🟩	🟩	🟨
13	🟥	🟨	🟥	🟨	🟩	🟩
14	🟥	🟨	🟨	🟩	🟩	🟩
15	🟨	🟨	🟥	🟨	🟨	🟨
Positive Increase	1	3	0	9	3	3
No change	11	9	5	5	6	8
Negative Increase	3	3	10	1	6	3

8.3.5 Individual changes to subjective ratings of Perceived Exertion during the Run-to fatigue

As detailed in chapter 6, every 2 minutes, and during the last 30 seconds of the fatiguing run, runners were asked to indicate perceived exertion ratings on a CR10 scale. Each runner's rating of perceived exertion, at the end of each quartile, is documented in Table 8.26, below. The range of RPE change over the course of the fatiguing run varied from 2.5 units, as runner 9 moved from an initial minimum rating of 6 to a maximum of 8.5, to 7 units for runner 13 who moved from an initial minimum rating of 2 to a maximum rating of 9. The average magnitude of RPE change across the fatiguing run, per runner, was 4.7.

Notably, the greatest increase in RPE units occurred between quartiles 1 and 2 (28.5) and progressively decreased across Quartiles 2 and 3 (25), and Quartiles 3 and 4 (16.5). This observation, however, does not suggest that more exertion was expended between quartiles 1 and 2 than between later quartiles. This observation merely reflects the fact that changes in exertion increased by more RPE units in the early phases of the fatiguing run. This observation likely reflects the fact that the initial ratings provided by some participants, in quartile 1, were quite low and subsequently increased, in some instances, by relatively large increments, between quartiles 1 and 2. To illustrate, 7 runners increased exertion ratings by at least 2.5 units between quartiles 1 and 2. In contrast, the largest change between quartiles 3 and 4 was 2 units, registered by only one runner. As perceptual ratings scale to anticipated end-points, rates of increase in perceived exertion are unlikely to be either linear or constant across changing conditions (for example: Eston, 2012). Borg considered such non-linear, somewhat positively accelerating growth functions, to be appropriate when determining intensity of experience (Borg, 2010).

Table 8.20. Runners Ratings of Perceived Exertion per Quartile

Runner	RPE at end of Q1	RPE at end of Q2	RPE at end of Q3	RPE at end of Q4
1	5	6	7	8.5
2	5	6	8	9.5
3	6	7	8	9
4	4	6.5	7.5	9
5	5	6.5	7.5	9
6	3.5	5	9	9
7	3.5	5	7.5	8.5
8	5.5	6.5	8	9
9	6	7.5	8	8.5
10	4	6	7.5	9
11	4	7	8	9
12	4	7	8	9
13	2	4	7	9
14	3	6	8	9
15	4	7	9	9.5
Average RPE per Quartile	4.3	6.2	7.9	9
Total RPE per Quartile	64.5	93	118	134.5

Table 8.21: Changes to runner's ratings of perceived exertion across quartile transitions

Runner	RPE change Q.1 to 2	RPE change Q.2 to 3	RPE change Q.3 to 4	Total RPE change
1	1	1	1.5	3.5
2	1	2	1.5	4.5
3	1	1	1	3
4	2.5	1	1.5	5
5	1.5	1	1.5	4
6	1.5	4	0	5.5
7	1.5	2.5	1	5
8	1	1.5	1	3.5
9	1.5	0.5	0.5	2.5
10	2	1.5	1.5	5
11	3	1	1	5
12	3	1	1	5
13	2	3	2	7
14	3	2	1	6
15	3	2	0.5	5.5
Total unit change across transitions	28.5	25	16.5	70
Mean change per Quartile	1.9	1.7	1.1	4.7
Average RPE per Quartile	4.3	6.2	7.9	9

. 8.4. Collective observations: How metrics changed over the course of the fatiguing run

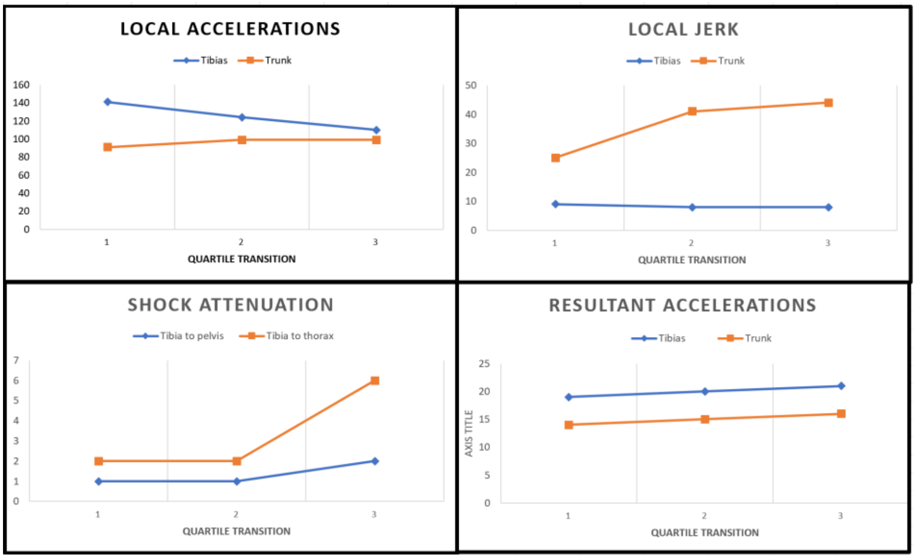
To contextualise how the various metrics changed over the course of the fatiguing run, Tables 8.22 and 8.2.3, below, were constructed. Notably, there does not appear to be any clear, coherent or shared patterns of change across the different metrics. The graphical illustration, Table 8.2.3, underlines the observation that there is no readily apparent, coherent agreement or common trend between the various metrics. Given that each of these metrics have been advocated as measures which sensitively change, subsequent to perturbations, such as injury and fatigue, as discussed in chapters 4 and 5, the lack of agreement between metrics is unexpected. However, although this observation is interesting, and potentially important, given the experimental constraints this finding clearly requires further investigation and replication.

As a final observation, patterns of change, evaluated using p values, provided different findings than those presented here, highlighting the extent of the potential discrepancies inherent, within this experimental realm, when different metrics and different methods of analysis are employed to investigate the same phenomenon (see Appendix O for comparison)

Table 8.22: Inter-measure comparisons across quartile transitions

Measure		Changes between Q1 & Q2	Changes between Q2 & Q3	Changes between Q3 & Q4
Local Accelerations	Tibias	141	124	110
	Trunk	91	99	99
Local Jerk	Tibias	9	8	8
	Trunk	25	41	44
Shock Attenuation	Tib to pel	1	1	2
	Tib to thor	2	2	6
Resultant Accelerations	Tibias	19	20	21
	Trunk	14	15	16
RPE unit increases per quartile transition		28.5	25	16.5

Table 8.23 Side by side comparisons of the trajectory of change between metrics



8.5 Conclusions

The objectives of this chapter were threefold:

1. To evaluate whether fatigue-induced changed uniformly across acceleration-derived metrics over the course of a moderately fatiguing run
2. To assess whether local accelerations and acceleration-derived metrics change in a consistent manner across measures, and across attachment locations, over the course of a fatiguing run
3. To examine whether, over the course of the fatiguing run, the patterns of change experienced across a cohort of runners, or is experienced in an individually idiosyncratic, runner-specific manner

The core findings of the analysis presented in this chapter suggest:

i. Incidence of group change, across selected metrics, was inconsistent. For example, a number of acceleration measures and one resultant acceleration measure, yet no local jerk or shock attenuation measures, exceeded the relevant SWC
ii. Both group and individual analyses suggest that left and right tibias commonly experience differences in the magnitudes of imposed acceleration shocks
iii. In contrast to the individual analysis of vertical tibial decelerations contained in Chapter 7, in this chapter, vertical tibial shocks were found to variously

increase and decrease over the course of the quartile transitions, i.e., vertical tibial deceleration did not demonstrate any directional consistency

iv. The various metrics, each conventionally assumed to provide indications of increasing fatigue, did not provide consistent and comparable insights across runners. There was no clear trajectory of common change shared amongst the various acceleration-derived metrics

v. Each runner exhibited a unique pattern of meaningful changes across metrics, thereby suggesting that individual compensatory responses, to running-induced exertion, are both commonplace and highly personalised

The implications of these findings will be further explored in chapter 9.

Chapter 9:

Discussion

9.1 Introduction

This thesis began by noting how remarkably proficient we are as endurance runners. We learn to run naturally, as a by-product of normal development. Yet the acquisition of running mastery requires extensive practice, over prolonged periods, and demands the gradual harnessing of densely networked neural and biological tissues collectives. However, despite our remarkable running proficiency, we are vulnerable to breakdown and frequently suffer injury and/or running-related discomfort. Predominantly, these occurrences are overuse in nature and, as noted in chapters 2 through 5, are exacerbated by fatigue-induced deteriorations to coordination serving to reduce our capacity to manage repetitive impact shocks.

Our understanding of running coordination, and how running coordination deteriorates, is limited by a lack of insightful assessment tools capable of providing a window into the deeper nature of the phenomenon. As running coordination is a multi-faceted phenomenon, it can be legitimately examined through a diversity of investigative lens. Currently, however, it remains unclear which measurable facets of running coordination are the most relevant to running proficiency, which are the most indicative of running deterioration and dysfunction, and whether different tools, advocated as assessing the same phenomenon, i.e., fatigue-induced detriments to running coordination, actually do so. Consequently, running coordination remains a somewhat equivocal and vaguely defined topic lacking a coherent and clarifying explanatory framework. Similarly, the outcomes of experimental studies, investigating running coordination and how it may change subsequent to fatigue, are frequently conflicting. This ambiguity recently led Sheerin and colleagues (2018) to conclude that, due to confounding influences –such as the use of different fatiguing protocols, assessment technologies and different running populations–, there is no clear consensus on how global and/or localized fatigue influences runners capacity to effectively manage the repetitive impact shocks imposed during running (Sheerin et al, 2018). Given this backdrop, the core purpose of this thesis was to investigate the complex phenomenon of running coordination, assessed through a battery of surrogate measures, changes subsequent to fatigue.

Specifically, the core experimental objectives of this thesis were three-fold:

- To examine the impact of running-induced fatigue on various running metrics commonly cited as sensitive to fatigue
- To examine and compare how these metrics change over the course of a moderately fatiguing run

- To determine, through analysis and comparison of runner-specific data, whether fatigue-induced deteriorations to running coordination progress in a coherent and shared, or disparate and individually customised, manner across the individuals of the running cohort

The objective of this discussion is to integrate the relevant conceptual conclusions emanating from the literature review chapters with the key findings and insights emerging from the empirical investigations, and to contextualise the combined learning outcomes against the stated thesis objectives.

9.2 Theoretical and conceptual thesis outcomes

The core objectives of the review chapters, 2 through 5, were to examine current understanding of key, conventionally under-developed, concepts relating to the larger running coordination phenomenon. Accordingly, Chapters 2 through 5 delved into the embedded processes, mechanisms and constraints underpinning running coordination, and in so doing reviewed the relevant evidence encompassed within that broad remit.

9.2.1 Chapter 2 – The uniqueness of human running coordination

This chapter outlined the deep neurobiological underpinnings of running coordination, and in so doing explored the multiple evolutionary innovations that enable humans to run in such an unlikely, and superficially risky, locomotive gait. Also described are the various processes through which continued practice progressively embeds running coordination habits within neural and peripheral tissue structures.

A persistent theme permeating this chapter is that our capacity to run, in such a seemingly dangerous manner, is founded on a platform of complexly entwined, evolutionary bestowed, neurobiological innovations. The coordinated integration of these modern and ancient evolutionary innovations blends the outputs of neuronal top-down and mechanical bottom-up control processes. Ultimately, it is this distributed multi-level control that enables the human runner to robustly negotiate varied challenges and terrains, for an acceptable neuronal investment; an acceptable energetic cost, and an acceptable exposure to survival threatening de-stabilisations. A key point, rationalised within this chapter, is that running coordination is vulnerable to any perturbation that impedes the flow, clarity or interpretation of sensorimotor information. An implication of this rationalisation is that coordination is subject to change under the influence of multiple mediating factors, operating across various timescales.

9.2.2 Chapter 3 – The robust running ape

This chapter begins by observing that, although we are amongst Nature's most proficient runners, we are, as evidenced by available running injury data, nevertheless, susceptible to breakdown. The chapter describes the neurobiological processes through which we learn to run, and the various plastic neural and biological mechanisms serving to engrain our running coordination habits and proclivities. Also discussed are the robustness-bestowing organisational principles of modularity and degeneracy. Additionally, described within this chapter, are the mechanisms through which accumulating fatigue diminishes complexity; diminishes available movement degeneracies; modulates variability and subsequently drives increasing dysfunction.

As the chapter closes, I outline the evolutionary neuro-economics of running coordination: the driving forces which initially serve to embed efficient habits, but eventually encase us within restrictive, and ultimately dysfunctional, constraints. The practical relevance of the reviewed evidence, to running practice, is subsequently summarised. Finally, the chapter ends with a description of running coordination, emanating from the literature reviewed, as the overarching super-capacity which ultimately orchestrates how proficiently neural, muscular, cardiovascular and metabolic reserves are purposefully harnessed, or wastefully squandered. The definition of running coordination, offered in this chapter, summarises the interpretation and meaning of the term 'running coordination' used within the context of this thesis.

