



# **Good Vibrations: Real-Time Haptic Feedback Gait Retraining to Reduce Tibial Acceleration in Runners.**

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## Thesis Abstract

Overuse injuries are common in runners, with tibial fatigue fractures (TFF) being particularly severe in nature and challenging to manage. Acceleration of the tibia at foot-ground contact, measured from accelerometers attached over the distal bone, can be used as a surrogate of lower extremity loading in runners. Elevated tibial acceleration (TA) is a risk factor for TFF. The primary question of this thesis is, what is the effectiveness of a real-time haptic feedback gait retraining intervention in reducing TA in runners. A review of the literature describing the measurement of, and outlining the factors that can affect, the measurement of TA was carried out in conjunction with an experimental study to assess the reliability and variability of these measurements. TA was measured from eighty-five uninjured runners to establish a representative database at four different running speeds. Values ranged from 4.5 g to 20.6 g, and a regression analysis revealed that for typical runners, every 0.1 m/s increase in running velocity resulted in a 0.38 g increase in TA. There were considerable individual variations to this trend. These data also served to identify runners who might be classified as high-impact based on their TA magnitude. Together these studies established the parameters of assessment for a subsequent intervention. An in-depth systematic review of augmented feedback to reduce lower extremity loading in runners at risk of tibial fatigue fracture found moderate evidence to support short-term reductions in tibial acceleration using visual feedback modalities. No studies have exclusively used haptics as a feedback modality with runners. There was also moderate evidence to suggest that eight sessions over two weeks was an appropriate stimulus dose, and that feedback withdrawal may be important to reduce the reliance on feedback. A final laboratory-based intervention study sought to investigate the effectiveness of a real-time haptic feedback system on 18 high TA runners. All but one runner reduced their TA immediately post-intervention. At the group level, when running on a treadmill a 50% reduction in TA was observed post-intervention, and 41% after 4-weeks. The reductions in TA were 28% and 17% at these same timepoints when running over ground. 61% of runners who completed the feedback programme and returned for a follow-up assessment were classed as positive responders to the intervention. The dominant strategies used by runners to reduce their TA were to adopt a higher cadence, while reducing their foot impact velocity and leg stiffness during initial stance. There was a high degree of individual variability in the mechanical strategies used, highlighting the need for a personalised, data-driven approach to understanding the response of each runner. The haptic feedback intervention used in this study appears to be as effective, but less invasive and expensive, compared to other more established modalities, such as visual feedback. This new approach to movement retraining has the potential to revolutionise the way runners engage in gait retraining with the next steps taking them out of the laboratory and into a normal training environment.

# Table of Contents

<b>LIST OF FIGURES.....</b>	<b>6</b>
<b>LIST OF TABLES.....</b>	<b>7</b>
<b>LIST OF COMMON ABBREVIATIONS .....</b>	<b>8</b>
<b>LIST OF CO-AUTHORED WORKS ARISING FROM THE DOCTORAL THESIS .....</b>	<b>9</b>
<b>CANDIDATE CONTRIBUTIONS TO CO-AUTHORED PUBLICATIONS.....</b>	<b>11</b>
<b>ATTESTATION OF AUTHORSHIP.....</b>	<b>12</b>
<b>SHARED DATA .....</b>	<b>13</b>
<b>ETHICAL APPROVAL.....</b>	<b>13</b>
<b>ACKNOWLEDGEMENTS .....</b>	<b>14</b>
<b>CHAPTER 1: BACKGROUND, RATIONALE AND SIGNIFICANCE OF THE STUDY.....</b>	<b>16</b>
OVERVIEW .....	16
1.0 BACKGROUND.....	17
1.1 RATIONALE AND SIGNIFICANCE OF THE RESEARCH .....	23
1.2 OBJECTIVE, AIMS AND STRUCTURE OF THE THESIS.....	24
1.3 SPECIFIC AIMS OF THE CHAPTERS 2-7.....	26
1.4 LIMITATIONS OF THE THESIS .....	28
<b>CHAPTER 2: THE MEASUREMENT OF TIBIAL ACCELERATION IN RUNNERS. A REVIEW OF THE FACTORS THAT CAN AFFECT TIBIAL ACCELERATION DURING RUNNING AND EVIDENCE-BASED GUIDELINES FOR ITS USE.....</b>	<b>30</b>
2 PREFACE.....	30
2.0 INTRODUCTION .....	31
2.1 METHODS.....	32
2.2 CHAPTER 2 NOVEL CONTRIBUTIONS .....	52
<b>CHAPTER 3: THE ONE-WEEK AND SIX-MONTH RELIABILITY AND VARIABILITY OF THREE-DIMENSIONAL TIBIAL ACCELERATION IN RUNNERS.....</b>	<b>53</b>
3 PREFACE.....	53
3.0 INTRODUCTION .....	54
3.1 METHODS.....	55
3.2 RESULTS .....	58
3.3 DISCUSSION AND IMPLICATIONS .....	62
3.4 CONCLUSION .....	63
3.5 CHAPTER 3 NOVEL CONTRIBUTIONS .....	64

<b>CHAPTER 4:</b>	<b>65</b>
<b>THE INFLUENCE OF RUNNING VELOCITY ON RESULTANT TIBIAL ACCELERATION IN RUNNERS.</b>	<b>65</b>
4 PREFACE.....	65
4.0 INTRODUCTION .....	66
4.1 METHODS.....	67
4.2 RESULTS.....	70
4.3 DISCUSSION AND IMPLICATIONS .....	73
4.4 CONCLUSION .....	75
4.5 CHAPTER 4 NOVEL CONTRIBUTIONS .....	76
<b>CHAPTER 5: THE EFFECTIVENESS OF AUGMENTED FEEDBACK TO REDUCE LOWER EXTREMITY LOADING IN RUNNERS AT RISK OF TIBIAL FATIGUE FRACTURE. A SYSTEMATIC REVIEW.</b>	<b>78</b>
5 PREFACE.....	78
5.0 INTRODUCTION .....	79
5.1 METHODS.....	80
5.2 RESULTS.....	83
5.3 DISCUSSION .....	94
5.4 CONCLUSIONS.....	101
5.5 PRACTICAL RECOMMENDATIONS .....	101
5.6 CHAPTER 5 NOVEL CONTRIBUTIONS .....	102
<b>CHAPTER 6: THE EFFECTIVENESS OF REAL-TIME HAPTIC FEEDBACK GAIT RETRAINING FOR REDUCING RESULTANT TIBIAL ACCELERATION WITH RUNNERS.</b>	<b>103</b>
6 PREFACE.....	103
6.0 INTRODUCTION .....	104
6.1 METHODS.....	105
6.2 RESULTS.....	111
6.3 DISCUSSION .....	115
6.4 CONCLUSIONS.....	116
6.5 CHAPTER 6 NOVEL CONTRIBUTIONS .....	117
<b>CHAPTER 7: THE BIOMECHANICAL EFFECTS OF REAL-TIME HAPTIC FEEDBACK GAIT RETRAINING ON RUNNERS WITH HIGH TIBIAL ACCELERATION.</b>	<b>118</b>
7 PREFACE.....	118
7.0 INTRODUCTION .....	119
7.1 METHODS.....	120
7.2 RESULTS.....	124



7.3	DISCUSSION .....	128
7.4	CONCLUSIONS.....	131
7.5	CHAPTER 7 NOVEL CONTRIBUTIONS .....	132
<b>CHAPTER 8: DISCUSSION AND CONCLUSIONS. ....</b>		<b>133</b>
8	CONTEXT AND FRAMEWORK .....	133
8.0	SUMMARY AND MAIN FINDINGS .....	133
8.0.7	SPECIFIC LIMITATIONS OF THE THESIS.....	140
8.1	FUTURE RESEARCH DIRECTIONS.....	142
8.2	CONCLUSIONS.....	143
9	<b>REFERENCES.....</b>	<b>144</b>
10	<b>APPENDICES .....</b>	<b>159</b>
APPENDIX 1. THE ONE-WEEK AND SIX-MONTH RELIABILITY AND VARIABILITY OF THREE-DIMENSIONAL TIBIAL ACCELERATION IN RUNNERS. PUBLISHED IN SPORTS BIOMECHANICS. ....		160
APPENDIX 2. THE INFLUENCE OF RUNNING VELOCITY ON RESULTANT TIBIAL ACCELERATION IN RUNNERS. PUBLISHED IN SPORTS BIOMECHANICS.....		161
APPENDIX 3. THE MEASUREMENT OF TIBIAL ACCELERATION IN RUNNERS. A REVIEW OF THE FACTORS THAT CAN AFFECT TIBIAL ACCELERATION DURING RUNNING AND EVIDENCE BASED GUIDELINES FOR ITS USE. PUBLISHED IN GAIT & POSTURE.....		162
APPENDIX 4. THE EFFECTIVENESS OF REAL-TIME HAPTIC FEEDBACK GAIT RETRAINING FOR REDUCING RESULTANT TIBIAL ACCELERATION WITH RUNNERS. PUBLISHED IN PHYSICAL THERAPY IN SPORT .....		163
APPENDIX 5. THE RELIABILITY AND VARIABILITY OF THREE-DIMENSIONAL TIBIAL ACCELERATION DURING RUNNING. PRESENTED AT THE INTERNATIONAL SOCIETY OF BIOMECHANICS IN SPORTS CONFERENCE .....		164
APPENDIX 6. THE INFLUENCE OF RUNNING VELOCITY ON RESULTANT TIBIAL ACCELERATION IN UNINJURED RUNNERS. PRESENTED AT THE SPORTS MEDICINE AUSTRALIA CONFERENCE. ....		170
APPENDIX 7. HAPTIC GAIT RETRAINING IS EFFECTIVE AT REDUCING TIBIAL IMPACTS DURING RUNNING. PRESENTED AT SPORTS MEDICINE NEW ZEALAND CONFERENCE .....		172
APPENDIX 8. AGREEMENT FOR USE OF SHARED DATA: CHAPTER 3.....		174
APPENDIX 9. ETHICS APPROVAL: CHAPTERS 3 AND 4.....		175
APPENDIX 10. INFORMATION AND CONSENT FORMS: CHAPTER 3 .....		176
APPENDIX 11. ETHICS APPROVAL: CHAPTERS 4, 6 AND 7 .....		180
APPENDIX 12. INFORMATION AND CONSENT FORMS: CHAPTER 4 .....		181
APPENDIX 13. ETHICS AMENDMENT: CHAPTERS 6 AND 7.....		184
APPENDIX 14. INFORMATION AND CONSENT FORMS: CHAPTERS 6 AND 7 .....		185
APPENDIX 15. REAL-TIME HAPTIC FEEDBACK SYSTEM AND LabVIEW PROGRAMME DETAILS.....		189
APPENDIX 16. CHAPTER 5: MODIFIED DOWNS AND BLACK SCORE OF METHODOLOGICAL QUALITY.....		192