9.2.3 Chapter 4 – Running Stiffness: Evolutions simple 'satisficing' solution

This chapter focusses specifically on the intertwined concepts of running stiffness, impact shock and shock attenuation. During the ground contact event, the suddenly imposed shock of impact drives a diversity of multi-dimensional 'threats'; yet simultaneously offers momentum-conserving and energy-saving opportunities. The subsequent coordination challenge, presented to the runner, is how to self-organise so as to effectively negotiate the trade-off between moderating risks and promoting proficiency. Runners negotiate this trade-off by calibrating the stiffness of the running system to alternatively mitigate the risks posed by excessive shock loading, whilst productively redirecting and recycling the energy and momentum conserving opportunities provided by shock loading. Appropriate management of running stiffness enables runners to calibrate the background tensional integrity of the running body to the dynamics of the impact event. From this perspective, the skilled modulation of running stiffness is a strategy that is both elegantly efficient, and evolutionarily economical. However, as with all facets of movement coordination, running stiffness and shock management capacities are vulnerable to diminishing capacities, driven, for example by injuries, biological *wear and tear* or accumulating fatigue.

Also, in this chapter, research relating to the threat imposed by ground contact is contextualised against the explanatory framework of the stiffness concept.

Research relating to running-imposed, predominantly tibial, accelerations is also outlined, against both performance and injury contexts.

Finally, the chapter closes by suggesting that the evidence reviewed illustrates that the management of impact shock loadings –via the calibration of running stiffness– is a practice-enhanced, coordinated skill, which ultimately enables the productive deployment of available neural and biological resources to simultaneously manage risk, and enhance running proficiency. Running stiffness, accordingly, emerges as a coordinated solution to the challenges and opportunities implicit in the ground contact event.

9.2.4 Chapter 5 – Smoothness: an unexplored window into coordinated running proficiency

Within this chapter, a diverse body of literature is reviewed through the lens of movement smoothness. Conventionally, movement smoothness is assessed via measurement of its opposite, kinematic jerk, where jerk is formally defined as the rate of change in acceleration. In effect, smoothness is the converse of jerk. Smoothness increases as jerk reduces, and vice versa. Smoothness emerges as a natural coordinated outcome of the intimate integration between multi-level control systems collaboratively combining to sensitively detect, and agilely respond to, imposed perturbations in ways that minimise disruption to rhythmical locomotion. Accordingly, smoothness is an indication of sensitive and fine-grained movement control. Here, the evidence underpinning why smoothness may be a universal indicator of coordinated movement proficiency is outlined.

Despite its apparent relevance to a variety of movement outcomes, however, smoothness remains only vaguely defined within the sports science literature. Accordingly, a working definition of running smoothness is offered to help better frame this discussion. Notably, the available evidence illustrates that a variety of influences, all serving to inhibit different coordination qualities, ultimately drive the progressive erosion of smoothness. Once again, the evidence suggests that accumulating fatigue plays an influential role in detracting from movement smoothness, and thereby elevating jerk. As ever, the progressive sensorimotor *wear and tear* associated with declining health, injury, pain sensitisation and fatigue status, compounded by inevitable cycles of overuse, underuse, misuse and disuse, all combine to escalate jerk and diminish smoothness.

Finally, although evidence suggests smoothness metrics provide insights into coordination proficiency, the most appropriate means and methods to measure, monitor and analyse smoothness remain unresolved and controversial (see, for example, Balasubramanian et al, 2015; Guilde et al, 2018).

9.2.5 Key outcomes of the literature review

i. Running coordination can be conceptualised through a variety of theoretical lens and quantified using a broad diversity of assessment measures. Nevertheless, it remains unclear which measure, or measures, provides the most pertinent information, or how these measures correspond, or fail to correspond, with each other
ii. Although certain measures, –such as local accelerations, jerk, measures of shock management (i.e., local stiffness and shock attenuation) and resultant accelerations– are all theorised to be vulnerable to fatigue-driven deterioration, only elevated peak vertical tibial decelerations are consistently associated with negative outcomes
iii. A recurring theme, emanating from the literature review, was that running coordination is an innately personalised phenomenon, shaped by a host of runner-specific influences. In endeavouring to illuminate the extent of this phenomenon, individual analyses, across a range of metrics, were conducted for each runner and compared across the cohort.

9.3 Key Experimental Outcomes

In the following sections the experimental findings, for pre- and post-fatigue runs and for the fatiguing run, are blended, reviewed and the key outcomes outlined.

9.3.1 Local Accelerations

9.3.1.1 Tibial accelerations

Peak tibial shock —the highest acceleration recorded at the tibia during stance— is a common means of assessing the mechanical stress of ground contact (Milner et al, 2006; Sheerin et al, 2018). Although local peak accelerations are a surrogate loading measure, previous work suggests tibial shock provides a more direct and informative estimate of the mechanical stress exerted on the tibia than ground reaction forces (GRFs) (Milner et al, 2006). Prior evidence has linked increasing vertical tibial shocks to an increasing incidence of tibial fatigue fractures in runners (Milner et al, 2006; Pohl et al, 2008). Supporting this perspective, previous investigations suggested that greater vertical shocks were more strongly associated with increased running-related injury (RRI) risks than peak vertical forces (Van Gent et al, 2007; Liberman et al, 2010), and suggest that runners who habitually experience greater vertical shocks are more vulnerable to overuse injury; subsequently leading others to conclude that shock loading is the single most informative indicator of RRI probability (Davis, Bowser and Mullineaux, 2016; Zifchock et al, 2008; Hreljac, 2004).

Nevertheless, the relationships between peak tibial accelerations and running performance remain unclear and under-explored (Zifchock et al, 2008). Notably, and relevantly, investigations evaluating changes in tibial shock over the course of fatiguing runs have reported mixed results. Abt and colleagues (2011), for example, reported no changes in any kinematic or acceleration-dependent variables subsequent to exhaustive treadmill running (Abt et al, 2011). In contrast, other studies concluded that tibial shocks did increase with treadmill running-induced fatigue (Mizrahi et al, 2000); whereas findings were unclear when runners ran both overground and on a treadmill (García-Pérez et al, 2014). These findings highlight existing controversies relating to whether and how fatigue changes locally experienced shock accelerations.

Although the predominance of prior work assessed vertical accelerations using uni-axial sensors fixed to the anteromedial portion of the tibia, recent work makes it increasingly evident that fatigue does not drive local acceleration changes exclusively in the sagittal, but also across mediolateral and anteroposterior, planes. The incidence, relevance, and importance of potential changes to these horizontal accelerations, however, remains unclear (for example: Giandolini et al, 2016).

The capacity to calibrate trunk behaviour to imposed movement demands, is an outcome of effective coordination and as such is vulnerable to fatigue-induced deterioration. Hence recent research suggests that fatigued running not only effects the impact shock experienced by the lower body, but also changes the magnitudes of accelerations experienced at the upper-and lower-trunk (Schutte et al., 2018; Möhler, Ringhof, Debertin, & Stein, 2019; Möhler, Marahrens, Ringhof, Mikut & Stein, 2020).

Accordingly, for example, Koblbauer and colleagues (2013), investigating novice runners, demonstrated that runners compensated for running-induced fatigue by changing aspects of trunk control (Koblbauer et al., 2013). Similarly, Morin and colleagues (2005) showed that, during prolonged exhaustive running, the vertical motion of the trunk reduced in tandem with increases in leg stiffness and decreases in shock attenuation as fatigue accumulated (Morin et al., 2005). As a final example, Koblbauer and colleagues (2014) also observed changes in trunk control following running-induced fatigue, further demonstrating that the coordinated control of the upper-body is vulnerable to fatigue driven change. Nevertheless, studies evaluating changes to pelvic, and especially thoracic, acceleration behaviours during running activities remain sparse.

9.3.1.1. Vertical tibial accelerations

In agreement with some, but not all, of the reviewed research, the magnitude of peak vertical tibial decelerations exceeded the smallest worthwhile change, for that metric, between pre- and post-fatigue conditions. Critically, however, although vertical decelerations increased between pre- and post-fatigue conditions, no such changes were evident across the quartiles of the fatiguing run. This finding suggests that, although there was a meaningful increase in peak vertical tibial deceleration before and after the fatiguing run, there was no meaningful increase in peak vertical tibial deceleration during the fatiguing run. Conversely, over the course of the fatiguing run, the magnitudes of peak vertical tibial accelerations meaningfully increased, yet the magnitudes of peak vertical tibial accelerations between the pre- and post-fatigue intervals did not. The range of vertical accelerations experienced at the tibias, however, meaningfully increased between both pre- and post-fatigue conditions and across the quartiles of the continuous, fatiguing run. These discrepancies in results, between subtly differing experimental conditions, underline the difficulties inherent in constructing a coherent explanatory framework encompassing the conflicting results characterising this field of study.

Analysis of individual data revealed that all vertical metrics demonstrated multiple instances of meaningful change between pre- and post-conditions. In the case of both vertical acceleration and vertical range metrics, these changes variously resulted in increasing or decreasing acceleration magnitudes. Notably, however, in the case of vertical tibial deceleration, all changes occurred exclusively in one direction. Specifically, every meaningful change to vertical tibial decelerations, of which there were 10 between both tibias, resulted in an increase in the magnitude of deceleration shock. Vertical tibial deceleration was therefore unique amongst the tibial metrics, as every other metric variously exhibited both increases and decreases in acceleration magnitudes between conditions.

Individual analysis of runners data over the course of the fatiguing run, however, did not support this directional consistency. Instead, during the fatiguing run there were 31 incidences of meaningful change to vertical tibial decelerations. Of these changes 17 resulted in increases, whilst 14 resulted in decreases. Accordingly, if only pre-to post-fatigue data are considered, it appears vertical decelerations increase, in a uni-directional manner, as fatigue progresses. Nevertheless, analysis of the quartile data illustrates that, under the marginally different conditions of the fatiguing run, vertical tibial decelerations variously increase and decrease (An observation highlighting the difficulties inherent in comparing the results of differently designed investigative studies within this domain).

9.3.1.2 Mediolateral tibial accelerations

Group analyses of tibial mediolateral (ML) acceleration data, collected during both the pre- and post-fatigue intervals and across the quartiles of the fatiguing run, did not reveal any meaningful changes between conditions. Individual analysis of pre- and post-fatigue data, however, highlighted multiple instances of runner-specific change, to ML metrics. Notably, however, ML tibial accelerations variously increased and/or decreased without any consistent direction of change.

Interestingly, between pre- and post-fatigue conditions, 2 runners simultaneously increased the magnitude of medial decelerations experienced at the right tibia, while decreasing the magnitude of medial decelerations experienced at the left tibia; thereby highlighting that left and right legs can be subjected to different acceleration experiences during running. However, beyond the observation that ML metrics shared the same general trends as vertical tibial accelerations, in that most changes occurred between the 1st and 2nd quartile and that there were more incidences of change than instances of no change, there were no clear, discernible patterns to this change.

9.3.1.3 Anteroposterior tibial accelerations

Neither grouped analysis of the pre- and post-fatigue, nor of the fatiguing run, detected any change in anteroposterior (AP) metrics. Again, however, analysis of individual data, from the pre- and post-fatigue intervals and the fatiguing runs, highlighted multiple instances of AP change. Collectively, the AP metrics followed the same trends as other local tibial accelerations in that meaningful changes variously resulted in increases or decreases in acceleration magnitudes and that most meaningful changes occurred between quartiles 1 and 2, with the least number of changes occurring between quartiles 3 and 4. Furthermore, once again, incidences of meaningful change were more frequent than instances of no change.

9.3.1.4 Differential acceleration experiences between tibias

Remarkably little running-specific research has investigated the potential differences between the acceleration loading experiences of left and right tibias. Yet, notably, in the pre- and post-fatigue analysis, documented in chapter 7, the difference between the vertical deceleration means experienced by left and right tibias exceeded the smallest worthwhile change, with tibial decelerations experienced by the left tibias being greater than those experienced by the right. Similarly, analysis of the quartile data from the fatiguing run, documented in chapter 8, established that the vertical decelerations experienced by left and right tibias meaningfully differed, as did the vertical accelerations experienced by both tibias. Furthermore, in a previously unreported finding, although the vertical accelerations experienced by the right tibias were greater than those experienced by the left, the magnitude of decelerations experienced by the left were greater than those experienced by the right.

Continuing this theme, within both the pre- and post-fatigue intervals and the fatiguing run analyses, the differences between the magnitudes of peak medial decelerations experienced by left and right tibias exceeded the relevant SWC. Similarly, the difference between peak lateral accelerations, between tibias, also exceeded the relevant SWC. Specifically, medial decelerations experienced at the left tibia were higher than those experienced at the right. In contrast, the magnitudes of lateral accelerations experienced at the right tibia were higher than those experienced at the left. These findings held across both analyses. Finally, the peak posterior decelerations**,** experienced by left and right tibias during the pre- to post-fatigue intervals, also meaningfully differed, with the left tibia experiencing higher peak decelerations.

Notably, upon inspection of the patterns of change for individual runners, the different tibial accelerations observed for each runner were such that no runner experienced the same permutation of meaningful changes at the left and right tibias. This observation held for the pre- to post-fatigue interval runs, and across the quartile transitions of the fatiguing run. These observations suggest that differential changes to left and right tibial accelerations are both pervasive and, perhaps, peculiar to the specific runner.

As an additional noteworthy observation, across the quartiles of the fatiguing run, tibial acceleration metrics followed two trends. Firstly, most meaningful changes occurred between the 1st and 2nd quartiles, whereas the least number of changes occurred between quartiles 3 and 4 and, secondly, incidences of meaningful change were more frequent than instances of no change.