## List of Figures

FIGURE 1: CONCEPTUAL FRAMEWORK FOR THE ACCUMULATION OF TISSUE STRESS .....	18
FIGURE 2: FLOW OF THESIS THEMES AND CHAPTERS .....	25
FIGURE 3: ACCELEROMETER ATTACHMENT METHODS. ....	56
FIGURE 4: RESULTANT TIBIAL ACCELERATION FROM RUNNING AT 3.0 M/S .....	57
FIGURE 5: INDIVIDUAL TIBIAL ACCELERATION RESULTS FOR BASELINE, ONE WEEK AND SIX MONTHS. ....	60
FIGURE 6: INDIVIDUAL RESULTANT TIBIAL ACCELERATION OBSERVATIONS AT 2.7, 3.0, 3.3 AND 3.7 M/S AND A LINEAR REGRESSION LINE. ....	71
FIGURE 7: EXAMPLE INDIVIDUAL RESULTANT TIBIAL ACCELERATIONS DEPICTING (A) TYPICAL INCREASES (A) LARGE INCREASES (A) INCONSISTENT CHANGES, WITH INCREASING RUNNING VELOCITY.....	72
FIGURE 8: FLOW DIAGRAM OF SYSTEMATIC SEARCH .....	84
FIGURE 9: FLOW CHART DESCRIBING THE STUDY RECRUITMENT SCREENING AND STUDY DESIGN. ....	106
FIGURE 10: BLOCK DIAGRAM OF REAL-TIME HAPTIC FEEDBACK SYSTEM. ....	108
FIGURE 11: TYPICAL PARTICIPANT RUNNING ON THE TREADMILL WHILE RECEIVING REAL-TIME HAPTIC FEEDBACK VIA CUSTOM SOFTWARE. ....	109
FIGURE 12: GAIT RETRAINING SCHEDULE .....	110
FIGURE 13: RESULTANT TIBIAL ACCELERATION DURING TREADMILL RUNNING AT PRE-INTERVENTION, POST-INTERVENTION AND 4-WEEKS POST-INTERVENTION ASSESSMENTS.....	113
FIGURE 14: RESULTANT TIBIAL ACCELERATION DURING OVERGROUND RUNNING AT PRE-INTERVENTION, POST-INTERVENTION AND 4-WEEKS POST-INTERVENTION ASSESSMENTS.....	114
FIGURE 15: LOWER BODY KINEMATIC MARKER LOCATIONS .....	122
FIGURE 16: INDIVIDUAL PRE- AND POST-INTERVENTION CHANGES IN (A) RESULTANT TIBIAL ACCELERATION .....	127

## List of Tables

TABLE 1: SUMMARY OF LITERATURE RELATED TO TIBIAL ACCELERATION MEASUREMENT AND ANALYSIS .....	36
TABLE 2 SUMMARY OF LITERATURE RELATED TO INTRINSIC AND EXTRINSIC FACTORS THAT CAN MODIFY TIBIAL ACCELERATIONS .....	45
TABLE 3: PEAK RESULTANT ACCELERATION BETWEEN-SESSION VARIABILITY AND RELIABILITY FOR FOUR RUNNING SPEEDS. ....	61
TABLE 4: PARTICIPANT CHARACTERISTICS USED IN CHAPTER 4.....	68
TABLE 5: RESULTANT TIBIAL ACCELERATION CHARACTERISTICS BY RUNNING VELOCITY.....	71
TABLE 6: LEVELS OF EVIDENCE. ....	82
TABLE 7: PARTICIPANT AND STUDY CHARACTERISTICS FOR CHAPTER 5.....	85
TABLE 8: KEY CLINICAL FINDINGS AND BIOMECHANICAL OUTCOMES.....	91
TABLE 9: RESULTANT TIBIAL ACCELERATION SCREENING THRESHOLDS.....	107
TABLE 10: MULTI-POINT CRITERIA TO DEFINE POSITIVE RESPONDERS TO INTERVENTION.....	111
TABLE 11: PARTICIPANT CHARACTERISTICS USED IN CHAPTER 6.....	111
TABLE 12: GROUP-BASED RESULTANT TIBIAL ACCELERATION RESULTS FOR TREADMILL AND OVER GROUND RUNNING. ....	112
TABLE 13: PARTICIPANT CHARACTERISTICS USED IN CHAPTER 7.....	124
TABLE 14: BIOMECHANICAL VARIABLES BEFORE AND AFTER THE GAIT RETRAINING INTERVENTION. ....	125
TABLE 15: STEP RATE SUB-GROUP ANALYSIS OF BIOMECHANICAL VARIABLES BEFORE AND AFTER THE GAIT RETRAINING INTERVENTION. ..	126
TABLE 16: FOOTSTRIKE ANGLE SUB-GROUP ANALYSIS OF BIOMECHANICAL VARIABLES BEFORE AND AFTER THE GAIT RETRAINING INTERVENTION.....	128

## List of Common Abbreviations

AVLR	average vertical loading rate
CI	confidence interval
COM	centre of mass
ES	effect sizes
FB	feedback
GRF	ground reaction force
IMU	inertial measurement unit
IVLR	instantaneous vertical loading rate
IQR	Interquartile range
k	stiffness
Md	median
m <sub>e</sub>	effective mass
PTA-A or TA-A	peak axial tibial acceleration
PTA-R or TA-R	peak resultant tibial acceleration
RFS	rearfoot strike
TA	tibial acceleration
TFF	tibial fatigue fracture
SD	standard deviation
VIP	vertical impact peak

## List of Co-Authored Works Arising from the Doctoral Thesis

### Peer-reviewed journal publications

#### Manuscripts published

1. Sheerin, K. R., Besier, T. F., Reid, D., & Hume, P. A. (2017). The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners. *Sports Biomechanics*, 17(4) 531-540. <http://doi.org/10.1080/14763141.2017.1371214> (Appendix 1.)
2. Sheerin, K., Besier, T., Reid, D. (2018). The influence of running velocity on resultant tibial acceleration in runners. *Sports Biomechanics*. DOI: 10.1080/14763141.2018.1546890 *Published online ahead of print* (Appendix 2.)
3. Sheerin, K., Reid, D., Besier, T. (2019). The measurement of tibial acceleration in runners. A review of the factors that can affect tibial acceleration during running and evidence based guidelines for its use. *Gait & Posture*. 67, 12–24. <http://doi.org/10.1016/j.gaitpost.2018.09.017> (Appendix 3.)
4. Sheerin, K., Reid, D., Taylor, D., Besier, T. (2020). The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners. *Physical Therapy in Sport* (43), 173-180. DOI:10.1016/j.ptsp.2020.03.001 (Appendix 4.)

#### Manuscripts submitted for publication

1. Sheerin, K., Reid, D., Taylor, D., Besier, T. The biomechanical effects of real-time haptic feedback gait retraining on runners with high tibial acceleration. Submitted to *Journal of Applied Biomechanics*.

#### Manuscripts formatted for submission

2. Sheerin, K., Reid, D., Taylor, D., Besier, T. The effectiveness of augmented feedback to reduce lower extremity loading in runners at risk of tibial fatigue fracture. A systematic review. To be submitted to *Sports Medicine*.

### Peer-reviewed conference presentations

1. Sheerin, K., Besier, T., Reid, D., & Hume, P. (2016). The reliability and variability of three-dimensional tibial acceleration during running. *International Society of Biomechanics in Sports Conference*. Tsukuba, Japan. (Appendix 5.)
2. Sheerin, K., Reid, D., & Besier, T. (2017). The influence of running velocity on resultant tibial acceleration in uninjured runners. *Sports Medicine Australia Conference*. Langkawi, Malaysia. (Appendix 6.)

3. Sheerin, K., Reid, D., & Besier, T. (2017). The influence of running velocity on resultant tibial acceleration in uninjured runners. *Journal of Science & Medicine in Sport*, 20(Supplement 3), 85–86. <http://doi.org/10.1016/j.jsams.2017.09.366> (Appendix 6.)
4. Sheerin, K., Reid, D., Taylor, D. & Besier, T. (2019). Haptic gait retraining is effective at reducing tibial impacts during running. Sports Medicine New Zealand Conference. Dunedin, NZ. (Appendix 7.)

**Invited speaker**

Presented at the 5<sup>th</sup> Biennial Sport and Exercise Physiotherapy New Zealand Symposium, March 2019, Tauranga. Running gait retraining: What, why, how?

## Candidate Contributions to Co-authored Publications

### Chapter 2

Sheerin, K., Reid, D., Besier, T. (2019). The measurement of tibial acceleration in runners. A review of the factors that can affect tibial acceleration during running and evidence based guidelines for its use. <i>Gait &amp; Posture</i> . 67, 12–24.	Sheerin: 90% Reid: 5% Besier: 5%
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### Chapter 3

Sheerin, K. R., Besier, T. F., Reid, D., & Hume, P. A. (2017). The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners. <i>Sports Biomechanics</i> , 17(4) 531-540.	Sheerin: 90% Besier: 5% Reid: 2.5% Hume: 2.5%
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### Chapter 4

Sheerin, K., Besier, T., Reid, D. (2018). The influence of running velocity on resultant tibial acceleration in runners. <i>Sports Biomechanics</i> . Published online ahead of print	Sheerin: 85% Besier: 10% Reid: 5%
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### Chapter 5

<u>Sheerin, K.</u> , Reid, D., Taylor, D., Besier, T. The effectiveness of augmented feedback to reduce lower extremity loading in runners at risk of tibial fatigue fracture. A systematic review. To be submitted to <i>Sports Medicine</i> .	Sheerin: 85% Reid: 10% Taylor: 5% Besier: 5%
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### Chapter 6

<u>Sheerin, K.</u> , Reid, D., Taylor, D., Besier, T. (2020). The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners. <i>Physical Therapy in Sport</i> (43), 173-180.	Sheerin: 80% Reid: 5% Taylor: 5% Besier: 10%
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### Chapter 7

<u>Sheerin, K.</u> , Reid, D., Taylor, D., Besier, T. The biomechanical effects of real-time haptic feedback gait retraining on runners with high tibial acceleration. Submitted to <i>Journal of Applied Biomechanics</i> .	Sheerin: 80% Reid: 5% Taylor: 5% Besier: 10%
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We, the undersigned, hereby agree to the percentages of participation to the chapters identified above:



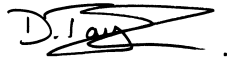
Kelly Sheerin



Duncan Reid



Thor Besier



Denise Taylor



Patria Hume

### **Attestation of Authorship**

I hereby declare that this submission is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person (except where explicitly defined in the acknowledgements), nor material which to a substantial extent has been submitted for the award of any other degree or diploma of a university or other institution of higher learning.

Chapters 2–7 have been submitted (or are in the process of being prepared for submission) for consideration as separate papers for publication in international peer-reviewed journals. The candidate was the main contributor and principal author for each of these papers. All co-authors have approved the inclusion of the papers they were involved in as chapters for this thesis. Individual contributions for these chapters are outlined in the introduction (Chapter 1).



December 2019



## **Shared Data**

Chapter 3 includes data that were collected collaboratively with Mr. Alex Wilson. The subsequent processing and analysis of this data were carried out independently. The agreement outlining this arrangement is provided in Appendix 8.

## **Ethical Approval**

Auckland University of Technology Ethics Committee (AUTEC) granted ethical approval for research carried out in Chapters 3, 4, 5 and 7.

- Ethics Approval #13/51 for Chapter 3 (Appendix 9.)
- Information and Consent Forms: Chapter 3 (Appendix 10.)
- Ethics Approval #15/181 for Chapters 4, 6 and 7 (Appendix 11.)
- Information and Consent Forms: Chapter 4 (Appendix 12.)
- Ethics Amendment: Chapters 6 and 7 (Appendix 13.)
- Information and Consent Forms: Chapters 6 and 7 (Appendix 14.)

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**Ehara taku toa i te toa takitahi, engari he toa takitini**

***My success should not be bestowed onto me alone, as it was not individual success but success of a collective.***

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# **Chapter 1: Background, Rationale and Significance of the Study**

## **Overview**

Section 1 of this chapter contains a review of the relevant background literature for the thesis. The issue of tibial fatigue fracture injuries in runners is defined in Section 1.0, with the issues of surrounding the current management of tibial fatigue fractures outlined in Section 1.0.2. In Section 1.0.3 there is a review of the biomechanical risk factors associated with tibial fatigue fractures, with specific reference to tibial acceleration. In the final parts of the background (Sections 1.0.4-1.0.6), the concepts of running gait retraining and augmented feedback are outlined. Haptics is introduced as a specific modality of feedback (Section 1.0.7), but due to the lack of research on haptics in running, previous studies that have used this form of feedback in walking gait are outlined.

The rationale and significance of this research, along with the thesis aims are detailed in Sections 1.1 and 1.2, respectively. Finally, the overall thesis structure is provided in Section 1.3.