9.3.2 Pelvic accelerations

The group analysis, detailed in chapter 7, revealed that the magnitude of pre- and post-condition pelvic medial and mediolateral range acceleration metrics exceeded the relevant SWCs for those measures. Over the course of the fatiguing run, however, no differences between quartile means exceeded pelvic SWCs.

Analysis of individual runner's data illustrated that, for both pre- and post-interval and the fatiguing run, there were substantial incidences of meaningful change, across all pelvic metrics. Furthermore, for every pelvic metric meaningful change variously resulted in increases or decreases to acceleration magnitudes, i.e., there were no discernible trends of either predominant increase or decrease.

9.3.3 Thoracic accelerations

Group analysis of thoracic acceleration data did not detect any meaningful changes between pre- and post-fatigue condition means. Over the course of the fatiguing run, however, the differences between quartile means exceeded the relevant SWC on two occasions. Specifically, posterior thoracic deceleration increased between quartiles 1 and 2, and the range of anteroposterior accelerations increased between quartiles 2 and 3.

As with other sensor locations, over the course of the fatiguing run, there were multiple incidences of quartile differences exceeding the relevant SWCs at the thorax. Again, incidences of meaningful change were more frequent than incidences of no change. Interestingly, although in the case of tibial accelerations incidences of meaningful change were most common between quartiles 1 and 2, in the case of upper-body sensor locations, pelvic and thoracic accelerations changes were least frequent across the first quartile transition, and most frequent later in the fatiguing run.

9.3.2 Local Jerk

A growing body of literature, reviewed in chapter 5, suggests that movement smoothness, as evaluated by quantifying its opposite, jerk, is a sensitive indicator of movement coordination proficiency. Extensive research reviewed in chapter 5 suggests jerk is an informative and sensitive measure capable of detecting changes in coordinated movement control. As example, although not running specific, a recent study found that a lack of smoothness, within the postural sway profiles of NCAA Division 1 College football players, served as a predictor of subsequent injury (Wilkerson, Gupta, and Colston, 2018). Specifically in relation to running, although previous research established differences in the magnitudes of peak jerk experienced at the lower leg between regular and novice runners, measurements were calculated indirectly using video analysis techniques (Hjerlac, 2004). Whether, and to what extent, local jerk may change subsequent to running-induced fatigue, accordingly, remains largely unexplored and despite the use of jerk metrics in other health-related domains, jerk is not commonly evaluated in running contexts.

Nevertheless, despite the apparent theoretical support for jerk as a sensitive measure of deteriorating movement proficiency within the relevant literature, the grouped tibial analysis did not reveal any changes greater than the smallest worthwhile change for any jerk metrics between either pre- to post-fatigue conditions, or across the quartiles of the fatiguing run.

Similarly, analysis of the individual tibial jerk data, collected during both the pre- and post-fatigue intervals and the fatiguing run, detected few incidences of change greater than the smallest worthwhile change for jerk metrics. Notably, the sparsity of incidences of meaningful change evaluated using smallest worthwhile changes, contrasted with analysis, using inferential statistics which identified numerous incidences of significant change between pre- and post-fatigue intervals and across the quartiles of the fatiguing run (see Appendix J).

9.3.3 Shock Attenuation

The concept of shock attenuation is inevitably entwined with that of system stiffness, such that both are commonly advocated as related and sensitive metrics capable of identifying fatigue-induced changes to running proficiency (For example:

Morin et al, 2005). Theoretically, in relation to shock attenuation, 2 competing theories have been advocated by different authorities (Giandolini et al, 2016). One suggests that exercise-induced fatigue diminishes the capacity to dampen the severity of impact shocks at the tibia, sacrum and/or thorax. And some research, suggesting that fatigued running increases tibial, sacral and thoracic accelerations, supports this contention (Verbistsky et al, 1998; Mizrahi et al, 2000; Mercer et al, 2003). The counterclaim is that runners maintain and/or decrease the magnitudes of impact, and subsequently the negative consequences of exacerbated shock, by naturally and innately adjusting stride patterns. Both Abt and colleagues (2011) and Clansey and colleagues (2012), for example, did not observe any changes to peak tibial and head acceleration and/or shock attenuation after fatigued running.

Accordingly, although recently it has been suggested that direct measurement of acceleration timeseries at upper and lower body locations provides a uniquely direct means of calculating SA (for example: Castillo & Lieberman, 2018). Whether this is the case remains unclear, and whether SA increases or decreases, or both, subsequent to fatigue remains unresolved.

9.3.3.1 Shock attenuation between tibia and pelvis

Upon group analyses of shock attenuation metrics, neither the pre- and post-fatigue SA means experienced between tibias and pelvis, nor the SA experienced between tibias and thorax, exceeded the smallest worthwhile change for SA metrics. Similarly, no changes in SA magnitudes, across the quartiles of the fatiguing run, exceeded the relevant SWCs.

However, upon individual analysis the differences in means between pre- and post-fatigue conditions exceeded the relevant SWC for 5 of the 15 runners. For two runners, this change occurred between tibia and pelvis, and for the other three, occurred between tibia and thorax. Between both tibia and pelvis, and tibia and thorax, the direction of meaningful changes, experienced by runners, occurred in different directions, i.e., changes variously resulted in increases and decreases in magnitude.

Across the quartiles of the fatiguing run, the shock attenuation experienced between tibial and pelvic sensors exceeded the relevant SWC on only 3 occasions, whereas the shock attenuation experienced between tibia and thorax exceeded the relevant SWC on 10 occasions across quartile transitions. Interestingly, on 3 occasions when changes to SA – P and SA – T occurred simultaneously in the same runner, i.e. both metrics changed, for the same runner, across the same quartile transition. Notably, however, on 2 of these 3 occasions’ the direction of change was in opposite directions, i.e., if SA – P increased, SA – T decreased.

These findings highlight that, although both metrics are closely related (in terms of sensor locations and methods of calculation), they are not interchangeable and outcomes, and subsequent interpretations, are likely to differ substantially depending on the anatomical locations between which the SA metric is assessed. Although both metrics assess the same phenomenon, differing only in the location of the trunk-mounted sensor, there seems little correspondence between the two. In the absence of any external measure of validity, there does not appear to be any clear means of evaluating whether one metric is right and another wrong or of establishing which measure is a better reflection of fatigue-induced coordinative change. These findings support recent arguments, by Edwards and colleagues (2019) for example, highlighting the limitations of assuming that different metrics will provide consistent results regardless of attachment location.

Notably, the results of the shock attenuation analysis serve to re-enforce a recurring theme of this thesis, specifically, that when the same metric is calculated at different anatomical locations, or in this case between different anatomical locations, results are not reliably transposable. The same measures, accordingly, assessed between different anatomical locations, may provide conflicting information leading to very different behavioural interpretations.

9.3.3.4 Overview of individual shock attenuation changes

Despite a coherent theoretical argument suggesting why SA could, or should, provide relevant insight into the processes of fatigue-induced coordinative change, the empirical reality demonstrated in the investigations included in this thesis serves to question whether this is the case. SA outcomes exhibited meaningful change, but those changes did not appear to exhibit any consistency, either in terms of direction of change and/or between measures of the same phenomenon assessed between different locations. This is especially noteworthy as, in commercial and practical contexts, results from investigations conducted using sensors positioned at one location are being transposed to others in the belief that the insights gleaned from one location hold across locations. The evidence here, however, suggests this is not the case.

9.3.4 Resultant Accelerations

Measuring local accelerations using a triaxial accelerometer, and calculating the resultant, is a method of capturing all axes of acceleration and presenting them in a single metric, without the need to precisely align the device (Sheerin et al, 2018). Resultant accelerations (RA) have been advocated as sensitive measures of running induced fatigue in the literature, for over 25 years (Lafortune and Hennig, 1995). Recently, the potential value of RA assessment has been re-visited and reiterated by Sheerin and colleagues (2018). Nevertheless, Gandolini and colleagues (2015), upon reviewing studies investigating impact shock, published over the previous 2 decades, concluded that the overwhelming majority had focussed exclusively on vertical impact, and although a small few had considered mediolateral accelerations, none had accounted for anteroposterior accelerations (Gandolini et al, 2015).

Accordingly, despite receiving theoretical support, the question of whether triaxial resultant accelerations modulate subsequent to fatigue, remains under-investigated and unresolved. Similarly, whether RAs assessed at other anatomical locations change following fatigue remains unanswered and, as yet, no published study has quantified the RAs experienced simultaneously at multiple anatomical sites during fresh-to-fatigued running.

9.3.4.1 Tibial resultant accelerations

Within the context of the grouped analysis, and further highlighting the potentially differential loading experiences imposed on each tibia during running, the differences between left tibial RA means between pre- and post-fatigue conditions exceeded the SWC for that metric, whereas the magnitude of right tibial RA did not change. Similarly, during the fatiguing run, the difference between left tibial RA means, between quartiles 3 and 4, exceeded the relevant SWC. Again, there was no meaningful change at the right tibia.

Upon application of the relevant SWCs to the pre- and post-fatigue data, the magnitude of differences exceeded the SWC for tibial RA on 20 occasions, across both tibias (67%). 10 of the 15 runners demonstrated meaningful change in RA – L following the fatiguing run. Of these 10 runners, 9 increased, and one decreased, the magnitude of RA – L. Similarly, 10 runners demonstrated significant change in the magnitude of RA - R following fatigue. Of these 10 runners, 9 increased and 1 reduced RA magnitude. Interestingly, 1 runner decreased RA at the left, whilst simultaneously increasing RA at the right, tibia. Similarly, over the course of the fatiguing run, differences in tibial RA means exceeded the relevant SWC across 69% of all quartile transitions. At both tibias individual runner’s RAs variously increased and decreased across sensor locations, i.e. there was no coherent directional pattern of change.

Notably, the RAs for both tibias changed simultaneously on 19 occasions. However, on 3 of these occasions’ the magnitude of the RAs at both tibias changed in different directions, i.e., RA increased at 1 tibia, whilst decreasing at the other. Accordingly, on only 16 occasions (18%), from 87 opportunities, did the change, experienced by both tibias, occur during the same transition and in the same direction.

9.3.4.2 Pelvic resultant accelerations

Upon group analysis of the pre- and post-condition intervals, no pelvic SA metrics meaningfully changed during between pre- to post-conditions or across the quartiles of the fatiguing run. Upon individual analysis, however, 5 runners exhibited differences greater than the relevant SWC between pre- and post-fatigue conditions. 3 runners increased pelvic RA, whilst 2 decreased pelvic RA. Similarly, across the quartiles of the fatiguing run, pelvic RAs exceeded the relevant SWC on 19 occasions across the 44 quartile transitions (42%). These changes were distributed across all pelvic metrics and these changes resulted in both increases and decreases to pelvic RA magnitudes.

9.3.4.3 Thoracic resultant acceleration

When the grouped thoracic RA changes were compared against the relevant SWC, for both pre- and post-fatigue intervals and across the quartiles of the fatiguing run, no meaningful differences were detected. At an individual level, however, between pre- and post-fatigue conditions, 13 runners exhibited changes to thoracic RAs that exceeded the relevant SWC.

Similarly, across the quartiles of the fatiguing run, thoracic RAs exceeded the relevant SWC a total of 24 times, i.e., 58% of the opportunities to change. Of these changes, 15 resulted in increases, and 9 resulted in decreases, in RA magnitudes. Interestingly, both RA – P and RA – T changed, simultaneously, i.e. during the same quartile transitions, on 9 occasions. However, on 4 of these occasions the RA metrics changed in different directions, i.e., if RA – P increased, RA – T decreased, and vice versa.

Notably, the pelvic and thoracic runner-specific resultant acceleration data illustrates that, although rarely documented or investigated, upper-body attachment sites experience extensive local acceleration changes over the course of moderately fatiguing runs. Finally, its additionally worth noting that due to the diversity of directional and temporal resultant acceleration changes, no runners exhibited the same pattern of RA change across upper- and lower-limb sensor locations across either running intervention.

As a final generalised observation, within the experimental cohort, fatigue-induced change appeared highly customised to individual runners, and there was no evidence of a consistent, coherent, shared direction of change across individuals following fatigue. Instead, different runners experienced various permutations of increases or decreases, at different locations and across different quartile transitions.

9.4 Discussion

A central organising principle, throughout this thesis, has been the use of fatigue to perturb, and subsequently examine, changes in running coordination between non-fatigued and fatigued states. In this section, the experimental outcomes of the thesis are contextualised against 5 core observations emerging from the analyses contained in chapters 7 and 8, each highlighting a distinct facet of the running coordination phenomenon.

9.4.1 Ongoing coordinative change is pervasive and not necessarily driven by fatigue

There is a perception, reflected in both the design of empirical investigations and within practical coaching lore, that coordinative change over the course of a fatiguing run is primarily driven by fatigue. Accordingly, coordinative changes are assumed to be more common towards the end of fatiguing efforts. This coordinative change is subsequently perceived as driving increased risk of negative outcomes – primarily running-related injury and/or increasingly detrimental alterations to habitual running patterns.