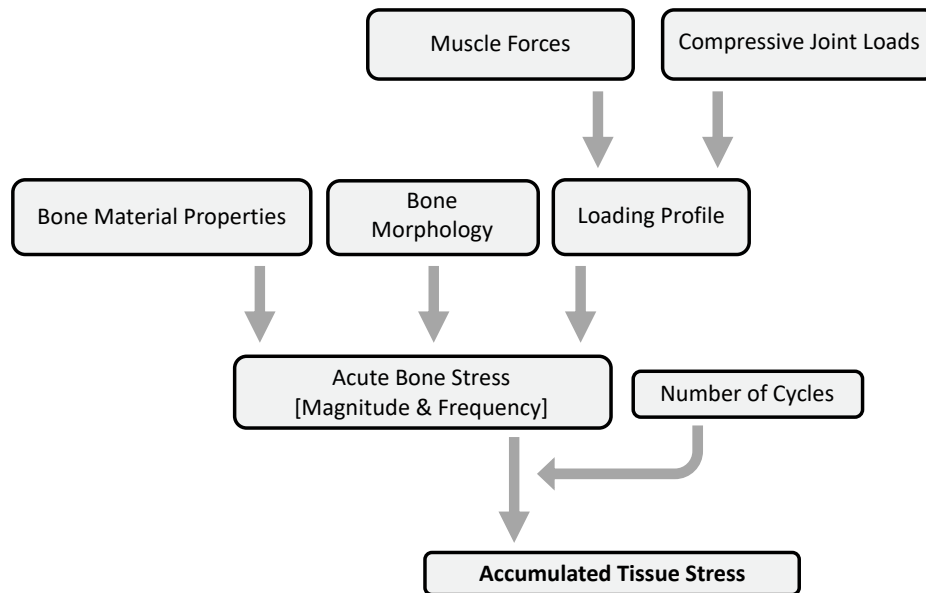
## **1.0 Background**

### **1.0.1 Defining the Issue**

Running is a popular form of exercise, an integral component of many sports, and a sport unto itself. Running is also known to have a positive influence on fitness, as well as physical and mental health [1,2]. Runners have been shown to have a 30% lower overall risk of mortality, and a 45% lower risk of death due to heart disease, compared with non-runners [1]. Despite these positive outcomes, running is associated with a particularly high annual injury incidence [3,4] ranging between 19% and 79%, depending on the study population and definition of injury [4]. The increasing participation of non-professionals in endurance sports such as marathon running, has also led to an increase in stress injuries amongst this group [5]. Within this large number of injuries, fatigue fractures (commonly termed stress fractures) are among the top ten injuries sustained in runners [6], at a rate of 0.7 per 1000 hours of training, with tibial fatigue fractures (TFFs) accounting for between 35% and 49% of all fatigue fractures in runners [7,8]. The most common site for TFFs is the antero-medial border, at the junction of the middle and distal thirds [8,9], which often corresponds to site of the narrowest bone width [10].

### **1.0.2 Mechanisms of Tibial Fatigue Fracture**

While there are many contributing factors to the development of TFFs [11], from a mechanical perspective, TFFs develop due to the accumulation of bone microdamage from repeated submaximal loading which leads to mechanical fatigue [12,13]. Muscle forces and compressive joint loads generated during running act on the tibia bone [14], which will have individually characteristic morphology and material properties. The accumulated tissue stress in the tibia bone will be the product of the acute bone stress generated by the forces and the number of cycles the runner is exposed to (i.e. their running cadence and overall distance) (Figure 1) [15]. Under normal physiological conditions this accumulated tissue stress is managed by bone remodeling [16,17]. However, in situations where damage accumulation exceed bone remodeling, and bone stress continues, fatigue fracture can occur [18,19].



**Figure 1: Conceptual framework for the accumulation of tissue stress**

### 1.0.3 Current Management of Tibial Fatigue Fractures

In addition to causing considerable pain, fatigue fractures often lead to lost days training due to the need for reduced impact (usually 4-8 weeks) to the affected area, and subsequent rehabilitation. Conservative management (rest from weight-bearing activities) is generally considered to be the most effective approach for treating bony injuries [7,20–23]. The recommended rest is often for prolonged periods, and ranges from 3 to 16 weeks for those with TFFs (mean  $8.3 \pm 2.8$  weeks) [20,24]. Some clinicians elect for limb casting as an initial treatment option, with a recommended immobilisation period ranging from 3 to 12 weeks (average of  $6.4 \pm 2.1$  weeks) [24]. In either case, this extended period of reduced or modified activity can result in substantial reductions in cardiovascular and muscular function [25]. Considering the negative repercussions of reduced weight-bearing activity and immobilisation on bone and other tissues, the disruption to active lifestyles, and the occasional ending of athletic careers entirely, the treatment options for runners suffering TFFs are suboptimal. There is clearly a need to develop robust strategies to improve the rehabilitation of TFFs, but more importantly, to develop strategies to prevent their occurrence in the first instance.

### 1.0.4 Biomechanical Risk Factors for Tibial Fatigue Fractures

With respect to the kinetics of running, when the foot strikes the ground, its velocity rapidly decelerates to zero and large ground reaction forces (GRF) are generated [26]. This momentum change produces compressive loading of the lower limbs, and results in an impact shock

transmitted through the musculoskeletal system, with local segment peak accelerations occurring at successively later times [27,28]. While many factors influence bony remodeling and ultimately the manifestation of a fatigue fracture [11], biomechanics and running load dictate the level of mechanical stress on bone during running [29,30]. A number of running kinetic and kinematic factors have been implicated in the increase in loading on the tibia [31], and while the magnitude of the GRFs have not found to be predictive of injury, force loading rates have differentiated healthy runners, from those who have experienced TFFs [32]. The measurement of these variables requires facilities equipped with motion capture systems and force plates. However, with developing small wearable technologies, the measurement of tibial acceleration has become easier as a surrogate measure of lower extremity loading with runners [33,34].

Specifically with regard to TFF, tibial acceleration magnitude in the axial direction (TA-A) has successfully predicted history of TFF in female runners in 70% of cases [35], and when a binary logistic regression model was applied, for every 1g increase in TA-A, the likelihood of having a history of TFF increased by a factor of 1.4 [36]. Additionally, when tracked prospectively, a small sample of runners who went on to sustain TFF demonstrated TA-A magnitudes almost double that of uninjured controls [37]. There are still gaps in our knowledge regarding the use of this measurement tool specifically regarding some of the key considerations and potential limitations, such as the effects of running velocity, technique, fatigue and surface characteristics.

The importance of measuring all components of tibial acceleration which contribute to the overall severity of the lower leg impact was proposed some time ago [38]. More recently some researchers have reported measuring the vector sum from all components of a triaxial accelerometer from runners [39–41]. Otherwise known as the resultant tibial acceleration (TA-R), this provides a single metric representation of these components [41,42], which may be preferable in instances where repeatability of measurement is prioritised and sensor alignment is not strictly controllable, such as in cases where athletes are attaching devices themselves. While it would seem logical that TA-R would be related to other GRF loading variables associated with TFFs in runners, there is only very limited understanding of these relationships [42]. Additionally, if accelerometers are to be used as a viable tool for coaches and clinicians, as well as for intervention studies, verification of measurement reliability is needed. Currently this is limited to a single study of where TA-A is measured from three participants running at a single speed [43]. Additional investigation of the reliability and movement variability of TA-R is required, ideally at a range of speeds representative of those runners' train at.

### **1.0.5 Gait Retraining**

The development of TFFs in runners is multifactorial [44], and if improper mechanics are a contributor, it makes sense to address these factors to ensure that a return to running does not result in a reoccurrence of the injury [45]. Similarly, it also makes sense to address technique related faults placing uninjured runners at risk as an injury prevention strategy [46]. There is a growing body of evidence to alter running technique using gait retraining. It is also possible to target specific biomechanical variables related to injuries, at least for patello-femoral pain syndrome and chronic exertional compartment syndrome [47–49]. Despite these positive indications, it should be noted that creating a lasting change in the motor performance of running is not a trivial undertaking. There are a number of motor learning-related complexities, such as the feedback protocol design, the retraining environment, and the feedback timing and withdrawal, that cannot be ignored [50]. A comprehensive understanding of the underlying motor control and mechanical principles, as well as the feedback to be employed, are essential [51].

### **1.0.6 Augmented Feedback**

In the past runners might have relied on coaches, therapists and sport scientists to provide them with feedback on their technique and their faults. While sometimes effective, the techniques used often stem from a subjective art rather than an objective analysis. Previously most available technologies, such as GPS watches, or accelerometer embedded heart-rate straps and foot-pods, have been limited to providing feedback on pace, cadence and distance, and not variables specifically related to running technique or biomechanics. Recent advancements in computing power and sensor technology have enabled biomechanical variables to be measured and processed at high speed [52]. These metrics can now be integrated with wearable feedback systems to provide real-time training for relatively fast movement patterns.

Augmented feedback is the provision of information from an external source, to supplement the body's own internal feedback, with the aim of enhancing motor learning [51]. This has been shown to positively influence changes to motor performance in runners [47–49]. A number of feedback strategies have been studied, using various forms and combinations of audio and visual feedback [47–49]. Despite the apparent success of these approaches, they are not without limitations. An inherent downside of most visual feedback is the requirement for a fixed screen, thereby limiting retraining to a lab or clinic environment. While audio feedback has greater potential to be integrated into a normal training environment, there are safety concerns when hearing is also required for the sensing of traffic and other hazards that could be injurious for runners.



### **1.0.7 Haptic Feedback**

Tactile and kinesthetic perception conveyed through vibrations or pressure on the skin, or via receptors in muscles and tendons, that allow us to feel the position of our body is known as haptic feedback [50]. Cutaneous feedback is a subset of haptics whereby the focus is on providing sensations of pressure via the skin mechanoreceptors [53,54], which can take the form of various modalities such as skin stretch or vibration. While the feedback used in this research takes the form of vibration cutaneous feedback, the term haptic feedback will be used throughout for clarity.

Vibrotactile motors have been used with good effect in a range of sports [52,55,56], however evidence for their use in influencing motor learning is limited to two rowing studies [53,54]. Haptic feedback has been used more widely in walking gait retraining to reduce knee adduction moments in healthy individuals and those with osteoarthritis [57–59]. Haptic feedback has also been tuned to provide independent feedback about multiple gait parameters within a single session [52,55]. The use of haptic feedback with runners is limited to a single exploratory study with six participants, while they also received simultaneous visual and auditory feedback [56]. Additional motivations for the further development of wearable haptic devices is that they lend themselves to use in normal training environments. They also leave vision and hearing free for other tasks, which is particularly important for safety.

Haptic feedback is typically initiated based on parameter error signals, meaning that learners receive intermittent, rather than continuous feedback. Advanced feedback designs have successfully been used, such as audible beeps scaled to error as feedback for running gait [39], and different patterns of haptic stimuli targeting multiple gait parameters for walking [52]. However, given the paucity of research using haptic feedback in running, a simplified approach was initiated for the purposes of this research. Step-by-step feedback was provided immediately after each foot contact where the peak TA-R exceeded the defined threshold. Additional detail on the haptic feedback used in this thesis is outlined in Appendix 15.

Previous gait retraining studies using haptic feedback have provided the stimuli to the distal lower extremity [52,57,60], and while this has been reported as a successful approach with walking, with the limb velocity and subsequent ground impact shock associated with running, this location was found to be impractical during pilot testing. Specifically, movement and slippage of the feedback device resulted in unwanted interference with the tibial mounted accelerometer. Solutions to these issues could have been sought through further redesign, but it was instead decided to relocate feedback to the wrist in a watch. This change solved the issue of the device moving, but additionally, runners reported during pilot testing that the wrist-watch location was familiar and

intuitive. Research has indicated that vibration stimulus can be well detected on the forearm [58,59], and we also found that the circular arrangement of a watch strap propagated the vibration stimulus around the wrist, rather than isolating it to cutaneous region proximate to the vibrating motor.

## **1.1 Rationale and Significance of the Research**

Despite being the focus of clinicians and researchers for some time, runners of all levels continue to sustain tibial fatigue fracture injuries, which can substantially disrupt training and overall health and wellbeing. The research on biomechanical risk factors is substantial, but few solutions, outside a laboratory environment, have demonstrated the potential to reduce the risk of runners sustaining these injuries.

The research presented in this thesis builds upon previous work to further investigate the measurement of tibial acceleration in runners, specifically with regard to the effects of running velocity on tibial acceleration magnitude, as well as the relationships between tibial acceleration and other key biomechanical variables. The work in this thesis provides the first evaluation of haptics as a sole feedback modality with runners. This new approach to movement retraining has the potential to revolutionise the way runners train. Combining the proven concepts of movement retraining with new technology, there is a potential to empower individuals to modify their own movement patterns/joint load to ensure lasting changes, and minimise the risk of injury in the long-term. This approach has the potential to be as effective, but less invasive and expensive, as other proven approaches, while also opening the door to the retraining gait in a normal training environment.

These studies not only make novel contributions to research, but also have wider implications that could be translated into new technological developments, and practical strategies for runners to prevent future injury occurrence.

## **1.2 Objective, Aims and Structure of the Thesis**

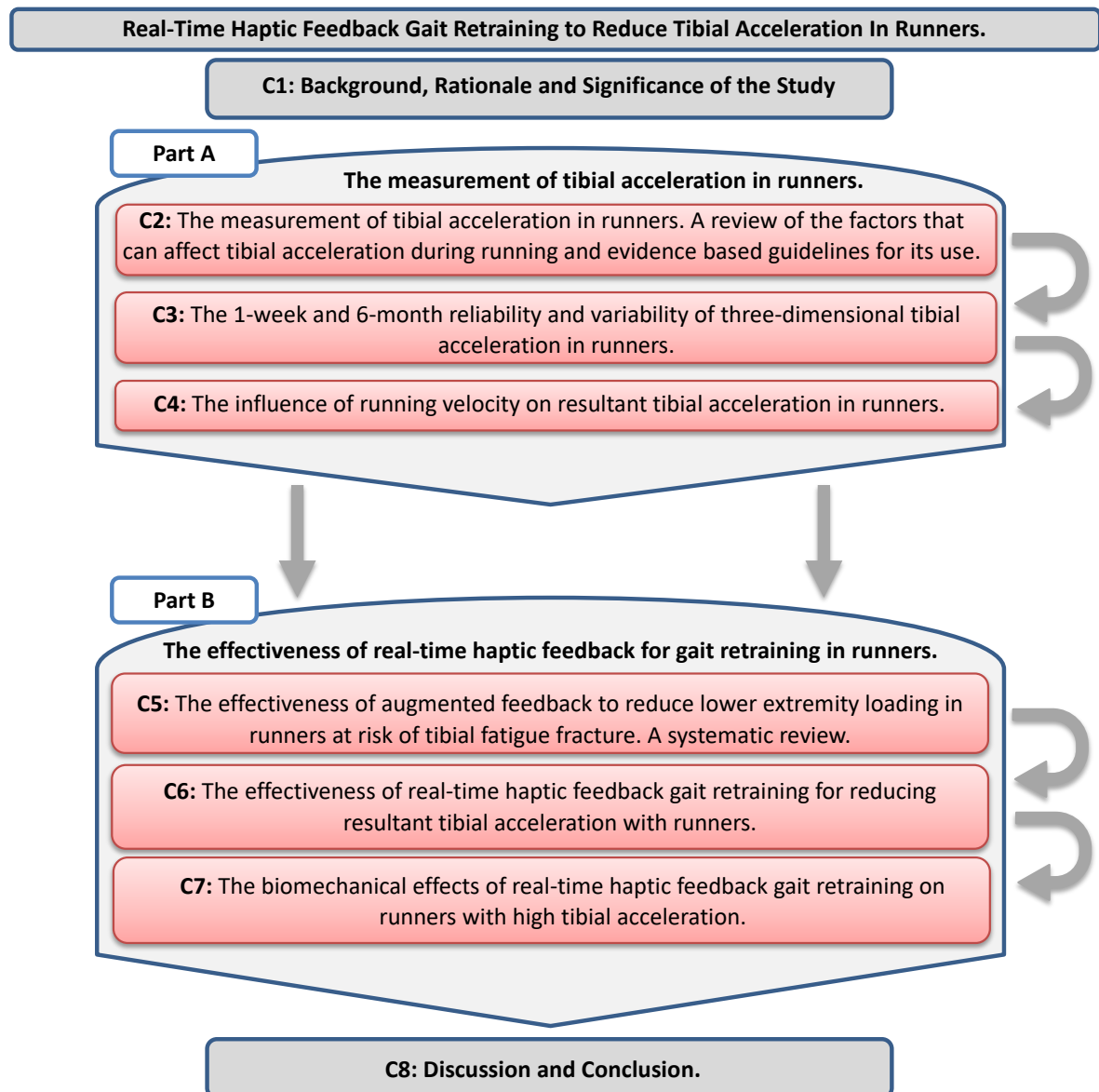
This thesis follows a manuscript structure and comprises six research chapters designed to address the thesis aim and objectives (Figure 2). The manuscripts are presented as they have been submitted to, or accepted for, the target journals, and as such the repetition of some information occurs. Each chapter begins with a preface, and ends with an outline of the novel contributions, which serves to demonstrate the sequential progression and brings together the thesis as a cohesive whole. The referencing format has been standardised in a numerical style, with a single reference list prepared for the thesis.

This thesis had two preliminary objectives, which built towards answering the primary question; 'What is the effectiveness of a real-time haptic feedback gait retraining intervention in reducing TA in runners?'. The thesis was divided into two parts to address the preliminary objectives, with each part comprising of three chapters. Chapter 1 provides the background context and thesis aims. The objective of Part A was to establish the methodology for the measurement, as well as the processing and interpretation of tibial acceleration data. It began with a narrative literature review on this topic (Chapter 2), specifically outlining the factors that can affect tibial acceleration during running and providing evidence-based guidelines for its use. Chapter 3 was an experimental study establishing the short- and long-term reliability and variability for the measurement of resultant tibial acceleration in runners. Chapter 4 included three sub-sections, the first of which outlines normative values for resultant tibial acceleration. This flows into establishing a method for estimating tibial fatigue fracture injury risk classification based on the normative ranges observed. The third sub-section assesses the effects of running velocity on resultant tibial acceleration magnitude.

The objective of Part B was to establish the effectiveness of real-time haptic feedback from resultant tibial acceleration for gait retraining in runners. Chapter 5 is a systematic review that establishes the evidence for augmented feedback gait retraining to reduce lower extremity loading. This review identifies gaps in the current research, as well as important factors that should be considered from a motor learning perspective, as well as practical recommendations for the use of augmented feedback with runners. Chapter 5 provided a retraining framework, and with the convergence of findings from the previous three chapters, the final experimental study (Chapters 6 and 7) used a custom-made haptic feedback watch, linked to a tibial accelerometer, to promote gait changes from runners who were pre-screened and identified as high tibial-loaders. The effectiveness of the haptic feedback on promoting changes in resultant tibial acceleration, both on a treadmill and outdoors, was assessed in Chapter 6. The self-selected strategies used by runners

to achieve reductions in resultant tibial acceleration were explored in Chapter 7, with further analyses of the strategies adopted by runners who differed in their step rate and foot strike patterns at baseline.

Finally, the thesis concludes with a general discussion integrating findings from the preceding chapters (Chapter 8). Contributions to the literature, practical implications, study limitations and directions for further work are discussed.



**Figure 2: Flow of thesis themes and chapters**

### 1.3 Specific aims of the Chapters 2-7

**Chapter 2** – *The measurement of tibial acceleration in runners. A review of the factors that can affect tibial acceleration during running and evidence-based guidelines for its use.*

- To update the current knowledge of the measurement of tibial acceleration in runners and to provide recommendations for those intending on using this assessment method in research or clinical practice.

**Chapter 3** – *The 1-week and 6-month reliability and variability of three-dimensional tibial acceleration in runners.*

- To determine between-session reliability and variability of resultant tibial acceleration at a range of running velocities.

**Chapter 4** – *The influence of running velocity on resultant tibial acceleration in runners.*

- To investigate the impact of increasing running velocity on resultant tibial acceleration in injury-free runners.
- To collect a representative database using tibial acceleration as a key biomechanical variable.
- To identify runners who might be classified as high-impact based on their resultant tibial acceleration magnitude.

**Chapter 5** – *The effectiveness of augmented feedback to reduce lower extremity loading in runners at risk of tibial fatigue fracture. A systematic review.*

- To evaluate the evidence for the use of augmented feedback to alter lower extremity loading in runners at risk of tibial fatigue fractures and to provide recommendations that guide the implementation of augmented feedback interventions.

**Chapter 6** – *The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners.*

- To examine the effectiveness of a real-time haptic feedback gait retraining system on reducing resultant tibial acceleration with individual runners.
- To assess the retention of changes over four weeks, and the transfer of learning to overground running.

**Chapter 7** – *The biomechanical effects of real-time haptic feedback gait retraining on runners with high tibial acceleration.*

- To determine the self-selected strategies used by runners to achieve a reduction in resultant tibial acceleration.
- To investigate whether there were differences in strategies used between runners who presented at baseline with a rearfoot strike or a non-rearfoot strike landing pattern, as well as runners who were categorised as having either a low or high cadence at baseline.

## **1.4 Limitations of the Thesis**

There are general limitations to this thesis that should be outlined. Firstly, actual TFF injury risk or injury occurrence, were not determined or measured in this thesis. A number of subject-specific factors affect TFF risk, and therefore to accurately determine risk level, additional information on bone remodelling, as well as intrinsic factors (bone density, geometry, nutrition, mineral content, etc.) may also need to be accounted for. Tibial acceleration, was used as a proxy or surrogate measure of the forces experienced at the tibia. In Chapter 4 tibial acceleration was measured at four running velocities from a wide range of runners, and these data were used to establish a probabilistic determination of risk for subsequent runners. Specifically, runners whose tibial acceleration exceeded one standard deviation (SD) above the mean at their comfortable training pace, were invited to participate in the intervention that formed Chapters 6 and 7. Secondly, while it is recognised that actual stress and strain on bone tissue is what mechanically results in accumulated tissue stress (see Figure 1), bone stress and strain were not calculated. Thirdly, a pre-post experimental study design was used for Chapters 6 and 7, therefore cause and effect cannot be established. To include a control group, many more runners would have needed to be screened and identified as high tibial-loaders, and non-feedback 'control' treadmill running sessions organised for these runners. While this may have strengthened the experimental design, it was not practical within the bounds of this Doctoral research. Fourthly, the research that makes up this thesis was predominantly conducted in the laboratory, allowing an element of experimental control, this did inhibit the ecological validity.



## **Part A:**

### **The Measurement of Tibial Acceleration in Runners**

#### **Overview**

The first aim of Part A was to gain an understanding of the measurement of tibial acceleration in runners from the existing published literature. This information will help guide the methods for measurement, as well as the processing and interpretation of data, which will be fundamental for all subsequent studies in this thesis. The second aim was to establish the short- and long-term reliability and variability for the measurement of resultant tibial acceleration from a representative sample of runners similar to those that would participate in future experimental studies.

## **Chapter 2:**

# **The Measurement of Tibial Acceleration in Runners. A Review of the Factors That Can Affect Tibial Acceleration During Running and Evidence-Based Guidelines for its Use.**

*This chapter comprises the following manuscript:*

Sheerin, K., Reid, D., Besier, T. (2019). The measurement of tibial acceleration in runners. A review of the factors that can affect tibial acceleration during running and evidence-based guidelines for its use. *Gait & Posture*. 67, 12–24. <http://doi.org/10.1016/j.gaitpost.2018.09.017> (Appendix 3.)

## **2 Preface**

Impact loading in runners, assessed by the measurement of tibial acceleration, has attracted substantial research attention. Due to potential injury links, particularly tibial fatigue fractures, tibial acceleration is also used as a clinical monitoring metric. However, there are contributing factors and potential limitations that must be considered before widespread implementation.

This Chapter takes the form of a narrative literature review to update the current knowledge of the measurement of tibial acceleration in runners, and to provide recommendations for the use of this measurement technique in research or clinical practice provided. This review serves as an appropriate lead into the thesis, establishing the methods by which tibial acceleration should be measured, as well as intrinsic and extrinsic factors that impact tibial acceleration during running.