Conventionally, investigations seeking to uncover fatigue-induced effects on running behaviours, assume that change is primarily enforced and a consequence of accumulating fatigue. Subsequently, studies typically assess a specific metric ‘before’, and compare that assessment with a quantification of the same metric ‘after’, effortful running. Yet the investigation, detailed in chapter 8, evidenced that extensive meaningful changes, across all local and resultant acceleration measures and all sensor locations, were present during the early portion of the fatiguing run; thereby suggesting that instances of meaningful change may be an ever-present phenomenon and that change occurs even in the absence of substantial exertion or fatigue. This observation was unexpected and appears undocumented within the relevant running-related literature.

Explanatory frameworks, such as, for example, the Uncontrolled Manifold hypothesis (UMH), as noted in chapter 3, could provide a theoretical rationalisation for these observations. Under the assumptions of the UMH, only prioritised parameters (in any given context) are afforded the control resources necessary to tightly regulate behaviour. Conversely, non-prioritised parameters are less tightly regulated and are therefore free to deviate within broader, more expansive limits.

Accordingly, it may be the case that not all changes occur in direct response to diminished capacities. Some on-going change, as suggested in chapter 3 of the literature review, may be beneficially adaptive. Such changes may arise simply because there is a high cost to tightly regulating ongoing change, whereas there is often a low cost and low risk to allowing, in a non-priority dimension of the running action, more expansive, non-detrimental, variation. Thus, there may be a distinction between adaptive and mal-adaptive change. There may be a cost/benefit trade-off, whereby aspects of the impact event which need to be tightly regulated, are tightly regulated, yet facets of the impact event which do not any pose immediate threat are allowed to diverge between broader limits.

This highlights a potentially important limitation of these studies, and the broader literature on this topic. Specifically, in some contexts, ongoing change may be a natural, beneficial and adaptive phenomenon; yet, in others, may be an indication of accumulating coordinative deficits driven by fatiguing exertion. In either case, as a result, change is detected in the outcome measure. One form of change is potentially benign; one form is potentially destructive. Currently, however, these distinctions cannot be discerned using the methodologies employed in conventional running protocols, nor the methodologies used in this thesis. Subsequently, unless enforced change can be distinguished from non-detrimental variation, interpreting the relevance and importance of detected change will remain difficult.

9.4.2 Sensitivity to specific conditions

A feature of the analyses conducted in chapters 7 and 8 is that experimental variables –such as, environmental parameters, treadmill conditions, footwear and nutritional status—, that commonly change between data collections, remained constant between both the pre- and post-fatigue, and the fatiguing, protocols.

Accordingly, the only changing variables, between both data collections, were a 1 kph increase in running speed for the pre- and post-fatigue intervals, the duration of the effort and an inevitably changing fatigue status. Nevertheless, within the context of the group analysis, changes observed during the pre- and post-intervals were not observed during the fatiguing run, and vice versa. As illustration, there was a meaningful increase in peak vertical tibial decelerations before and after the fatiguing run, yet there was no increase in peak vertical tibial decelerations during the fatiguing run. Conversely, over the course of the fatiguing run, the magnitudes of peak vertical tibial accelerations increased, but the magnitudes of peak vertical tibial accelerations between the pre- and post-fatigue intervals did not. These distinctions, following what could be considered relatively minor changes in running pace and fatigue status, are well within the range of distinctions commonly encountered between different experimental protocols in this field. In this light, the mixed findings evident in the literature, and noted in chapter 4, are perhaps unsurprising and potentially serve to illustrate the high sensitivity of these measures to subtly changing conditions.

9.4.2.1 Sensitivity to specific metrics

The analyses detailed in chapters 7 and 8 illustrate that although each metric has been advocated as a sensitive measure of exertional or fatigued status, there was no coherent ageement between metrics. Accordingly, whether runners can be judged to be running in a compromised state, or not (i.e., whether metrics are observed to change between non-fatigued and fatigued conditions), seems dependent on the metric used.

Accordingly, despite various levels of support for each assessment measure within the literature, subsequent interpretations of the experimental findings vary extensively based on the specific outcome measure selected. A subsequent concluding point is that the use of different metrics drives different observations, and subsequently different theoretical interpretations of results.

9.4.2.2 Sensitivity to time-point

There is an implicit assumption, within running contexts, that accumulating fatigue is the primary cause of coordinative change. Following this logic, fatigue-induced change is considered more likely, and more extensive, towards the end of strenuous runs. Accordingly, many running investigations follow a design, whereby an outcome measure is assessed at the beginning and end of an effortful run. The underpinning assumption is that compromised coordination, driven by exertion, is the primary driver of change and that coordination is most compromised in the presence of greater levels of fatigue. Accordingly, changes to outcome measures are considered most likely in the latter stages of effortful runs. Interestingly, however, across the quartiles of the fatiguing run, local acceleration, and the other acceleration-derived, metrics (whether evaluated using either smallest worthwhile change or inferential statistics) did not exhibit this pattern of change. For example, the highest incidence of meaningful change, recorded for local tibial accelerations, occurred between quartiles 1 and 2, and this incidence of change progressively declined over the subsequent two quartile transitions (see Tables 8.1.2 and 8.1.3). Yet, beyond this observation, there was no coherent pattern of change shared amongst the various metrics as the fatiguing run progressed. There was no evidence that metrics shared a common pattern of either increasing or decreasing change across the quartiles of the fatiguing run.

Notably, across the quartiles of the fatiguing run, the extent of meaningful change differed at each evaluated time-point. Accordingly, from a research design perspective, the extent of detected change seems likely to vary, perhaps extensively, dependent on the times-point selected for assessment and re-assessment. These observations suggest that: (a) the selection of running durations and assessment time-points is critical and will likely exert a substantial influence on results, and (b) serial or continuous metric monitoring may provide richer, more insightful and perhaps very different, interpretations of how coordinative change progresses over the course of fatiguing runs.

9.4.2.3 Sensitivity to location

Recently, the relationships between accelerations experienced at different attachment locations were questioned by Nedergaard and colleagues (2017), who suggested that sensors, attached to different anatomical landmarks, exhibited only weak to moderate correlations between different segmental accelerations. Subsequently, the authors cautioned against extrapolating trends from one location, to other locations and/or to whole-body contexts (Nedergaard et al, 2017; Edwards et al, 2019). The findings detailed in chapters 7 and 8 support these conclusions as the results of each metric appeared highly sensitive to location. Specifically, across measures, findings varied according to location, or in the case of shock attenuation

(SA), findings varied depending on whether the metric was calculated between tibia and pelvis, or tibia and thorax.

Using resultant acceleration as an example, remarkably, over the course of the fatiguing run (which contained a total of 87 individual quartiles), there were 23 occasions when RA - L and RA - R changed concurrently. Yet on 9 of these 23 occasions, change occurred in opposite directions, i.e., if RA -L increased, RA – R decreased and vice versa. Accordingly, meaningful changes to resultant accelerations, experienced by both tibias, occurred concurrently and in the same direction during only 16% of all quartile transitions.

In concluding this sub-section, and echoing the suggestions proposed in recent publications (Nedergaard et al, 2017 and Edwards et al, 2019), the analyses within this thesis suggest that findings, at any given attachment location, are specific to that location, and subsequent results are not reliably transposable to other anatomical sites, nor are they likely to be indicative of whole-body behaviours.

9.4.2.4 Sensitivity to assessed side

The question of whether each leg habitually experiences different shock loadings during running, remains contentious and inadequately understood. In relation to high impact tasks, such as jumping or hopping, some previous evidence suggests there are side-to-side differences which were, in this case, attributed to limb dominance (Brown et al, 2009; Ford et al, 2003). In contrast, other studies found no significant differences between dominant and non-dominant legs in a variety of impact-related tasks (Flanagan and Harrison, 2007; Hobara et al, 2013).

Specifically, in relation to running, an early study by Hamill and colleagues (1984) found no side-to-side differences in running kinetics between dominant and non-dominant legs, although the sample size was small (N = 5). More recent experimental work arrived at a similar conclusion, suggesting limb dominance had essentially no effect on lower extremity joint kinematics and kinetics (Brown et al, 2014), or leg and vertical stiffness (Pappas et al, 2015), during running. Similarly, Brown and colleagues (2014) were unable to establish an interaction between fatigue status and limb side in a running task, and subsequently concluded that both legs fatigued at similar rates. Other recent experimental work —using repeated treadmill sprints—also concluded that, although inter-limb differences were observed in some kinetic and kinematic characteristics, the lack of interaction between sprint repetitions and leg suggested both legs fatigued at similar rates (Girard et al, 2017). In relation to prior injuries, some work suggests that injury does induce asymmetries during a fatiguing run (for example: Radzak et al, 2017). Conversely, other running-specific investigations suggest that the magnitudes of common dynamic loading measures, assessed via accelerometery, are not higher on the previously injured leg (Schütte et al, 2018).

Notably, multiple asymmetries were observed within the context of the analyses conducted in chapters 7 and 8. In the group analysis conducted in chapter 7, for example, the accelerations experienced by left and right tibias differed for vertical deceleration, lateral acceleration, medial deceleration and posterior deceleration. In chapter 8, detailing the results of the fatiguing run, left and right tibial experiences meaningfully differed for vertical acceleration (but not deceleration) and for both lateral acceleration and medial deceleration.

Importantly, the analysis of runner-specific acceleration data revealed uniquely individually personalised patterns of change between both tibias. The differences between the impact loadings experienced by both tibias undermines a conventional, but in the light of these investigations perhaps erroneous, assumption. Specifically, the perception that both tibias experience very similar mechanical stresses during running. This perception has, historically, fuelled the assumption that attaching a sensor to a single tibia will adequately reflect the experiences of both (for example: Meardon et al, 2011). The results presented in this thesis, however, underline the lack of correspondence between changes experienced at the left and right tibias, across all measures.

It is important to note that although, within athletic running contexts, asymmetries are generally considered undesirable, it is unclear whether asymmetries are always, or even most often, detrimental to movement proficiency. Furthermore, and more specifically in relation to the running asymmetries detected in this thesis's investigations, it is not clear whether these asymmetries are:

a) Permanently embedded constraints or flexibly variable features modulating adaptively in response to subtly changing running conditions
b) Long-lasting and embedded deficits, or temporary means of dispersing work burdens and reallocating mechanical stressors

c) Injury risk factors and impediments to running performance or functionally protective compensatory strategies

9.4.3 Extensive inter- and intra-individual variability

An obvious outcome of the individual analyses, conducted on the various metrics, was the extent of inter-individual variation across runners, across metrics and across attachment locations. This variation was such that each individual runner demonstrated diverse permutations and patterns of change across the metrics and locations assessed. This phenomenon is exemplified by noting that the number of local accelerations experienced, across quartiles, ranged from 75, for a runner who recorded a final RPE of 8.5, to 17 for a runner who's final RPE was 9. Also worthy of note is the observation that, aside from vertical tibial decelerations during the pre- to post-intervals, no metric exhibited a strong or obvious trend, across both tibias, that was shared across all individuals who exhibited change. Individual analyses also illustrate that patterns of change are not always consistent within a single runner. As example, over the course of the fatiguing run, 10 runners experienced more local acceleration changes in upper-body that lower-body sites, only to shift in a later quartile to experiencing the predominance of changes in the other half of the body.

Interestingly, the runner who exhibited the lowest number of meaningful changes to tibial accelerations (3), between the pre- and post-fatigue intervals, had the 2nd lowest number of thoracic changes (3), and did not change either SA measure nor either RA measure. Yet this same runner had the second highest number of pelvic acceleration changes (7). In contrast, the runner who demonstrated the highest number of tibial changes, exhibited a very low incidence of change at both pelvic and thoracic sites. This is not, however, to suggest that certain runners do not demonstrate consistencies. One runner, for example, persistently demonstrated a high incidence of meaningful change across tibial acceleration metrics, while simultaneously exhibiting one of the lowest incidences of meaningful change at the pelvis. The intention, here, is not to try and derive or claim insight from individual or small cohort observations, but simply to note that how runners adapt, as runs progress, appears highly individually malleable.

9.4.4 Fatigue-induced changes to upper-body coordination outcomes

Coordinated control of the trunk must be closely regulated to facilitate a diversity of running-related objectives, for example, to avoid postural collapse during ground contact; to effectively attenuate impact shock; to manage the complex vortex of forces emanating from shoulder and hip drives; to control and counterbalance the powerful movements of the lower limbs. Managing these multiple stressors imposes significant mechanical and control demands on the musculature of the trunk (for example: Schütte et al, 2016). Supporting this conjecture, recent evidence suggests that fatigue-driven deteriorations in trunk stability may potentially serve as an early indicator of disintegrating coordinated control and declining running proficiency. Following this logic, it has recently been suggested that changes in lumbo-pelvic coordination, induced by running fatigue, are detectable using trunk mounted accelerometers (Schütte et al, 2015, 2016 & 2017; Lubetzky, Harel & Lubetzky, 2018).

Within the context of the evidence presented here, it seems clear that upper-body attachment sites experience high incidences of local acceleration change. This high incidence of local change, occurring concurrently at both upper- and lower-body sites, has not previously been documented. When the frequency of meaningful changes across both lower and upper body sensor attachment locations were compared, there were a total of 365 lower-body and 294 upper-body changes across the quartiles of the fatiguing run. Furthermore, between quartiles 1 and 2, more local acceleration changes occurred in the lower body than the upper body; yet, between quartiles 3 and 4, more changes were recorded at upper-body locations.

The findings in chapters 7 and 8 support the suggestion that the assessment of trunk-mounted metrics may provide a window into running coordination proficiency. Nevertheless, given the limited number of available studies, it remains unclear which, if any, trunk metrics provide a clear indication of running coordinative proficiency. Accordingly, currently, there is a need for more investigations combining triaxial sensor assessment of upper- and lower-body behaviours during progressively fatiguing runs. As detailed here, and given the dearth of insight in this area, such investigations may prove a fruitful avenue of exploration.