## 2.0 Introduction

Running is a popular activity, but the high participation rate is accompanied by a high incidence of injuries [61]. The majority of running-related injuries occur in the lower limbs, are chronic in nature, and are related to cumulative loading [4]. The repetitive impacts associated with running is thought to play an important role in the pathophysiology of many common running injuries, especially bony fatigue fractures (commonly termed stress fractures) [10,24,62]. In runners, between 35% and 49% of all fatigue fractures occur in the tibia [7,8,63,64].

While many factors influence bony remodeling and ultimately the manifestation of a fatigue fracture [11], biomechanics dictate the level of mechanical loading on bone during running [29,30]. When the foot strikes the ground, its velocity decelerates to zero and large ground reaction forces (GRF) are generated [26]. This momentum change produces compressive loading of the lower limbs, and results in an impact shock transmitted through the musculoskeletal system, with local segment peak accelerations occurring at successively later times [27,28]. To minimise damage to proximal structures the shock is attenuated, which is accomplished through an interaction of passive and active mechanisms [65–68]. A failure of the lower extremity muscles to adequately absorb the energy of impact may lead to an over-reliance on passive mechanisms for attenuation [68].

Direct in-vivo measurement of bone strain would be ideal for monitoring injury risk in runners, however this is invasive and impractical [69,70]. Measuring the tibial acceleration (TA) via segment mounted accelerometers is a commonly used proxy measurement for the impact forces experienced at the tibia by virtue of Newton's second law ( $F=ma$ ) [33,34]. While the relationship between TA and bone strain is unclear, and likely to be complicated by local muscle forces, peak TA measured via devices attached directly to the tibia bone have revealed reasonable correlations with key GRF parameters (vertical impact peaks  $r=0.7-0.85$ ; loading rates  $r=0.87-0.99$ ) [71]. While the correlations are weaker when using skin-mounted accelerometers, average loading rate ( $r=0.274-0.439$ ) and instantaneous loading rate ( $r=0.469$ ) of the vertical GRF have all been significantly correlated with peak TA [72]. The moderate correlation between peak TA and GRF is not surprising as the GRF represents the summed acceleration of all body segments. These points withstanding, the axial component of TA has been shown to discriminate between runners with and without tibial fatigue fractures [31], and between runners injured and uninjured limb [73]. Additionally, the likelihood of the history of tibial fatigue fracture has been shown to increase by a factor of 1.4 for every 1 g increase in axial TA [35].

Previous literature reviews on the use of accelerometers in running have highlighted some of the key elements for consideration, such as the attachment method and placement location of the accelerometer, and the need for a low mass multi-axis device for increased measurement accuracy [33,34]. Despite this, the scope of these reviews did not address many of the issues and potential limitations that must also be considered when measuring TA from runners, including the influence of running velocity, technique, fatigue and surface characteristics. The objective of this review is to update current knowledge of the measurement of TA in runners and to provide recommendations for those intending on using this assessment method in research or clinical practice.

## **2.1 Methods**

PubMed, Web of Science, SPORTDiscus and Google Scholar were searched to Jan 2018 using the following terms linked with the Boolean operators ('AND' and 'OR'): 'run\*', 'tibia\* acceler\*', 'shock', 'inertia\*' and 'biomech\*', with no limits. Additional relevant studies were identified using article reference lists. Titles, abstracts and full-texts of retrieved documents were sequentially reviewed to determine their relevance. Only papers published in English, that specifically measured TA during steady-speed running, were included. Papers were excluded if they only assessed sprinting, or where participants used bodyweight support, or any form of implement or aid.

Findings from the literature covering the selection (Section 2.2.2.1), placement (Section 2.2.2.4) and attachment (Section 2.2.2.5) of accelerometers, as well as data analysis (Section 2.2.3) and key outcome measures (Section 2.2.4) are consolidated in the first half of the review. The second half of the review assesses the intrinsic (Section 2.2.5) and extrinsic (Section 2.2.6) factors that impact TA during running.

### **2.1.1 Definition of Terms**

A number of terms are used interchangeably to describe different aspects of TA, including peak TA, peak shank deceleration, peak positive acceleration and tibial shock. For the purpose of this review, axial (TA-A), anterior-posterior (TA-AP), and medio-lateral tibial acceleration (TA-ML) are used where time-domain peak acceleration magnitude components from a device aligned to the long axis of the tibia are reported. Resultant tibial acceleration (TA-R) is where the peak acceleration magnitude from all axes are used to calculate the resultant vector.

## **2.1.2 Tibial Acceleration Measurement**

### *2.1.2.1 Device Selection*

Devices contain one, two or three accelerometers mounted at right angles, each reacting to the orthogonal component acting along their axis [74]. They operate relative to the Earth's gravitational field, constantly registering 9.81 m/s/s (1 g) as a reaction to gravitational acceleration [23]. The maximum contribution of the acceleration due to gravity is 1 g (when the shank is vertical), and some accelerometers will register 9.81 m/s/s or 1 g in this position at rest, while others may read zero [23]. During the stance phase of running, the tibia undergoes angular and linear motions, with tibial angular motion largely confined to the sagittal plane, rotating about the ankle joint [75]. The TA measured by an accelerometer is the summation of the acceleration due to gravity, angular motion and the linear acceleration resulting from ground impact [76], but depending on the angle of the shank at impact, the measured acceleration contribution due to gravity will vary [75].

Recent improvements have enabled sensors that are small, light and transmit wirelessly, allowing for monitoring outside of the laboratory environment [77,78]. Accelerometers can differ across a range of parameters, which can impact on the quality of the signal. One of the main differences can be the range captured; if the signal range exceeds the capture range of the device, the measured signal will be clipped at the extremities. Some devices capture to on-board memory cards, which often have restrictions to the speed of their read-write capacity. Additionally, wireless transmitting devices can exhibit a variable length signal delay, or complete dropout. While on-board processing of data can in some cases alleviate these problems, this can also result in a reduction in the fidelity of the data. Careful assessment of all of these points is necessary when selecting a device. Where accelerometer specifications are not aligned to the task, subsequent data interpretation may be questionable. It should also be mentioned that researchers and clinicians may have access to accelerometers, that also measure other data such as EMG or gyroscopes, however these units are typically greater mass and therefore less accurate for measuring TA [79].

### *2.1.2.2 Uniaxial and Triaxial Accelerometry*

The acceleration of the tibia occurs in three dimensions, often referred to with respect to a local tibial coordinate frame: axial, antero-posterior and medio-lateral [80]. Lafortune and Hennig [38] measured all three TA components using a triaxial accelerometer, and at 4.7 m/s the TA-AP component exhibited the highest peak values (7.6 g) followed by the TA-A (5.0 g) and TA-ML component (4.5 g). The TA-AP and TA-A components were reduced at 3.5 m/s, while TA-ML

components remained constant. The authors concluded that in order to accurately quantify the total acceleration passing through the musculo-skeletal system, it is important to measure all three components of acceleration. The existence of high TA-AP components supports the hypothesis proposed by MacLellan [81] who, using high-speed films of the shank, identified a horizontally transmitted shock at heel-strike. Despite these recommendations many researchers have solely reported peak TA-A [82–86] (Table 1). When measuring TA using a uniaxial accelerometer, or when there is the intention to extract the components relative to anatomically defined axes, there is a need for careful alignment of the device to the long axis of the tibia [82–86]. If the correct alignment is not achieved, the acceleration will not accurately reflect the actual TA-A. Using all axes from a triaxial accelerometer to calculate the TA-R is one method to eliminate the need to specifically align the device to the tibial coordinate frame, thus improving repeatability of the measurement [87].

Only a small number of studies have accounted for additional acceleration components applied to the tibia [39–41,87,88] (Table 1), with one research group reporting that cadence influenced the acceleration components independently, where an increase in cadence resulted in lower TA-A and TA-R peaks, but greater TA-AP acceleration [41]. These data were captured from a single subject, running over a highly variable terrain. Thompson et al. [88] reported TA-R calculated from two movement planes only (TA-A and TA-AP). The lack of a third axis, and therefore a true resultant vector, means that data could still be lost through axis misalignment. The resultant TA takes into account all three axes, therefore the magnitudes will always be larger than the TA-A on its own. Some runners will have a dominance of the axial component, in which case the magnitudes of TA-A and TA-R may be similar, however this is not always the case, and these variables are not interchangeable. Following the initial recommendations of LaFortune et al. [38], Glauber et al. [40] report no differences in TA-A between rearfoot and non-rearfoot strike runners, however TA-R were reported to be greater in non-rearfoot runners. While they did not report the individual components, the additional acceleration present in the resultant signal could only have come from components other than TA-A.

#### *2.1.2.3 Sampling Frequency*

Nyquist theory dictates that the minimum sampling frequency should be twice the highest frequency present in a signal [74]. The measurement of human motion adds signal noise, therefore an even higher sampling frequency (5-10 times the highest frequency) is required to obtain an adequate reconstruction [74]. Power spectral analyses have revealed that 99% of the TA signal power captured during running was below 60 Hz [71,75,89]. Based on the conventions previously

outlined, this would dictate a capture sampling frequency between 300 to 600 Hz. While most researchers report a sampling rate of at least 1000 Hz [38,67,71,82], some have sampled as low as 100 Hz [90], calling their results into question (Table 1).

#### *2.1.2.4 Accelerometer Placement on the Tibia*

The distal tibia is a common location of fatigue fractures in runners, making it an important site for the measurement of acceleration [35,91,92], but many researchers have also measured TA from the proximal tibia [90,93,94] (Table 1). These differing placements may not give comparable results. Running at 4.5 m/s the tibia angle at impact can vary by up to 20° from vertical [75,95]. The linear acceleration of the tibia is influenced by centripetal acceleration due to the sagittal plane angular motion, which acts in the opposite direction to TA-A [95]. The angular acceleration is dependent on the tibial angular velocity and the distance of the device from the axis (i.e. the ankle) [75]. Both measured and modeled estimates have indicated that the TA recorded on a device attached closer to the knee substantially underestimates the TA-A at the distal attachment [96]. Taking into account the contributions of gravity and the angular component of TA, Lake et al. [95] reported that the measured TA (at 4.5 m/s) needed to increase by 1.5-3 g depending on the subject and shod condition. Additionally, the correction for angular motion influenced the TA power spectrum, with a gain in signal power particularly prevalent in the 8-13 Hz frequency band. Despite these findings, most researchers don't examine the frequency components, and often simultaneous kinematics are not captured to allow for a correction for gravity and angular motions of the lower extremity [75,95].