9.4.5 The special case of vertical tibial deceleration

Although, as noted in chapter 4 and elsewhere, peak tibial deceleration, measured during ground contact, is considered a reliable method of quantifying impact shock, yet whether vertical tibial deceleration increases or decreases subsequent to fatigue remains controversial (Meardon et al, 2011; Greenhalgh et al, 2012).

In the context of the investigation detailed in chapter 7, all metrics variously increased and decreased following fatigue, except for vertical tibial deceleration. This metric changed 10 times between pre- and post-fatigue conditions and on each occasion increased by a magnitude exceeding the relevant SWC. Subsequently, across the quartiles of the fatiguing run, there were a total of 31 instances of meaningful change to vertical tibial deceleration. However, within this analysis the distribution of increases and decreases to vertical tibial deceleration, as evaluated by comparison to the relevant SWC, were more equitably distributed, with vertical deceleration shock increasing on 17, and decreasing on 14, occasions. Subsequently, a seemingly conclusive observation, based on the analysis results detailed in chapter 7, was strongly contradicted by the results of the analysis detailed in chapter 8, despite the similarity of conditions for both investigations. These conclusions, accordingly, suggest that perhaps vertical tibial deceleration is no different than the other metrics employed in the studies documented here, in that vertical tibial deceleration variously increases and decreases over the course of a fatiguing run. Accordingly, vertical tibial deceleration may not be a reliable indicator of compromised running proficiency.

9.5 The limitations of this thesis

The primary limitation of this work is the pairing of a complex experimental question with a sample size that, although larger than many of the studies on this topic within the literature, remains small. Similarly, there is a mismatch between the complexity of the topic under investigation and the simplicity of the analysis methods employed. Given the multiplicity of parameters that change in tandem, during running activity, and the complex inter-relatedness of the multiple facets of the coordination phenomenon, in retrospect, it was unrealistic to expect that analysis based on comparing maximum and minimum means would provide adequate insight into the true nature of such a complex, richly layered and multi-faceted phenomenon.

The application of smallest worthwhile changes (SWCs) served as a beneficial means of determining whether the differences noted between conditions were likely to exert real-World influences, or were merely a consequence of random biological variation and/or technological error (Hopkins, 2000). However, in part due to constraints imposed by COVID-19 restrictions, recruitment of participants for the test-retest reliability study was difficult, and limited to 3 participants. A larger cohort would have further refined the SWC calculations for all metrics.

In relation to the experimental cohort, although all shared the core attribute of being experienced and regular runners, they were a mixed group, in terms of gender, age, injury and training histories, and running abilities. Participants that shared more key commonalities (age, gender, performance standards, and so on), would perhaps provide less varied, more coherent, results.

Accurately assessing fatigue is an ongoing problem in this domain. As with many studies in this area, this investigation used a subjective CR-10 chart to record RPEs. Given the subjective, perceptual nature of RPEs, however, there is substantial scope for vagueness and variation across individual ratings. Although, as discussed in section 6.2.2, Borgs CR scales offer many practical advantages, appropriate implementation of CR scales demands clear and thorough education of both the experimenter and participants. Although all participants were familiarised with the CR–10 Scale prior to assessment and although care was taken to thoroughly explain the descriptors of effort detailed on the scale, a more comprehensive briefing would have promoted greater clarity and consistency (see, for example, Halperin and Emanuel, 2020).

Furthermore, it is worth noting that the levels of fatigue generated in this experimental protocol were not severe. In fact, an initial premise of the thesis was to determine whether moderate levels of fatigue, such as would regularly be encountered during moderately hard, but not extreme, running training sessions, influenced local accelerations and acceleration-derived metrics. Subsequently, desired levels of exertion were contextualised, described, and defined against the intensities encountered within regular training and/or competitive running efforts. However, it is inevitable that individual motivations and interpretations of the desired level of fatigue to be attained varied substantially between participants. Subsequently, although heart rates were also monitored to ensure they surpassed age-calculated criteria at the end of the fatiguing run, additional more perceptive indicators of exertion and/or fatigue would add insight and greatly benefit future investigations.

Finally, an unresolved problem within the realm of running coordination research is, as noted in this thesis, our ability to discern between the multitude of changing variables, and to distinguish between non-critical and critical indicators of changing running proficiency. In these terms, although monitoring multiple dimensions of change is easy; knowing which changes are most important, remains unclear.

9.6 Implications for future research

Technological innovations provide us with many potentially insightful assessment tools. For example, as recently noted, very few studies have used more than a single acceleration sensor in running contexts (Rabuffetti, Scalera, and Ferrarin, 2019). Accordingly, we know very little about how multiple acceleration time-series change, relative to each other, at different locations and along different axes during fatiguing runs. Notably, recent research, including that documented in this thesis, have found that running-induced fatigue greatly influences horizontal plane trunk accelerations (for example: Schütte et al, 2016).

Similarly, we know little about the extent of inter-individual customisation of acceleration signatures. The evidence presented in chapters 7 and 8 suggest this customisation is extensive and that runners do vary widely, and uniquely, in response to changing running conditions. As yet, however, the factors which drive and influence this personalised individualisation process are not well understood.

Furthermore, for many years, the study of running coordination has relied on investigations focussed on the analysis of mean peak magnitudes, primarily evaluated using p values as a threshold for change. In the future, it seems likely there will be a continued move away from the evaluation of peak values, and towards a more comprehensive quantification and tracking of the relationships between pertinent metrics.

Running coordination is, ultimately, the coherent integration, regulation and control of multiple, ever-modulating outputs to achieve desired running objectives, for an acceptable cost, in terms of psycho-emotional discomfort, energetic expenditure and neural processing commitment. Given this evident complexity, it seems unlikely that the continued investigation of the peak magnitudes of impact metrics will provide deeper, more sensitive understanding of this complex phenomenon.

Similarly, given this inherent complexity, a deeper understanding could be driven by the use of more subtle and complex analysis methods, which are more suited to studying large datasets. Recently developed and/or re-purposed analytical methods, such as, for example, statistical parametric mapping (Friston, 2003), may be well suited to addressing specific, currently unanswered, questions in this realm.

Similarly, techniques such as the use of the sample entropy of local acceleration time-series (as recently pioneered by Schütte and colleagues (2016, 2018), whereby non-linear mathematical algorithms (described by Richman and Moorman, (2000)) are used to quantify the uncertainty or unpredictability of the accelerometery time series, may be especially insightful in relation to coordination processes. Yet, clearly, such methods require future investigation and replication within this context.

Furthermore, an overwhelming majority of running-related investigations have focussed on the kinematics of the lower limbs. Accordingly, studies focussing on changes in the control of the trunk, in response to accumulating fatigue, are severely under-represented in the literature. Yet, as evidenced here, and in work cited here, trunk kinematics change extensively during fatiguing runs. Accordingly, further research, blending information gathered from upper- and lower-body sensor attachment sites, could contribute to a deeper understanding of the complex patterns of change that may, in the future, help to better identify fatigue-induced detriments to running coordination.

Finally, a clear implication of the experimental work within this thesis is that outcomes, in terms of the accelerometer derived metrics, are highly sensitive to specific conditions. In this thesis, a collection of metrics, derived from information obtained from the same sensors, on the same runners, provided different interpretations of how running coordination outputs changed subsequent to fatigue. Specifically, results suggest outcome measures were sensitive to:

- Metric
- Time-point of evaluation
- Attachment location

In summary, although limitations exist, so do opportunities. Opportunities presented by technological and methodological innovations. Clearly, however, researchers should endeavour to be aware of the complexities inherent within this domain and take steps to circumvent and obviate the key limitations outlined here.

9.7 Conclusions

The goal of this thesis was to endeavour to illuminate a complex phenomenon, running coordination. In retrospect, this was a naive and overly ambitious objective. Nevertheless, there were interesting findings, some previously unreported, that perhaps serve to illuminate, in some small way, our understanding of running coordination.

The investigations were designed around a central organising principle. Specifically, the observation that fatigue drives coordinative change. Yet, studies evaluating the effects of fatigue on running biomechanics have produced conflicting results, and the domain remains mired in controversies. Subsequently, as recently noted by Sheerin and colleagues (2018), as yet there is no consensus on how fatigue influences a runner’s capacity to effectively manage the repetitive impact shocks imposed during running (Sheerin et al, 2018). Bearing that lack of consensus in mind, the studies documented here serve to emphasise, and to some extent illuminate, certain facets of the running coordination phenomenon. More specifically, the key findings of this thesis suggest:

i. Change is pervasive, even in the absence of fatigue. The capacity to quantify change is not necessarily a limitation. Instead, the problem lies in identifying which specific dimensions of change are most indicative of compromised running proficiency
ii. Measures obtained from running acceleration timeseries are highly specific to:
    a. Metrics. Such that even related metrics, derived from the same source data, provide very different and inconsistent indications of current running status
    b. Time-points. Underlying running parameters appear to persistently change, regardless of fatigue status, accordingly impact-related evaluations may differ substantially between consecutive segments of fatiguing runs
    c. Attachment locations. Conclusions derived from data collected at one location cannot, validly, be transposed to other locations
iii. Running coordination outputs vary substantially on an inter- and intra-individual basis. This response variability appears highly sensitive to slightly differing conditions
iv. Although the predominance of published literature focuses on evaluation of lower-limb kinematics, as noted here, trunk kinematics also undergo substantial change between similarly paced, moderately fatiguing runs

v. Vertical tibial deceleration remains, within running contexts, the most regularly evaluated local acceleration metric. Nevertheless, whether and how vertical tibial deceleration changes, following increasing running-induced fatigue, remains controversial with frequently conflicting findings. The different findings of the studies detailed here, following only minor changes in experimental design, demonstrate the sensitivity of vertical tibial deceleration to subtle differences in experimental design

As a final observation: A notable outcome of the investigations detailed here was the expansive diversity of responses, displayed by individual runners, to running-induced fatigue. This diversity of responses emphasises the capacity of human runners to subtly modulate multiple dimensions of movement in order to accommodate the multiple perturbations imposed during every running experience. Clearly, however, this expansive adaptive flexibility, both between and within individual runners, ensures that low powered experimental designs using isolated outcome measures and group mean magnitudes, are unlikely to drive a clear and comprehensive understanding of such a complex phenomenon as running coordination.

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Appendices

Appendix A: The Borg Category Ratio (CR10) scale

1 - 10 Borg Rating of Perceived Exertion Scale	
0	Rest
1	Really Easy
2	Easy
3	Moderate
4	Sort of Hard
5	Hard
6	
7	Really Hard
8	
9	Really, Really, Hard
10	Maximal: Just like my hardest race

Appendix B: Ethical Approval

3rd July 2014

Dave Colins and John Kiely

School of Sports Tourism & the Outdoors University
of Central Lancashire

Dear Dave & John

Re: STEMH Ethics Committee Application Unique Reference Number: STEMH 184

The STEMH ethics committee has granted approval of your proposal application 'Coordinative variability over the course of a short moderately fatiguing run'. Approval is granted up to the end of project date* or for 5 years from the date of this letter, whichever is the longer.

It is your responsibility to ensure that

- the project is carried out in line with the information provided in the forms you have submitted
- you regularly re-consider the ethical issues that may be raised in generating and analysing your data
- any proposed amendments/changes to the project are raised with, and approved, by Committee
- you notify roffice@uclan.ac.uk if the end date changes or the project does not start
- serious adverse events that occur from the project are reported to Committee
- a closure report is submitted to complete the ethics governance procedures (Existing paperwork can be used for this purposes e.g. funder's end of grant report; abstract for student award or NRES final report. If none of these are available use e-Ethics Closure Report Proforma).

Yours sincerely

Gill Thomson
Chair

STEMH Ethics Committee

* for research degree students this will be the final lapse date

NB - Ethical approval is contingent on any health and safety checklists having been completed, and necessary approvals as a result of gained.

Appendix C: Participants' consent form

Version 1 – 30/01/14

Study Number:
Participant Identification Number for this trial:

CONSENT FORM

Title of Project: Coordinative variability over the course of a short moderately fatiguing run

Name of Researchers: Profs Dave Collins and Jim Richards, John Kiely

The following test will require you to have a number of small and light accelerometers attached to your body to assess the accelerations generated when you execute a double legged jump, and while you run.

Actions will be taken to ensure your data remain anonymous and to prevent you from being identified in any future report/publication.

The experimental procedure requires you to execute a standard double legged counter-movement jump, and to run at a self-selected pace. The procedure should cause no discomfort beyond that which you would normally experience during a moderately fatiguing training run. However, if you do feel undue discomfort you are free to withdraw from the protocol at any time, and without explanation.

Before any of the tests are conducted the institutional review board require written consent, please complete, providing you agree to the terms of the research.

Please initial box

1. I confirm that I have read and understand the information sheet dated 30/01/2014 ☐ (version1) for the above study and have had the opportunity to ask questions.

2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason, without my medical care or legal rights being affected. ☐

3. I agree to take part in the above study. ☐

Name of Participant		Date Signature
Name of Person taking consent	Date	Signature

(if different from researcher)

___________________________ _________________ _____________________

Researcher Date

Signature

1 for participant; 1 for researcher

Appendix D: Participant Information Sheet

Version Number: 3
Date: 02/07/2014

Title of study: Coordinative variability over the course of a short moderately fatiguing run

Researchers: Professor Dave Collins, Professor Jim Richards, John Kiely

You are being invited to take part in a research study, being conducted as part of a PhD programme. The study is being conducted in the Movement Analysis Unit, Brooke Building, at the University of Central Lancashire. Participation is entirely voluntary. Please read this information sheet before deciding if you would like to participate. If you have any further questions, or concerns, please contact: John Kiely, School of Sports Tourism and the Outdoors, UClan by phone at 07795636296, or by email at jkiely@uclan.ac.uk

Aim of study:

Our primary aim is to establish how key running coordination variables may change over the course of a short, moderately fatiguing run. There are also two secondary objectives, (i) to assess whether or not both legs exhibit fatigue in a similar fashion over the course of the run and (ii) to determine if a simple jump test can provide insight into how fatigue may be manifest in each leg.

Who can participate in the study?

You should be a regular runner and/or field sports player; be experienced with running on a motorised treadmill; have no currently active injuries or disorders; and no medical conditions that limit physical activity or expose you to undue risk through physical exertion. Participants must be over the age of 18, with a history of > 3 running sessions per week, for at least 40 weeks of the year, for at least the past 12 months.

Who will conduct the research?

The research will be conducted by a team of researchers made up of experts in running and human movement.

What we will ask you to do?

Before beginning the physical component of the experimental protocol you will be asked to complete a short questionnaire asking about prior training and injury history. This will take between 5 and 10 minutes.

Following this we will attach (with adhesive tape) 7 Trigno low-mass wireless accelerometry units to sites on leg and torso. You will then be asked to warm up, on a motorised treadmill, as per your normal pre-training warm up routine. When you are satisfied you are ready to exercise we will ask you to perform a number (5-7) of counter-movement jumps. Following this we will ask you to run, at a pre-agreed pace, for 90seconds. Subsequently, following a 3minute recovery period, we will ask you to run at a pre-agreed pace until (a) you no longer wish to continue or (b) you rate your subjective level of fatigue at 8 or greater on a perceived exertion scale of 0 to 10. Finally, subsequent to a 3minute recovery period, you will once more be asked to run for 90seconds at the same pre-agreed pace.

The levels of exertion required by the testing protocol should not exceed those encountered in a moderate intensity training session.

On the day of testing we will wish to put accelerometry units on legs and torso, hence we would ask you to bring a either shorts or running leggings and a well-fitting training top. Changing facilities are available. The full visit will take no longer than 1 hour to perform.

Do I have to take part?

No, participation is entirely voluntary. If you agree to take part, you can change your mind and withdraw at any time without giving a reason. Further, you can withdraw permission for the use of the anonymised information, collected during your visit, at any time until the completion of the testing protocol.

If you would like to participate, then please contact a member of the research team within two weeks of receipt of this letter.

Are there any risks or benefits?

There will be no direct benefits to you for taking part in this study. The movements we will ask you to perform are common to your normal training activities and should not place any unusual or unhabituated stress on your biology and joints, therefore the risk of injury is very low. **However, all physical exertion carries an inherent risk. In the event of injury or adverse reaction immediate remedial action will be taken.**

What will happen to the data from the study?

All relevant data (obtained from the pre-test questionnaire, and from physical testing) will be stored in line with UCLAN regulations and in accordance with the data protection act. Electronic data will be stored on a password protected PC and on the University network. All consent forms and other documents will be stored so that no names can be associated with them in a locked filling cabinet. Electronic data and forms will be kept for 5 years following the end of the project, and then destroyed.

Who has approved this study?
This study has been approved by the STEMH (Science, Technology, Engineering, Medicine and Health) ethics committee, for the University of Central Lancashire

Who can I contact to discuss any issues or to make a complaint?
Any complaint about the way you have been dealt with during the study or any possible harm you might suffer will be addressed. If you have any complaints about the study or how you have been treated in the study, please in the first instance contact the researchers using the details provided, they will do their best to answer your questions. Should you wish to take your concerns further, please address any issues to the University Officer for Ethics at OfficerForEthics@uclan.ac.uk. Please include the study name or description (so that it can be identified), the principal investigator or student investigator or researcher, and the substance of the complaint.

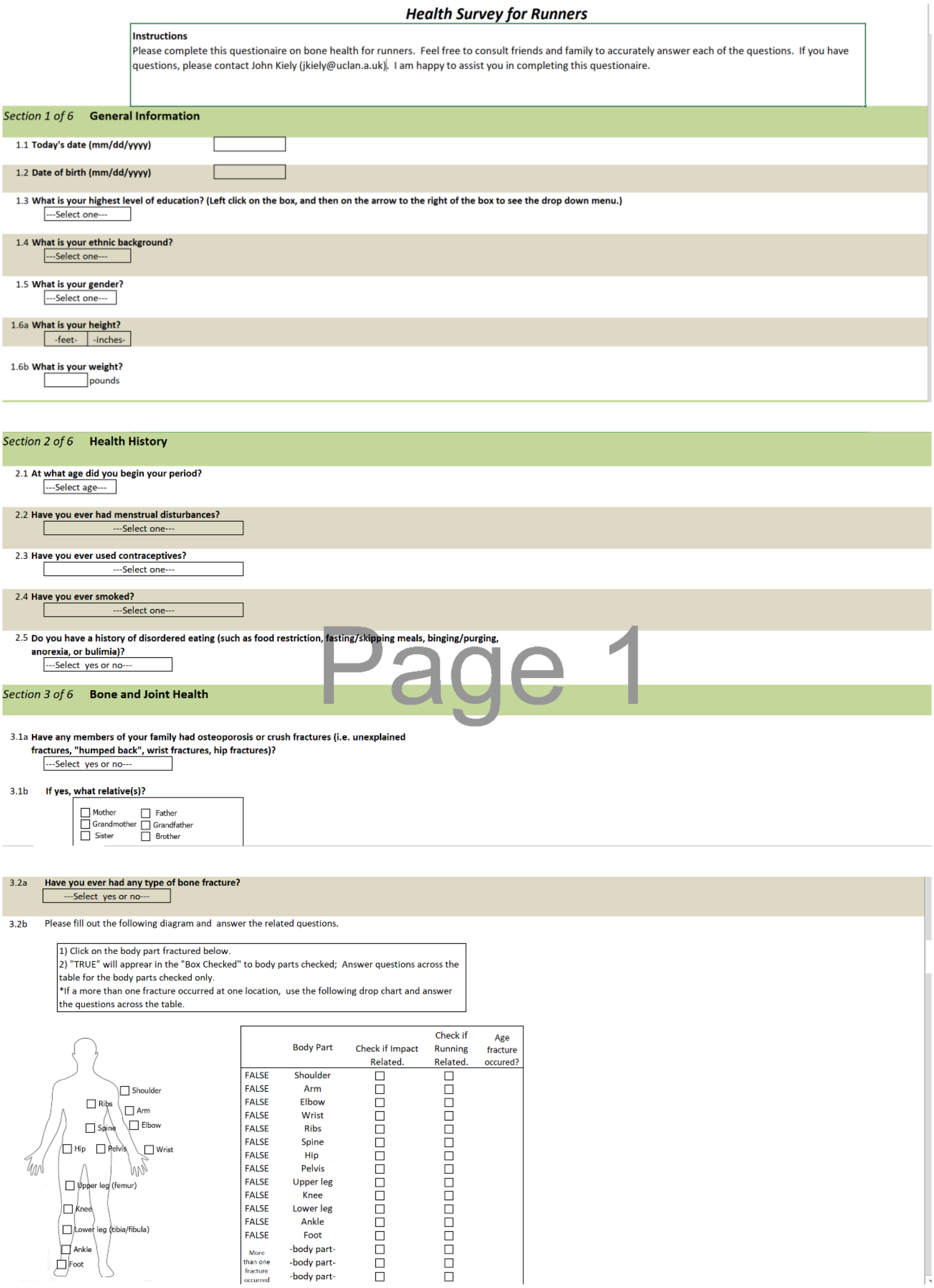
Thank you for taking the time to consider this study

If you have any queries, or if you may be interested in participating, please contact John Kiely at *jkiely@uclan.ac.uk*

Alternatively, feel free to contact any member of the research team. Full contact details are listed below:

- John Kiely: Institute of Coaching and Performance, SSTO, Greenbank Building, Gr153. By email at: jkiely@uclan.ac.uk or by phone at: 07795636296
- Professor Dave Collins: Institute of Coaching and Performance,SSTO, Greenbank Building, Gr153. By email at: djcollins@uclan.ac.uk or by phone at: 07595513540
- Professor James Richards: Allied Health Professions Unit, SSTO, Brook Building, BB118 By email at: jrichards@uclan.ac.uk or by phone at: 01772 89 4575

Appendix E: Health Survey for Runners Questionnaire



Health Survey for Runners

Instructions

Please complete this questionaire on bone health for runners. Feel free to consult friends and family to accurately answer each of the questions. If you have questions, please contact John Kiely (jkiely@uclan.a.uk). I am happy to assist you in completing this questionaire.

Section 1 of 6 General Information

1.1 **Today's date (mm/dd/yyyy)** ☐

1.2 **Date of birth (mm/dd/yyyy)** ☐

1.3 **What is your highest level of education? (Left click on the box, and then on the arrow to the right of the box to see the drop down menu.)**
---Select one---

1.4 **What is your ethnic background?**
---Select one---

1.5 **What is your gender?**
---Select one---

1.6a **What is your height?**
-feet- -inches-

1.6b **What is your weight?**
☐ pounds

Section 2 of 6 Health History

2.1 **At what age did you begin your period?**
---Select age---

2.2 **Have you ever had menstrual disturbances?**
---Select one---

2.3 **Have you ever used contraceptives?**
---Select one---

2.4 **Have you ever smoked?**
---Select one---

2.5 **Do you have a history of disordered eating (such as food restriction, fasting/skipping meals, binging/purging, anorexia, or bulimia)?**
---Select yes or no---

Section 3 of 6 Bone and Joint Health

3.1a **Have any members of your family had osteoporosis or crush fractures (i.e. unexplained fractures, "humped back", wrist fractures, hip fractures)?**
---Select yes or no---

3.1b **If yes, what relative(s)?**

☐ Mother ☐ Father
☐ Grandmother ☐ Grandfather
☐ Sister ☐ Brother

3.2a **Have you ever had any type of bone fracture?**
---Select yes or no---

3.2b Please fill out the following diagram and answer the related questions.

1) Click on the body part fractured below.
2) "TRUE" will apprear in the "Box Checked" to body parts checked; Answer questions across the table for the body parts checked only.
*If a more than one fracture occurred at one location, use the following drop chart and answer the questions across the table.

☐ Shoulder
☐ Ribs
☐ Arm
☐ Spine
☐ Elbow
☐ Hip
☐ Pelvis
☐ Wrist
☐ Upper leg (femur)
☐ Knee
☐ Lower leg (tibia/fibula)
☐ Ankle
☐ Foot

	Body Part	Check if Impact Related.	Check if Running Related.	Age fracture occured?
FALSE	Shoulder	☐	☐	
FALSE	Arm	☐	☐	
FALSE	Elbow	☐	☐	
FALSE	Wrist	☐	☐	
FALSE	Ribs	☐	☐	
FALSE	Spine	☐	☐	
FALSE	Hip	☐	☐	
FALSE	Pelvis	☐	☐	
FALSE	Upper leg	☐	☐	
FALSE	Knee	☐	☐	
FALSE	Lower leg	☐	☐	
FALSE	Ankle	☐	☐	
FALSE	Foot	☐	☐	
More than one fracture occurred	-body part-	☐	☐	
	-body part-	☐	☐	
	-body part-	☐	☐	

3.3a **Have you ever had any other type of bone or joint injury?**
☐ No ☐ Yes

3.3b

If yes, please check the box corresponding to the specific bone or joint injury and answer the questions in the following chart.