**Table 1: Summary of literature related to tibial acceleration measurement and analysis**

Reference	Participant Details Age (years)   Height (cm)   Weight (kg)	Accelerometer Placement (Sampling   Filtering Frequencies)	Steps / Time Recorded	Surface   Running Speed	Main Tibial Acceleration Results (g)	
Derrick [27]	10 RFS runners 25.3 ±6.5   NR   68.6 ±8.0	NR (3600 Hz   NR)	Steps = 10	Overground Speed NR	Normal stride length TA-A: 6.4 Preferred stride length TA-A: 6.2 Preferred stride length TA-A: 8.2	
Mizrahi [67]	14 M volunteer runners 24.2 ±3.7   175.5 ±5.9   73.2 ±8.3	Uniaxial acc. attached to proximal tibia (1667 Hz   NR)	Time = 20 s	Treadmill Varied (individual max effort)	Minute 1 TA-A: 6.9 ±2.9 Minute 15 TA-A: 10.5 ±4.7 Minute 30 TA-A: 11.1 ±4.2	
Derrick [68]	10 M uninjured university students 27.0 ±5.0   179.0 ±5.0   75.5 ±12.2	Uniaxial acc. attached to distal tibia (1000 Hz   NR)	Steps = 6	Overground 3.83 m/s ±5%	Preferred stride length TA-A: 6.1 Stride length +20% TA-A: 11.3 Stride length +10% TA-A: 7.9 Stride length -10% TA-A: 5.9 Stride length -20% TA-A: 5.7	
Hennig [71]	6 M 29   181.0   76.2	Triaxial acc. attached via bone pins to the proximal tibia (1000 Hz   60 Hz LP)	Steps: 5	Overground 4.5 m/s	TA-A: 5.32 ±1.5	
Milner [35]	TFF: 20 F RFS runners 26 ±9   NR   NR Control: 20 F RFS uninjured runners 25 ±9   NR   NR	Uniaxial acc. attached to distal tibia (960 Hz   NR)	Steps = 5	Overground 3.7 m/s ±0.2	TFF TA-A: 7.70 ±3.21 Control TA-A: 5.81 ±1.66	
Lafortune [75]	1 M recreational runner 32   179.0   76.0	Triaxial acc. attached via bone pins to the proximal tibia (1000 Hz   60 Hz LP)	Steps: 10	Overground	TA-A: 2.98 ±0.19 TA-A: 5.19 ±0.77	
DeBeliso [80]	10 M uninjured RFS runners 20-30   NR   78.9 ±11.9	Acc. attached to distal tibia (NR   NR)	Steps = 10	Treadmill 2.68 & 3.58 m/s	2.68 m/s BF TA-A: 3.9 ±1.4 BF innersole TA-A: 2.8 ±1.3 Shoe TA-A: 2.0 ±0.6 Shoe with innersole TA-A: 1.0 ±0.6	3.58 m/s 5.7 ±2.4 4.2 ±2.1 3.2 ±1.5 2.8 ±1.1
Lafortune [38]	6 M 29   181.0   76.2	Triaxial acc. attached via bone pins to the proximal tibia (1000 Hz   NR)	Steps: 5	Overground 4.5 m/s	TA-A - measured: 5.32 ±1.6 TA-A - estimated: 9.39 ±1.8	
Crowell [82]	4 M / 6 F RFS recreational runners with TA-A >8 g 26 ±2.0   172.0 ±0.07   81.5 ±21.0	Uniaxial acc. attached to distal tibia (1080 Hz   100 Hz LP)	NR	Overground 3.7 m/s	Baseline TA-A: 8.2 ±2.5 Post FB TA-A: 4.3 ±1.5 1 month post FB TA-A: 4.6 ±1.5	
Crowell [83]	5 F recreational runners 26.0 ±2.0   164.0 ±0.06   59.3 ±5.4	Uni-axial acc. attached to distal tibia (1080 Hz   100 Hz LP)	Steps = 20	Treadmill Self-selected (2.4-2.6 m/s)	Baseline TA-A: 9.0 ±1.6 Post FB TA-A: 7.2 ±4.9 10 min post FB TA-A: 6.3 ±3.5	



**Table 1: Continued**

Reference	Participant Details Age (years)   Height (cm)   Weight (kg)	Accelerometer Placement (Sampling   Filtering Frequencies)	Steps / Time Recorded	Surface   Running Speed	Main Tibial Acceleration Results (g)	
Dufek [84]	11 F pre-menarche 9.2 ±1.9   139.9 ±12.5   32.9 ±7.7 11 F normally menstruating 25.2 ±3.9   1.64.3 ±3.2   63.6 ±9.2 12 F post-menopausal 53.2 ±4.6   163.0 ±8.2   67.2 ±13.0	Uniaxial acc. attached to distal tibia (1000 Hz   NR)	Time = 45 s	Treadmill Varied speeds	Preferred velocity TA-A: 4.87 ±1.88 Preferred velocity +10% TA-A: 6.07 ±2.41 Preferred Velocity TA-A: 4.36 ±1.32 Preferred velocity +10% TA-A: 4.77 ±1.50 Preferred Velocity TA-A: 3.56 ±1.74 Preferred velocity +10% TA-A: 4.05 ±2.39	
Sheerin [87]	14 M recreational runners 33.6 ±11.6   177.2 ±6.6   75.6 ±9.5	Triaxial acc. attached to the distal tibia (1000 Hz   60 Hz LP)	Steps = 61 ±1.5	Treadmill 2.7 m/s 3.0 m/s 3.3 m/s 3.7 m/s	TA-R: 7.8 ±2.9 – 8.6 ±3.4 TA-R: 9.1 ±2.7 – 9.7 ±3.4 TA-R: 10.4 ±3.4 – 11.7 ±3.8 TA-R: 12.9 ±4.3 – 11.9 ±3.6	
Glauberman [40]	20 F uninjured distance runners 27.8 ±3.7   168.1 ±6.2   59.2 ±7.3	Triaxial acc. attached to distal tibia (NR   NR)	Time = 60 s	Treadmill 3.13 m/s	Natural RFS (control shoes) TA-R: 11.5 ±2.8 Natural non-RFS (control shoes) TA-R: 8.7 ±2.8 TA-A: 6.3 ±1.1 Natural RFS (baseline) TA-R: TA-A: 7.4 ±0.8 9.3 ±2.6 NR Natural RFS (altered strike) TA-R: 11.2 ±1.6 NR Non-natural RFS (baseline) TA-R: 13.1 ±1.2 NR Non-natural RFS (altered strike) TA-R: 9.5 ±1.6	
Thompson [88]	5 M   5 F uninjured RFS runners 26 ± 7.3   174.0 ±9.0   65.6 ±10.2	Bi-axial acc. attached to the distal tibia (1000 Hz   NR)	Steps: 10	Overground 2.25 ±0.19 m/s	TA-R shod: 11.27 ±1.73 TA-R BF: 11.32 ±1.48 TA-R BF RFS: 13.55 ±1.51	
Giandolini [41]	1 M elite trail runner 26   171   56.5	Triaxial acc. attached to the proximal tibia (1300 Hz   50 Hz LP)	Steps: 5530	Overground trail race Variable speed	TA-A: 8.41 ±3.37 TA-R: 10.4 ±3.42	
Wood [39]	3 M   6 F uninjured recreational runners 20 ±1.5   170.2 ±8.7   59.1 ±8.2	Triaxial acc. attached to the distal tibia (612 Hz   NR)	Steps: 20	Treadmill 3.13 ±2.5 m/s	Baseline PTA(R): 5.9 ±0.7 FB1 PTA(R): 5.30 ±0.80 No FB1 PTA(R): 5.60 ±1.10 FB2 TA-R: 5.20 ±0.60 No FB2 TA-R: 5.4 ±0.70 <i>Proximal</i> 4.34 ±1.29 g 5.03 ±1.45 g 5.71 ±1.64 g	

**Table 1: Continued**

Reference	Participant Details Age (years)   Height (cm)   Weight (kg)	Accelerometer Placement (Sampling   Filtering Frequencies)	Steps / Time Recorded	Surface   Running Speed	Main Tibial Acceleration Results (g)	
Garcia-Perez [90]	11 M   9 F uninjured recreational runners 34 ±8   172 ±8   63.6 ±8.0	Uniaxial acc. attached to proximal tibia (100 Hz   NR)	Time = 10 s	Treadmill & overground 3.81 ±0.40 m/s	<i>Overground</i> Pre-fatigue TA-A: 24.6 ±10.8 Post-fatigue TA-A: 22.2 ±10.3 <i>Treadmill</i> 15.3 ±6.8 17.2 ±9.5	
Gruber [91]	12 M   7 F habitual RFS runners 26.7 ±6.1   175.0 ±9.0   70.1 ±10.0 14 M   5 F habitual FFS runners 25.4 ±6.2   176.0 ±10.0   68.8 ±9.5	Uniaxial acc. attached to distal tibia (1200 Hz   60 Hz LP)	Steps = 11000	Treadmill 3.47 ±0.90 m/s 3.73 ±0.24 m/s	RFS TA-A: 5.07 ±1.49 FFS TA-A: 3.87 ±1.36	
O'Leary [92]	7 M & 9 F uninjured runners 20-30   1.73 ±0.09   68.4 ±12.0	Uniaxial acc. attached to distal tibia (2000 Hz   100 Hz LP)	Steps = 5	Overground 3.2 ±0.3 m/s	Without innersoles TA-A: 4.81 ±1.45 With innersoles TA-A: 4.05 ±1.69	
Verbitsky [93]	22 M uninjured runners 30.8 ±5.1   173.9 ±7.3   70.4 ±9.2	Uniaxial acc. attached to proximal tibia (1667 Hz   NR)	Time = 30 s every 5 min.	Treadmill Varied speeds	<i>Fatigue group</i> Baseline TA-A: 6.0 ±0.5 5 min TA-A: 7.5 ±0.8 10 min TA-A: 9.0 ±1.7 15 min TA-A: 9.5 ±0.9 20 min TA-A: 8.7 ±1.2 25 min TA-A: 8.8 ±1.2 30 min TA-A: 9.5 ±1.0	<i>Non-fatigue group</i> Baseline TA-A: 7.0 ±1.7 5 min TA-A: 6.9 ±0.8 10 min TA-A: 6.9 ±1.1 15 min TA-A: 6.3 ±0.7 20 min TA-A: 7.0 ±1.2 25 min TA-A: 6.3 ±1.1 30 min TA-A: 7.0 ±1.3
Lake [95]	2 M recreational runners NR   NR   NR	Uniaxial acc. attached to distal tibia (2000 Hz   60 Hz LP)	Steps = 5-10	Overground Self-selected speed (~4.5 m/s)	BF TA-A: 16.34 ±1.60 - 17.76 ±1.74 BF corrected TA-A: 17.87 ±2.07 – 20.73 ±2.56 Shod TA-A: 8.51 ±1.43 – 8.67 ±0.97 Shod corrected TA-A: 10.07 ±1.29 – 10.52 ±1.27	

**Table 1: Continued**

Reference	Participant Details Age (years)   Height (cm)   Weight (kg)	Accelerometer Placement (Sampling   Filtering Frequencies)	Steps / Time Recorded	Surface   Running Speed	Main Tibial Acceleration Results (g)	
Creaby [97]	11 M uninjured runners - Clinician guided FB 28.1 ±7.8   178.0 ±0.05   76.5 ±7.7 11 M uninjured runners - visual FB 22.7 ±4.6   179.0 ±0.05   78.8 ±9.2	Triaxial acc. attached to distal tibia (1500 Hz   100 Hz LP)	Time = 10 s	Treadmill 3.0 m/s	<i>Clinical FB</i> Baseline TA-A: 5.74 ±2.25 During FB TA-A: 4.37 ±1.86 Post FB TA-A: 4.13 ±1.82 7-days post FB TA-A: 4.48 ±1.53	<i>Visual acc. FB</i> 5.34 ±1.93 3.81 ±1.36 4.32 ±2.20 4.20 ±1.54
Sinclair [98]	12 M uninjured runners 23.7 ±2.3   176.5 ±5.8   75.6 ±7.6	Triaxial acc. attached to the distal tibia (1000 Hz   60 Hz LP)	NR	Treadmill 4.0 m/s	Conventional shoes TA-A: 4.28 ±2.28 Light shoes with added mass TA-A: 5.47 ±1.83	
Sinclair [99]	12 M experienced runners 24.3 ±1.1   178.1 ±5.2   76.8 ±9.0	Triaxial acc. attached to the distal tibia (1000 Hz   60 Hz LP)	Steps: 6	Overground 4.0 m/s	Conventional shoes TA-A: 6.60 ±3.65 Barefoot TA-A: 9.17 ±2.96 Barefoot inspired TA-A: 10.2 ±3.48	
Sinclair [100]	12 M uninjured runners 23.5 ±2.0   177.1 ±4.6   77.5 ±5.5	Triaxial acc. attached to the distal tibia (1000 Hz   60 Hz LP)	Steps: 5	Overground 4.0 m/s	Conventional shoes TA-A: 6.60 ±2.47 Energy return shoes TA-A: 7.03 ±2.79	
Sinclair [101]	12 M uninjured runners 23.1 ±5.0   178.0 ±0.1   77.1 ±7.9	Triaxial acc. attached to the distal tibia (1000 Hz   60 Hz LP)	Steps: 5	Overground 4.0 m/s	Conventional shoes TA-A: 6.73 ±1.79 Maximalist shoes TA-A: 7.99 ±2.32 Minimalist shoes TA-A: 9.54 ±4.29	
Sinclair [102]	12 F uninjured recreational runners 21.45 ±2.98   166.0 ±6.0   60.87 ±4.37	Triaxial acc. attached to distal tibia (1000 Hz   60 Hz LP)	Steps = 5	Overground 4.00 m/s ±5%	Normal shoes TA-A: 10.70 ±2.31 Cooled shoes TA-A: 12.75 ±4.62	
Sinclair [100]	15 M uninjured runners 21.02 ±2.02   176.6 ±5.3   76.82 ±6.27	Uniaxial acc. attached to distal tibia (1000 Hz   60 Hz LP)	Steps = 5	Overground 4.00 m/s ±5%	Conventional shoes TA-A: 5.25 ±1.43 Energy return shoes TA-A: 5.90 ±1.58	
Sinclair [103]	13 M uninjured runners 27.81 ±7.02   177 ±11   76.22 ±7.04	Triaxial acc. attached to distal tibia (1000 Hz   60 Hz LP)	Steps = 5	Overground 4.0 m/s ± 5%	BF TA-A: 5.72 ±1.34 Cross-fit TA-A: 5.17 ±1.82 Conventional TA-A: 4.55 ±1.29 Minimalist TA-A: 5.31 ±1.55	
Laughton [104]	15 RFS runners 22.46 ±4.0   169.75 ±6.07   66.41 ±8.58	Uniaxial acc. attached to distal tibia (960 Hz   100 Hz LP)	Steps = 5	Overground 3.7 m/s ±5%	No Orthotics TA-A: 7.18 ±2.98 Orthotics TA-A: 6.78 ±3.14 RFS TA-A: 7.82 ±3.16 FFS TA-A: 6.15 ±2.96	

*M – male; F – female; acc – accelerometer; TA-A - peak axial tibial acceleration; TA-R – peak resultant tibial acceleration; RFS - rearfoot strike; FFS - forefoot strike; BF – barefoot; FB – feedback; NR - not reported; LP - low pass.*

#### *2.1.2.5 Accelerometer Attachment*

To determine the best estimate of the acceleration of a segment of interest, an accelerometer attached directly to the bone is most accurate, however this is impractical for regular use [38,71,75,105] (Table 1). LaFortune et al. [105] compared the TA-A measured from bone and skin mounted accelerometers while runners ran overground. For some subjects the skin-mounted accelerometer overestimated TA-A by as much as twice the bone-mounted devices. While the dominant component of these peaks represented the impact, the signal also included acceleration components due to muscular action, and noise due to resonance in the compliant attachment of the accelerometer [89,105]. The absolute differences between the signals was large, but with a low-pass filter, signals from a skin-mounted device adequately represented bone accelerations [105]. There will always be oscillation of skin-mounted accelerometers, therefore it is important to know the characteristics of this oscillation. If the resonance frequency of the accelerometer and mounting system occurs at the same frequencies of those from ground impact (10-20 Hz) the measured acceleration will be elevated [89]. Ziegert and Lewis [106] studied the effect of soft tissue, by comparing the output of a surface-mounted accelerometer with that of a device connected to the tibia bone via a needle. When the leg was impacted with a device, a 1.5-gram surface-mounted accelerometer showed almost identical outputs to the bone, but a 34-gram accelerometer gave outputs with little resemblance to the bone acceleration, appearing to oscillate at its resonant frequency on the soft tissue. Three studies have reported the natural resonant frequency of the accelerometer as 250 Hz [71], 400 Hz [38] and 1000 Hz [105]. Henning, et al. [71] reported mathematically and experimentally deriving this frequency, however no methods outline or appropriate reference was provided.

Accelerometer oscillation can be minimised by tensioning the device attachments [107], with Clarke et al. [108] reporting that a preload force 'as tight as tolerable' improved reliability, both within and between sessions. Forner-Cordero et al. [79] conducted a series of experiments to determine the frequency characteristics of skin-mounted devices under varied attachment conditions, including using elastic bands, a method commonly used in recent research [84,97,109,110] (Table 1). They also outlined a test to validate the attachment integrity before recording clinical measurements, which involved subjects standing on their tiptoes, and falling freely onto their heels. While this test is unlikely to produce TA magnitudes representative of running, it did show low variability, and could discriminate between different attachment conditions [79]. Once again, without adequate preload force, the frequency of the accelerometer-mounting system was too low, close to the frequency range of the data, increasing measurement error. While there is still no clarity on what constitutes tensioning 'as much as tolerable', and

acknowledgement that this will differ for individuals, a simple test, such as the 'heel drop', could be an effective method to compute the frequency of the accelerometer mounting to allow confirmation of the integrity of attachment before testing begins.

### **2.1.3 Tibial Acceleration Analysis**

#### *2.1.3.1 Normalisation*

To account for variability in absolute magnitudes between sessions, normalisation of TA data has been proposed [108]. Expressing TA-A relative to the mean observed at the slowest running velocity, provided a 'shock ratio', which can be useful considering the absolute values of the peak accelerations are susceptible to noise and vibration. Focusing on the relative magnitudes of acceleration measures can be informative for many applications (e.g. cushioning properties of running shoes), however to be of use in the comparison of datasets, multiple, and consistent running velocities would be required.

#### *2.1.3.2 Frequency Content of Acceleration*

While time domain TA components are most commonly reported, the signal is formed by acceleration components of various frequencies, which are superimposed in the time domain signal [89]. The low frequency component (4–8 Hz) is the acceleration associated with voluntary leg motion, while the high frequency component (10–20 Hz) represents the rapid deceleration of the lower extremity at contact [89,91]. These low and high frequency ranges are also representative of the active and impact peaks of the vertical GRF, respectively [89,111]. The resonant frequency of the mounting system also contributes to the time domain signal.

It is possible to separate the frequency components using a frequency analysis [89,112]. A fast fourier transform provides the median power frequency of the acceleration signal, or alternatively a joint time-frequency distribution analysis can provide the instantaneous power spectrum [112]. Variations or changes in peak TA observed in the time domain may be a result of changes in low or high frequency bands, or changes in the resonant frequency of the mounting system [112]. These additional signal analysis approaches have been used to provide a more thorough characterisation of the signal components in a range of running studies [89,98,99,112].

#### *2.1.3.3 Signal Filtering*

All kinematic data contains a true signal representing human movement, as well as noise, therefore some pre-analysis filtering is required [74]. While both the true signal and noise occupy a wide band-width, noise is usually at the higher end of the frequency spectrum. If the cut-off is

set too low, the resulting signal will be incorrect, whereas if the cut-offs are too high, too much noise will remain in the signal [74]. Most studies measuring TA magnitude in the time domain during running report using low-pass filters with cut-offs between 40 Hz and 100 Hz [71,75,82,90,97,100,105,110], which were in some cases determined via power spectral analyses of the signal [75,105]. Selecting the appropriate filter cut-off frequencies is essential, as over or under filtering data can lead to inaccurate interpretations. A TA signal also contains low frequency components (4–8 Hz) associated with voluntary leg motion, and the acceleration of the body COM, therefore it is possible to supplement the low-pass filtering with a high-pass (e.g. 10 Hz), or use band-pass filter (e.g. 10-60 Hz) to exclusively reveal the frequency component related to the passive impact of running gait. These filtering methods do not appear to be widely used [89,91] (Table 1).

#### **2.1.4 Outcome Measures**

Where triaxial devices are used, TA signals can be resolved into three acceleration components. The coordinate system axes can be defined differently, but commonly the orthogonal axes are defined with respect to the tibia: TA-A, TA-AP and TA-ML. The TA-A corresponds to a line bisecting the proximal and distal ends of the tibia in both the frontal and sagittal planes. The medio-lateral axis runs perpendicular to the axial axis and parallel to a line joining the two malleoli, and the antero-posterior axis is mutually orthogonal to both the longitudinal and medio-lateral axes [38]. A number of additional variables can be calculated from the measured signals. The most commonly reported are peak TA-A magnitude, followed by peak TA-R [38–41,87,88]. A smaller number of studies have also reported peak positive TA-ML [38] and TA-AP [38,41], as well as time to peak positive [38,71,75,100–102,105], TA-A slope [100,102,103,113], TA-A loading rate [101], duration of peak positive [38], peak negative [38,40], duration of negative acceleration [38], and peak positive to peak negative acceleration [104,114] (Table 1). It should be noted, TA-A magnitude is currently the only parameter linked to running injury [35].

Despite the widespread use, publications describing the acceptable reliability of accelerometers attached to the tibia of runners is limited [43,115]. Clarke et al. [43] collected TA-A data from three subjects running on a treadmill at 3.8 m/s during five separate sessions. The mean within-session step-to-step variability was 6.8%, and the between-session variability was 5.6%. With the between-session variability falling inside the step-to-step variability, it was deemed that accurate comparisons could be made between sessions. Sheerin et al. [115] report the one-week reliability from 20 runners at a range of velocities (2.7-3.7 m/s) on a treadmill. While the TA-A results were acceptable at all velocities, they were generally larger for TA-A compared to TA-R for both the

percentage difference in the means (TA-A 0%-5.7%; TA-R 0.9%-5.1%) and the effect sizes (TA-A 0.01-0.17; TA-R 0.01-0.12), indicating slightly better session-to-session TA-R reliability.

## **2.1.5 Intrinsic Factors That Can Modify Tibial Accelerations**

### *2.1.5.1 Running Velocity*

The seminal work analysing the effect of running velocity report consistently increased peak TA magnitude with faster running velocities (3.5 and 4.7 m/s) across all components of TA (TA-A, TA-AP and TA-ML) from a single recreational runner, using a bone-mounted accelerometer [105]. This increase in TA-A was also reported at a series of faster running velocities (spanning 3.4 to 5.4 m/s) from 10 well-trained runners [108]. Further to this, linear regression analysis revealed that average TA-A increased by 34% for each 1.0 m/s increase in running velocity. Individual linear relationships varied between 0.15 and 0.68, and while the best-fit linear relationship was described as in 'good agreement' with the experimental data, no supporting statistics were provided [108].

All subsequent studies reporting TA-A while running velocity was manipulated as an independent variable confirm that running at faster velocities was associated with increased TA-A, irrespective of running surface, footwear, running experience, or whether the velocity was fixed, or self-selected [84,116–120] (Table 1 & Table 2). While the focus of two of these studies were on shock attenuation between the tibia and the head, the results provided insight into the characteristics of TA-A change with increasing velocity [84,120]. Investigating the characteristics of shock attenuation across a range of running velocities up to a runners' maximum, Mercer et al. [120] report that the average TA-A remained constant for both 50% and 60% of maximal velocity, but increased at faster velocities. The TA-A variability (SD) remained relatively constant for the first four velocities, before increasing at 90% and 100% of maximal velocity. In a subsequent study, a mixed model design was used to examine the impact of attenuation characteristics of different groups female runners (pre-pubescent girls, normally menstruating women and postmenopausal women) [84]. Participants ran on a treadmill at their preferred velocity (1.9 to 2.6 m/s) and at a velocity 10% faster, while TA-A values ranging from 3.6 to 6.1 g were recorded. The authors claimed that the results demonstrated the anticipated response for velocity, with all groups exhibiting greater peak TA-A during faster running. However, with deeper analysis, it is evident that the TA-A measured from the prepubescent girls were larger than those measured from the normally menstruating women, despite running slower. While speculative, this could be as a result of the younger girls having a reduced tibia mass, and therefore reduced effective mass [76]. These studies were limited by small samples sizes, and the fact that comparisons were made against

the percentage of their individual maximum [120], or comfortable running velocity [84], rather than an absolute velocity.

There is still an absence of normative TA values for runners at a range of running velocities. Sheerin et al. [121] measured TA-R for 82 runners running at four different treadmill velocities (2.7-3.7 m/s) and report mean values ranging from 9.8 +2.7 g at the slowest velocity to 12.1 +3.1 at the faster velocity. Values from individual runners were spread with 4.5 g the lowest recorded at 2.7 m/s, and 20.6 g the highest recorded at 3.7 m/s. A moderate positive correlation ( $r=0.42$ ) was reported between velocity and TA-R, and a regression analysis that revealed that for every 1 m/s increase in velocity TA-R would increase by 3.7 g.