	Past Injury	Body Part	Right or Left	Running Related?	Injury Date (mm/yyyy)	Lost training days (in days)	Number of episodes	Who diagnosed injury?	Did you receive treatment for this injury?	What treatment did you receive?
☐	Bursitis			☐ Yes					☐ Yes	
☐	Compartment syndrome			☐ Yes					☐ Yes	
☐	Dislocation			☐ Yes					☐ Yes	
☐	ITB syndrome	leg		☐ Yes					☐ Yes	
☐	Ligament sprain			☐ Yes					☐ Yes	
☐	Low back pain	back	N/A	☐ Yes					☐ Yes	
☐	Muscle Strain			☐ Yes					☐ Yes	
☐	Patellofemoral syndrome	leg		☐ Yes					☐ Yes	
☐	Shin splints	lower leg		☐ Yes					☐ Yes	
☐	Shin splints	lower leg		☐ Yes					☐ Yes	
☐	Stress fracture			☐ Yes					☐ Yes	
☐	Tendon strain			☐ Yes					☐ Yes	
☐	Tendonitis			☐ Yes					☐ Yes	
☐	Turf Toe	toe		☐ Yes					☐ Yes	
☐	Other (please list here):			☐ Yes					☐ Yes	

3.4a **Have you ever had a bone or joint surgery?**
---Select yes or no---

3.4b **If yes, please list the type:**
and date of surgery: --Month-- ---year--- ---Month--- ---year--- -Month-- ---year--- -Month-- ---year--- ---Month--- ---year--- --Month--

Section 4 of 6 **Running History**

4.1 **Do you run for fitness, recreation or competition?**
---Select yes or no---

4.2 **How would you classify your level of running?**
---Select one---

4.3 **What type of surface do you run on (check all that apply)?**
☐ Concrete ☐ Asphalt
☐ Trail ☐ Treadmill
☐ Track ☐ Grass
☐ Hills ☐ Flats

4.4 **Do you change directions while running?**
---Select yes or no---

4.5 **What is your preferred running shoe brand and model?**

4.6a **Do you wear orthotics?**
---Select yes or no---

4.6b **If yes, what kind?**
---Select one---

4.7 **Do you have a history of injury that has interrupted your training for more than 1 week?**

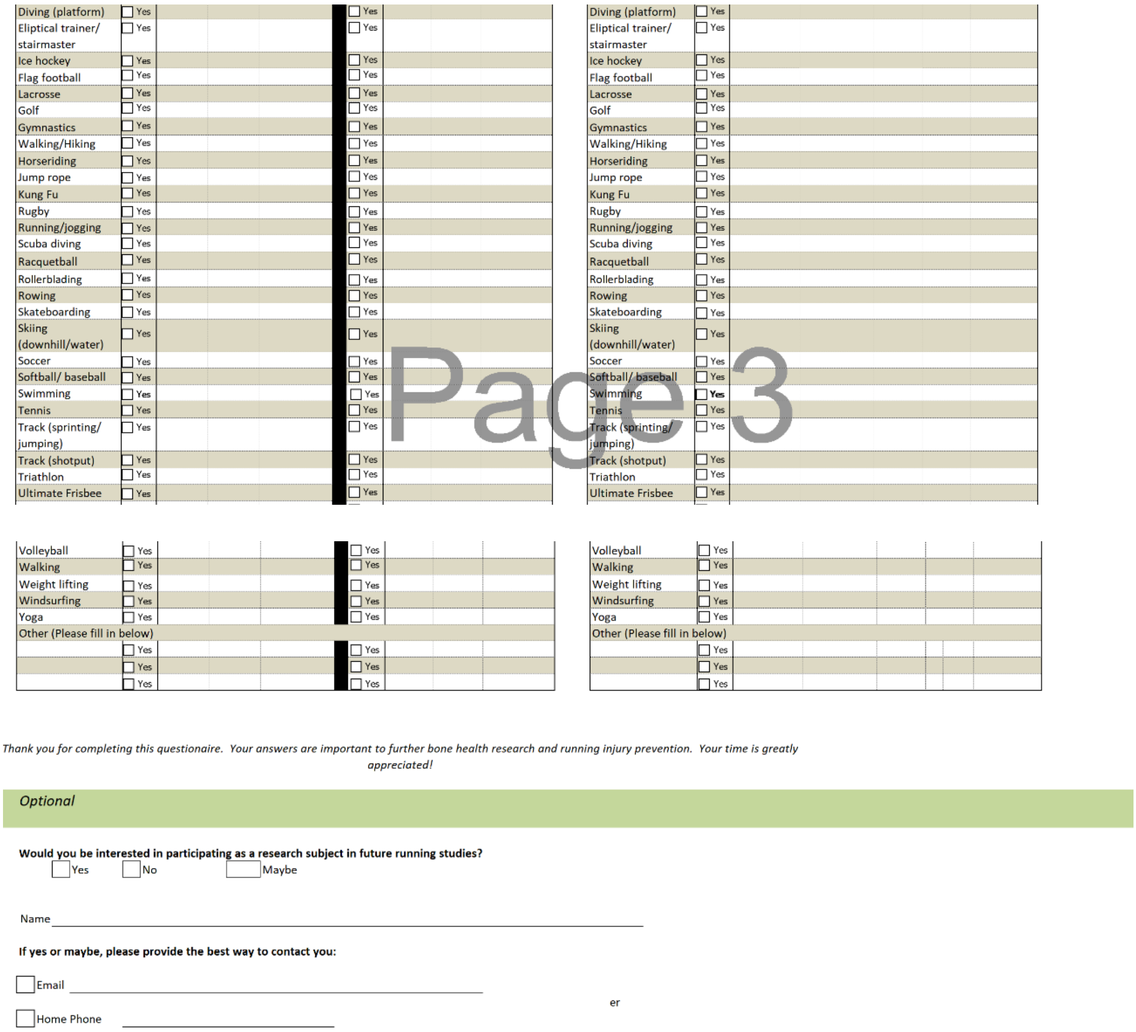
		Current	***3 months before injury**
4.8a	**How many miles a week do you typically run?**		
4.8b	**On average, how many hours per week do you run?**		
4.9	**How many days do you run per week?**	-Select one-	-Select one-
4.10	**How many months do you run out of the year?**	-Select one-	-Select one-
4.11	**What is your current 5k time (mm:ss)?**	:	:
4.12	**What is your estimated 1 mile run time (maximal effort) (mm:ss)?**	:	:
4.13	**Do you run everyday?**	-Select one-	-Select one-
4.14	**Do you run more than one time per day?**	-Select one-	-Select one-

☐ Cell Phone ____________________

☐ Work Phone ____________________

Contact information will be used only to identify potential particpants for future research studies. By providing your contact information, you agree to future contact from study investigators regarding upcoming studies; however, you are under no obligation to participate.

Thank you for your interest in advancing knowledge of running performance!



Diving (platform)	☐ Yes		☐ Yes
Eliptical trainer/ stairmaster	☐ Yes		☐ Yes
Ice hockey	☐ Yes		☐ Yes
Flag football	☐ Yes		☐ Yes
Lacrosse	☐ Yes		☐ Yes
Golf	☐ Yes		☐ Yes
Gymnastics	☐ Yes		☐ Yes
Walking/Hiking	☐ Yes		☐ Yes
Horseriding	☐ Yes		☐ Yes
Jump rope	☐ Yes		☐ Yes
Kung Fu	☐ Yes		☐ Yes
Rugby	☐ Yes		☐ Yes
Running/jogging	☐ Yes		☐ Yes
Scuba diving	☐ Yes		☐ Yes
Racquetball	☐ Yes		☐ Yes
Rollerblading	☐ Yes		☐ Yes
Rowing	☐ Yes		☐ Yes
Skateboarding	☐ Yes		☐ Yes
Skiing (downhill/water)	☐ Yes		☐ Yes
Soccer	☐ Yes		☐ Yes
Softball/ baseball	☐ Yes		☐ Yes
Swimming	☐ Yes		☐ Yes
Tennis	☐ Yes		☐ Yes
Track (sprinting/ jumping)	☐ Yes		☐ Yes
Track (shotput)	☐ Yes		☐ Yes
Triathlon	☐ Yes		☐ Yes
Ultimate Frisbee	☐ Yes		☐ Yes

Diving (platform)	☐ Yes	
Eliptical trainer/ stairmaster	☐ Yes	
Ice hockey	☐ Yes	
Flag football	☐ Yes	
Lacrosse	☐ Yes	
Golf	☐ Yes	
Gymnastics	☐ Yes	
Walking/Hiking	☐ Yes	
Horseriding	☐ Yes	
Jump rope	☐ Yes	
Kung Fu	☐ Yes	
Rugby	☐ Yes	
Running/jogging	☐ Yes	
Scuba diving	☐ Yes	
Racquetball	☐ Yes	
Rollerblading	☐ Yes	
Rowing	☐ Yes	
Skateboarding	☐ Yes	
Skiing (downhill/water)	☐ Yes	
Soccer	☐ Yes	
Softball/ baseball	☐ Yes	
Swimming	☐ Yes	
Tennis	☐ Yes	
Track (sprinting/ jumping)	☐ Yes	
Track (shotput)	☐ Yes	
Triathlon	☐ Yes	
Ultimate Frisbee	☐ Yes	

Volleyball	☐ Yes		☐ Yes
Walking	☐ Yes		☐ Yes
Weight lifting	☐ Yes		☐ Yes
Windsurfing	☐ Yes		☐ Yes
Yoga	☐ Yes		☐ Yes
Other (Please fill in below)			
	☐ Yes		☐ Yes
	☐ Yes		☐ Yes
	☐ Yes		☐ Yes

Volleyball	☐ Yes	
Walking	☐ Yes	
Weight lifting	☐ Yes	
Windsurfing	☐ Yes	
Yoga	☐ Yes	
Other (Please fill in below)		
	☐ Yes	
	☐ Yes	
	☐ Yes	

Thank you for completing this questionaire. Your answers are important to further bone health research and running injury prevention. Your time is greatly appreciated!

Optional

Would you be interested in participating as a research subject in future running studies?

☐ Yes ☐ No ☐ Maybe

Name ____________________

If yes or maybe, please provide the best way to contact you:

☐ Email ____________________

er

☐ Home Phone ____________________

Appendix F: Tabulated Participant information

P No	Age	Gender	Height	Weight	Leg length	DOM leg	Injury history	End HR	Run Time	Interval pace	RtF pace	End RPE	Activity	Est. 5k
1	54	M	177.5	85.3	93	L	Mixed	170	10	14	12.5	8.5	Tri	22.5
2	54	M	174	72.1	88	R	Mixed	170	10	14.2	12.8	10	Tri	22.8
3	42	M	175	69.3	89	R	L.ITB	161	14	16.2	13.7	9	Tri	21
4	42	M	182.6	69.3	96	R	Mixed	198	13	15.8	13.3	9	Tri	23
5	38	M	172	72.1	86	R	Mixed	193	10	16.3	14.8	10	Tri	19.3
6	37	F	160	52	82	R	L.Knee	166	8	16.1	13.6	9	Tri	21.3
7	21	M	174.5	62.7	83	R	L.Knee	161	30	15.7	13.2	8	Runner	17
8	24	M	184	93.7	89	R	R.ACL	189	8	15.2	13.7	9	Football	25
9	36	M	175	82	83	L	Mixed	164	18	15.1	13.6	8	Runner	21
10	46	M	174	72.5	83	R	L. HS	171	9	15.9	14.4	9	Tri	21
11	48	M	182.4	93.5	86	R	Mixed	161	9	15.3	13.8	9	Mixed	23.5
12	50	M	185	84	90	R	Mixed	174	10	15	13.5	10	Tri	21.5
13	66	F	150	56.3	76	R	Mixed	168	15	11.5	10	9	Tri	29
14	34	M	187	82.7	89	R	Mixed	178	12	14.9	13.4	9	Runner	21.3
15	39	M	173	73.5	82	R	Mixed	187	9.5	16	14.5	9.5	Football	20

Appendix G: Test-Retest Ethics Approval

University of Central Lancashire
Preston PR1 2HE
01772 201201
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17 June 2021

Jim Richards / John Kiely
School of Sport and Health Sciences
University of Central Lancashire

Dear Jim / John

Re: STEMH Ethics Review Panel Application
Unique Reference Number: **STEMH 184 CA Addition 11 06 21**

The Health Ethics Review Panel has approved your proposed amendment to your application ‘Coordinative variability over the course of a short moderately fatiguing run’.

As part of your approval please provide the Ethics, Integrity and Governance Unit regular feedback on the conditions which prevail to ensure the research team are adhering to the regulatory conditions imposed as a result of COVID-19.

Yours sincerely

Simon Alford
Deputy Vice-Chair
Health Ethics Review Panel

Appendix H: Test-Retest Smallest Worthwhile Changes for all experimental metrics

The Smallest Worthwhile Change (SWC) was calculated, as per the recommendations of Hopkins (2000), as 1.5 times the typical error. Typical error is estimated by calculating the standard deviation of the test-retest data obtained under conditions where true scores are not expected to change (Hopkins, 2000). The acceleration test-retest data, for 3 specifically recruited participants, was collected for 2 similarly paced 60 second intervals executed 10 minutes apart and in the absence of any intermediate fatiguing activity. Typical error, limits of agreement and SWC were calculated for all local accelerations, local jerk, shock attenuation and resultant accelerations metrics and applied to the experimental data. In the charts, below, Typical Error is abbreviated as TE and Smallest worthwhile Change as SWC.

Acceleration metrics

<table>
<tr><th colspan="10">Tibial Accelerations</th></tr>
<tr><th>Location</th><th>Tib X max</th><th>Tib X min</th><th>Tib X ROM</th><th>Tib Y max</th><th>Tib Y min</th><th>Tib Y ROM</th><th>Tib Z max</th><th>Tib Z min</th><th>Tib Z ROM</th></tr>
<tr><td>TE</td><td>0.188</td><td>0.171</td><td>0.147</td><td>0.103</td><td>0.136</td><td>0.113</td><td>0.139</td><td>0.091</td><td>0.203</td></tr>
<tr><td>SWC</td><td>0.282</td><td>0.256</td><td>0.220</td><td>0.154</td><td>0.203</td><td>0.169</td><td>0.208</td><td>0.137</td><td>0.305</td></tr>
<tr><th colspan="10">Pelvic Accelerations</th></tr>
<tr><th>Location</th><th>Pel X Max</th><th>Pel X Min</th><th>Pel X ROM</th><th>Pel Y Max</th><th>Pel Y Min</th><th>Pel Y ROM</th><th>Pel Z Max</th><th>PelZ Min</th><th>Pel Z ROM</th></tr>
<tr><td>TE</td><td>0.049</td><td>0.105</td><td>0.184</td><td>0.082</td><td>0.198</td><td>0.216</td><td>0.070</td><td>0.016</td><td>0.056</td></tr>
<tr><td>SWC</td><td>0.074</td><td>0.158</td><td>0.276</td><td>0.123</td><td>0.296</td><td>0.324</td><td>0.148</td><td>0.034</td><td>0.118</td></tr>
<tr><th colspan="10">Thoracic Accelerations</th></tr>
<tr><th>Location</th><th>Thor X Max</th><th>Thor X Min</th><th>Thor X ROM</th><th>Thor Y Max</th><th>Thor Y Min</th><th>Thor Y ROM</th><th>Thor Z Max</th><th>Thor Z Min</th><th>Thor Z ROM</th></tr>
<tr><td>TE</td><td>0.056</td><td>0.065</td><td>0.119</td><td>0.056</td><td>0.100</td><td>0.124</td><td>0.026</td><td>0.031</td><td>0.100</td></tr>
<tr><td>SWC</td><td>0.083</td><td>0.098</td><td>0.178</td><td>0.083</td><td>0.149</td><td>0.187</td><td>0.039</td><td>0.047</td><td>0.150</td></tr>
</table>

Jerk metrics

Tibial Jerk									
Location	Tib X max	Tib X min	Tib X ROM	Tib Y max	Tib Y min	Tib Y ROM	Tib Z max	Tib Z min	Tib Z ROM
TE	196.519	117.845	230.830	455.721	329.156	689.472	231.642	327.259	527.265
SWC	294.778	176.767	346.246	683.582	493.733	1034.208	347.464	490.889	790.897
Pelvic Jerk									
Location	Pel X Max	Pel X Min	Pel X ROM	Pel Y Max	Pel Y Min	Pel Y ROM	Pel Z Max	PelZ Min	Pel Z ROM
TE	51.048	92.618	194.449	143.153	51.846	108.317	14.653	10.843	25.489
SWC	76.572	138.927	291.673	214.730	77.769	162.476	21.980	16.264	38.234
Thoracic Jerk									
Location	Thor X Max	Thor X Min	Thor X ROM	Thor Y Max	Thor Y Min	Thor Y ROM	Thor Z Max	Thor Z Min	Thor Z ROM
TE	21.958	82.619	145.011	48.626	62.128	108.317	99.594	154.311	227.194
SWC	32.937	123.928	217.517	72.939	93.193	162.476	149.391	231.466	340.791

Shock Attenuation metrics

Shock Attenuation		
Location	TE	SWC
Tib to Pel	7.661	11.492
Tib to Thor	8.782	13.172

Resultant Acceleration metrics

RESULTANT ACCELERATIONS		
Location	TE	SWC
Tibias	0.099	0.149
Pelvis	0.138	0.206
Thorax	0.117	0.176

Appendix I: Local Acceleration incidences of significance between pre- and post-fatigue conditions

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Left Tibia										
Positive increase	**7**	**0**	**8**	**5**	**0**	**7**	**3**	**3**	**5**	**38**
No change	**5**	**5**	**7**	**10**	**5**	**7**	**9**	**5**	**7**	**60**
Negative increase	**3**	**10**	**0**	**0**	**10**	**1**	**3**	**7**	**3**	**37**
Right Tibia										
Positive increase	**7**	**0**	**9**	**4**	**3**	**3**	**3**	**1**	**5**	**35**
No change	**4**	**6**	**4**	**6**	**8**	**5**	**8**	**7**	**8**	**56**
Negative increase	**4**	**9**	**2**	**5**	**4**	**7**	**4**	**7**	**2**	**44**
Pelvis										
Positive increase	5	4	3	1	8	1	5	8	1	36
No change	4	8	5	7	4	6	8	7	8	57
Negative increase	6	3	7	7	3	8	2	0	6	39
Thorax										
Positive increase	2	6	4	1	9	3	4	4	7	40
No change	5	4	4	6	3	5	7	7	4	45
Negative increase	8	5	7	8	3	7	4	4	4	50

Appendix J: Local Jerk incidences of significance between pre- and post-fatigue conditions

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Left Tibia										
Positive increase	4	2	5	7	3	7	5	6	3	**43**
No change	8	7	8	7	6	5	5	6	5	**45**
Negative increase	3	6	2	1	6	3	5	3	7	**36**
Right Tibia										
Positive increase	7	4	8	4	3	4	5	4	5	**44**
No change	7	8	4	8	8	8	5	8	9	**65**

Negative increase	1	3	3	3	4	3	5	3	4	**29**
Pelvis										
Positive increase	4	3	5	6	3	5	4	1	5	**36**
No change	9	9	7	5	9	5	8	9	9	**70**
Negative increase	2	3	3	4	3	5	3	5	1	**29**
Thorax										
Positive increase	6	4	6	7	5	8	6	4	6	**52**
No change	5	5	5	6	3	4	4	3	4	**39**
Negative increase	4	6	4	2	7	3	5	8	5	**44**

Appendix K: Local acceleration incidences of significance across quartiles

Left tibia: Significant change across quartiles

Direction of change	**Vertical**			**Mediolateral**			**Anteroposterior**			**Incidence of change**
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	7	0	5	1	3	3	4	4	4	31
No change	6	10	9	10	9	9	8	6	8	75
Negative increase	2	5	1	4	3	3	3	5	3	29
Quartile 2 to 3										
Positive increase	4	0	7	5	2	6	3	2	3	32
No change	9	9	5	7	11	7	10	11	11	80
Negative increase	2	6	3	3	2	2	2	2	1	23
Quartile 3 to 4										
Positive increase	3	1	6	1	0	1	3	1	3	19
No change	10	9	7	12	11	13	9	12	8	90
Negative increase	1	4	1	1	3	0	2	1	3	16

Right tibia: Significant change across quartiles

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	5	2	5	5	2	6	5	2	7	39
No change	9	7	8	8	9	8	4	8	5	66
Negative increase	1	6	3	2	3	1	6	5	3	30
Quartile 2 to 3										
Positive increase	5	0	4	4	1	5	3	1	2	25
No change	9	13	11	9	10	8	10	12	11	93
Negative increase	1	2	0	2	4	2	2	2	2	17
Quartile 3 to 4										
Positive increase	5	0	5	2	2	1	2	1	4	22
No change	9	12	9	11	11	11	10	11	8	92
Negative increase	0	2	0	1	1	2	2	2	2	12

Pelvis: Significant change across quartiles

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Upward	Downward	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	4	2	2	3	0	2	4	3	3	23
No change	7	12	13	10	14	12	7	10	8	92
Negative increase	4	1	0	2	1	1	4	2	4	19
Quartile 2 to 3										
Positive increase	5	0	6	1	0	3	4	1	5	25
No change	10	10	9	14	14	12	7	12	7	95
Negative increase	0	5	0	0	1	0	4	2	3	15
Quartile 3 to 4										
Positive increase	2	2	2	1	1	2	4	2	2	18
No change	11	11	11	13	13	11	8	11	11	100
Negative increase	1	1	1	0	0	1	2	1	1	8

Thorax: Significant change across quartiles

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Accel	Decel	Range	Lateral	
Quartile 1 to 2										
Positive increase	5	2	7	5	1	6	4	4	5	39
No change	5	7	6	9	7	8	6	10	7	65
Negative increase	5	6	2	1	7	1	5	1	3	31
Quartile 2 to 3										
Positive increase	5	2	5	4	0	7	3	2	4	32
No change	10	8	8	10	8	8	7	12	6	77
Negative increase	0	5	2	1	7	0	5	1	5	26
Quartile 3 to 4										
Positive increase	0	2	3	2	5	1	3	2	2	20
No change	11	9	9	11	8	9	10	9	12	88
Negative increase	3	3	2	1	1	4	1	3	0	18

Appendix L: Local jerk incidences of significance across quartiles

Left tibia: Significant change across quartiles

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	7	0	5	1	3	3	4	4	4	31
No change	6	10	9	10	9	9	8	6	8	75
Negative increase	2	5	1	4	3	3	3	5	3	29
Quartile 2 to 3										
Positive increase	4	0	7	5	2	6	3	2	3	32
No change	9	9	5	7	11	7	10	11	11	80
Negative increase	2	6	3	3	2	2	2	2	1	23
Quartile 3 to 4										
Positive increase	3	1	6	1	0	1	3	1	3	19

No change	10	9	7	12	11	13	9	12	8	90
Negative increase	1	4	1	1	3	0	2	1	3	16

Right tibia: Significant change across quartiles

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	5	1	5	4	1	5	4	1	2	28
No change	8	12	9	10	10	9	8	12	10	88
Negative increase	2	2	1	1	4	1	3	2	3	19
Quartile 2 to 3										
Positive increase	3	1	4	4	2	4	3	2	4	27
No change	11	12	10	10	10	9	11	11	11	95
Negative increase	1	2	1	1	3	2	1	2	0	13
Quartile 3 to 4										
Positive increase	1	2	3	2	4	2	2	3	0	19
No change	11	10	9	10	10	8	10	12	12	92
Negative increase	3	3	3	3	1	5	3	0	3	24

Pelvis: Significant change across quartiles

Direction of change	Vertical			Mediolateral			Anteroposterior			Incidence of change
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	1	2	2	4	1	4	3	2	2	21
No change	13	12	10	7	9	9	10	12	11	93
Negative increase	1	1	3	4	5	2	2	1	2	21
Quartile 2 to 3										
Positive increase	5	2	4	4	3	3	3	3	3	30
No change	10	9	10	11	9	12	11	7	9	88
Negative increase	2	4	1	0	3	0	1	5	3	19
Quartile 3 to 4										
Positive increase	0	4	2	2	3	2	3	2	2	20
No change	11	10	11	10	9	10	10	12	11	94

Negative increase	4	1	2	3	3	3	2	1	2	21

Thorax: Significant change across quartiles

Direction of change	**Vertical**			**Mediolateral**			**Anteroposterior**			**Incidence of change**
	Accel	Decel	Range	Lateral	Medial	Range	Anterior	Posterior	Range	
Quartile 1 to 2										
Positive increase	12	0	7	4	0	3	0	2	2	30
No change	2	8	8	9	9	10	11	10	10	77
Negative increase	1	7	0	2	6	2	4	3	3	28
Quartile 2 to 3										
Positive increase	4	4	3	4	1	3	4	1	4	28
No change	8	9	7	10	10	11	9	10	10	84
Negative increase	3	2	5	1	4	1	2	4	1	23
Quartile 3 to 4										
Positive increase	3	3	3	0	4	0	0	3	1	17
No change	11	10	10	12	11	11	12	11	11	99
Negative increase	1	2	2	3	0	4	3	1	3	19

Appendix M: Occasions when the difference between Shock Attenuation quartile means exceeded the Smallest Worthwhile Change for shock attenuation metrics

	SA – Tibia to Pelvis			SA – Tibia to Thorax		
Runner	**1 To 2**	**2 To 3**	**3 To 4**	**1 To 2**	**2 To 3**	**3 To 4**
1	🟩	🟥	🟨	🟩	🟩	🟨
2	🟨	🟨	🟨	🟨	🟨	🟨
3	🟨	🟨	🟨	🟨	🟨	🟥
4	🟨	🟨	🟨	🟨	🟨	🟥
5	🟨	🟨	🟨	🟨	🟨	🟨
6	🟨	🟨	🟨	🟨	🟨	🟨

7	🟧	🟧	🟧	🟥	🟧	🟧
8	🟧	🟧	🟧	🟧	🟧	🟥
9	🟧	🟧	🟧	🟧	🟧	🟧
10	🟧	🟧	🟩	🟧	🟧	🟥
11	🟧	🟧	⬛	🟧	🟧	⬛
12	🟧	🟧	🟧	🟧	🟧	🟧
13	🟧	🟧	🟧	🟧	🟩	🟩
14	🟧	🟧	🟧	🟧	🟧	🟧
15	🟧	🟧	🟧	🟧	🟧	🟥

Appendix N: Occasions when the difference between Shock Attenuation quartile means achieved significance

	SA – Tibia to Pelvis			SA – Tibia to Thorax		
Runner	**1 To 2**	**2 To 3**	**3 To 4**	**1 To 2**	**2 To 3**	**3 To 4**
1	🟥	🟩	🟥	🟥	🟥	🟥
2	🟧	🟩	🟥	🟧	🟧	🟥
3	🟥	🟧	🟧	🟧	🟧	🟧
4	🟩	🟥	🟩	🟧	🟥	🟥
5	🟧	🟥	🟥	🟥	🟥	🟧
6	🟧	🟥	🟧	🟧	🟧	🟥
7	🟥	🟩	🟧	🟩	🟥	🟧
8	🟥	🟩	🟧	🟧	🟩	🟧
9	🟧	🟧	🟧	🟧	🟧	🟧
10	🟧	🟧	🟥	🟧	🟧	🟧
11	🟩	🟩	⬛	🟩	🟥	⬛
12	🟧	🟥	🟧	🟩	🟩	🟧

13	🟧	🟧	🟧	🟧	🟧	🟧
14	🟥	🟧	🟧	🟧	🟩	🟧
15	🟧	🟧	🟥	🟧	🟩	🟧

Appendix O: Inter-measure comparison, using incidence of significance, across quartile transitions

Measure		**Changes between Q1 & Q2**	**Changes between Q2 & Q3**	**Changes between Q3 & Q4**
Local Accelerations	**Tibias**	128	102	70
	Pelvis & thorax	116	97	68
Local Jerk	**Tibias**	97	106	77
	Pelvis & thorax	95	100	77
Shock Attenuation	**Tibia to pelvis**	7	9	6
	Tibia to thorax	5	9	5
Resultant Accelerations	**Tibias**	14	15	9
	Pelvis & thorax	9	13	11
Unit increases to Rates of Perceived Exertion per transition		28.5	25	16.5