#### *2.1.5.2 Stride Rate and Stride Length*

In the first three of six studies to assess the influence of stride rate and stride length on TA magnitude, stride rate was manipulated to 5% and 10% slower, and 5% and 10% faster, than subjects' preferred, while controlling velocity at 3.8 m/s [43]. Runners adapted to a stride rate 10% and 20% slower, and a 10% and 20% faster than their preferred, while running at their preferred velocity [122], and finally under the same stride rate conditions at 3.8 m/s [68]. Peak TA-A showed a positive linear trend with increased stride length across all three studies [68]. This increase is likely due to a simultaneous decrease in effective mass, which has been closely linked to knee angle (and therefore stride length) at impact [76,123].

Independently manipulating stride length and rate at different velocities has further expanded the understanding of the relationship of TA-A with these fundamental variables. Running with a longer than preferred stride length, leads to increased TA power spectral density [120,124], which were four times greater when stride length, as opposed to stride rate, was varied [124]. When TA-A was compared between preferred stride length and a stride length constrained to 2.5 m at various velocities [125], magnitudes increased by approximately 24% per 1 m/s increase in running velocity. This is lower than the 42% [124] and 34% [108] increases previously reported, however when stride length was constrained, there was no clear relationship between TA-A and running velocity [125]. These results support the notion that kinematic factors, such as the particular orientation of the hip, knee and ankle joints for a given stride length, might be critical in determining TA magnitude.



**Table 2 Summary of literature related to intrinsic and extrinsic factors that can modify tibial accelerations**

Reference	Participant Details Age (years)   Height (cm)   Weight (kg)	Accelerometer Placement (Sampling   Filtering Frequencies)	Steps / Time Recorded	Surface   Running Speed	Main Tibial Acceleration Results (g)		
Greenhalgh [117]	9 M hockey players 21.0 ±1.69   175.75 ±6.56   78.13 ±12.11	Triaxial acc. attached to distal tibia (1000 Hz   60 Hz LP)	Steps = 6	Varied surfaces 3.3 m/s ±5% 5.0 m/s ±5%	3.3 m/s Synthetic surface TA-A: 4.2 ±1.2 – 5.0 ±1.2 3.3 m/s Concrete TA-A: 4.8 ±1.8 – 5.5 ±1.8 5.0 m/s Synthetic surface TA-A: 7.4 ±2.6 – 9.1 ±2.7 5.0 m/s Concrete TA-A: 10.5 ±2.4 – 8.4 ±2.7		
Boey [118]	18 M & 20F uninjured runners Untrained: 22.3 ±1.8   173.2 ±8.9   65.0 ±9.1 Rec.: 22.3 ±2.5   176.3 ±9.2   65.8 ±8.1 Trained: 25.4 ±5.0   178.0 ±7.9   63.3 ±5.1	Triaxial acc. attached to distal tibia (1024 Hz   60 Hz LP)	Steps = 15	Concrete - 3.1 m/s Concrete - SS Track - 3.1 m/s Track - SS Trail - 3.1 m/s Trail – SS	<i>Untrained</i> TA-A: 10.55 ±2.20 TA-A: 10.99 ±2.98 TA-A: 10.38 ±1.83 TA-A: 10.88 ±2.79 TA-A: 9.55 ±1.53 TA-A: 9.84 ±2.21	<i>Recreational</i> 9.39 ±2.98 9.83 ±2.67 9.31 ±2.96 9.91 ±2.75 8.93 ±2.62 9.83 ±2.42	<i>Well-trained</i> 9.11 ±2.33 10.62 ±3.21 8.99 ±2.22 10.27 ±2.95 8.38 ±1.98 10.17 ±2.77
Montgomery [119]	15 recreational runners NR   NR   NR	Triaxial acc. attached to mid-tibia (1500 Hz   60 Hz LP)	Steps = 8	Varied surfaces 2.88 ±0.35 m/s 4.25 ±0.37 m/s	2.88 m/s Overground: 5.1 Motorised treadmill: 5.4 Non-motorised treadmill: 3.7		4.25 m/s 8.0 8.1 6.2
Sinclair [126]	10 M RFS recreational runners 20.42 ±3.55   178.75 ±5.81   76.58 ±6.52	Triaxial acc. attached to distal tibia (1024 Hz   60 Hz LP)	Steps = 10	Overground 3.5 m/s ±5% 5.0 m/s ±5%	3.5 m/s BF TA-A: 6.85 ±3.51 BF inspired shoes TA-A: 5.54 ±1.31 Conventional shoes TA-A: 2.28 ±0.64		5.0 m/s 12.81 ±5.74 7.92 ±4.30 4.54 ±1.14
Giandolini [127]	12 M & 8 F uninjured RFS recreational runners 19.7 ±1.3   177 ± 79   70.7 ±9.0	Uniaxial acc. attached to distal tibia (1000 Hz   50 Hz LP)	Time = 10 s	Treadmill Self-selected	<i>MFS training</i> Baseline TA-A: 6.80 ±1.55 1-month TA-A: 6.57 ±2.12 2-month TA-A: 7.47 ±1.71 3-month TA-A: 6.70 ±1.46		<i>Shoe training</i> 5.60 ±1.04 5.73 ±1.53 6.18 ±1.90 6.67 ±1.48
Fu [128]	13 M uninjured recreational RFS runners 23.7 ± 1.2   173.7 ± 5.7   65.7 ± 5.2	Biaxial acc. attached to the proximal tibia (1000 Hz   100 Hz LP)	Steps = 10	Varied surfaces 3.33 ±0.17 m/s	Concrete TA-A: 2.4 ±3.1 Synthetic track TA-A: 10.9 ±3.5 Grass TA-A: 11.1 ±3.4 Treadmill TA-A: 11.6 ±3.0 Treadmill EVA TA-A: 10.3 ±3.1		
McNair [129]	10 M RFS runners 75 ±6   NR   NR	Uniaxial acc. attached to distal tibia (1000 Hz   NR)	Steps = 8	Treadmill 3.5 m/s	<i>Shoe conditions</i> Double density EVA with a cantilever outsole: 10.0 Double density EVA: 10.3 Air filed chambers within double density EVA: 10.0 Encapsulated double density EVA: 9.8 BF: 14.0		

Table 2: Continued

Reference	Participant Details Age (years)   Height (cm)   Weight (kg)	Accelerometer Placement (Sampling   Filtering Frequencies)	Steps / Time Recorded	Surface   Running Speed	Main Tibial Acceleration Results (g)	
Chambon [130]	15 M uninjured recreational runners 23.9 ±3.2   177.0 ±3.0   73.0 ±8	Triaxial acc. attached to the middle medial tibia (2000 Hz   50 Hz LP)	Steps = 5	Overground (flat and 10° incline) 3.3 m/s ±5%	<i>Flat: New</i> Viscous TA-A: 5.37 ±1.86 Intermediate TA-A: 4.22 ±1.17 Elastic TA-A: 4.09 ±0.87 <i>Flat: Fatigued</i> Viscous TA-A: 5.56 ±1.76 Intermediate TA-A: 4.68 ±1.19 Elastic TA-A: 5.03 ±1.37	<i>Inclined: New</i> 5.22 ±1.55 4.39 ±1.83 3.73 ±1.17 <i>Inclined: Fatigued</i> 6.11 ±1.77 4.29 ±1.36 5.35 ±1.77
Clark [131]	36 F injury free runners (>30min per run; >3x weekly)	Triaxial acc. attached to proximal tibia (2000 Hz   60 Hz LP)	Steps = 16	Treadmill 2.8 m/s	<i>Day 1</i> No pill TA-A: 4.17 ±1.96 No pill TA-AP: 1.92 ±0.37 Contraceptive pill: TA-A: 4.99 ±2.02 Contraceptive pill: TA-AP: 1.79 ±0.35	<i>Day 14</i> 4.24 ±2.02 1.89 ±0.40 4.67 ±2.46 1.78 ±0.37
Clansey [110]	Uninjured recreational runners (RFS) with elevated TA-A (>9 g) 12 M (intervention) 33.3 ±9.0   180.0 ±0.1   77.2 ±11 12 M (control) 33.9 ±11.3   180.0 ±0.1   75.1 ±6.9	Triaxial acc. attached to distal tibia (1500 Hz   60 Hz LP)	NR	Treadmill 3.7 m/s (1% gradient)	<i>Intervention</i> Pre-intervention TA-A: 10.67 ±1.85 Post-intervention TA-A: 7.39 ±1.48 1 month post-intervention TA-A: 8.30 ±1.82	<i>Control</i> 9.78 ±1.68 9.99 ±1.97 9.68 ±1.87
Butler [132]	12 uninjured high arch runners 20.9 ±3.0   170.0 ±0.07   68.36 ±5.75 12 uninjured low arch runners 21.8 ±3.2   173.0 ±0.11   70.04 ±7.35	Uniaxial acc. attached to distal tibia (1080 Hz   NR)	NR	Treadmill   Self-selected training pace	<i>Beginning of run</i> High arch cushion shoes TA-A: 5.5 ±0.7 High arch motion control shoes TA-A: 4.5 ±0.4 Low arch cushion shoes TA-A: 4.6 ±1.4 Low arch motion control shoes TA-A: 5.7 ±1.7	<i>End of run</i> 5.9 ±0.9 4.6 ±0.6 NR NR

*M – male; F – female; acc – accelerometer; TA-A – peak axial tibial acceleration; TA-R – peak resultant tibial acceleration; RFS – rearfoot strike; FFS – forefoot strike; BF – barefoot; FB – feedback; NR – not reported; LP – low pass.*

### *2.1.5.3 Fatigue*

While a complex phenomenon, exercise induced fatigue is an important factor in the development of fatigue fractures [133]. Increases in TA-A towards the end of high intensity treadmill running bouts designed to induce central fatigue (related to a failure in neural drive), have been reported, in some cases by as much as 100% [67,134–137]. Derrick et al. [134] suggested that increases in knee flexion angle and foot inversion at contact may be responsible for the increased TA-A, and that these adaptations decrease the effective mass of the system, therefore increasing TA-A. Citing a spring-damper model simulating human running vertical GRFs [138], it is reasoned that increased TA-A should not necessarily be linked to an increased injury potential, suggesting that decreasing the effective mass will increase the TA-A, while at the same time decreasing the impact forces [134]. These conclusions are contradictory to the evidence linking increased TA-A magnitude with tibial fatigue fracture development in runners [31,35]. These views do highlight that the evidence is not clear and that researchers disagree on this topic. Contrasting the evidence that TA increases with central fatigue [67,134–137], Abt et al. [109] report no changes in any kinematic or acceleration variables after the exhaustive treadmill run. Unclear findings were also reported in a subsequent study where fatigue effects on TA-A were compared when runners ran both overground and on a treadmill [90]. On average, TA-A increased during the treadmill run, but this was not replicated with overground running. Additional kinematic variables were not captured, and therefore the characteristics of the adaptations could not be analysed further.

To which extent local muscle fatigue effects TA in running has not been demonstrated [139]. Several studies have used a human pendulum approach to control kinematic variables such as joint position and impact velocity, while reproducing impact parameters which closely resemble those of normal running [94,139,140]. In contrast to experiments on central fatigue, across a range of different protocols, localised muscle fatigue was found to cause a decrease in TA-A magnitude and slope at impact [94,139,140]. It is thought that these changes are a result of the reduction in the force generating capacity of the muscle due to fatigue. The implications of these findings are not likely to be fully appreciated until more extensive evaluations of the roles of individual muscles on segment and joint stiffness, and how this translates to the actual running environment [94]. Overall, there have been inconsistencies in the fatigue protocols, the varying levels of runners used, and a lack of understanding of the implications of effective mass during ground impact in running. These factors have meant that the effect of both central and localised muscle fatigue on TA is inconclusive.

#### *2.1.5.4 Joint Kinematics*

Lower extremity joint positions at contact are closely connected to stiffness and effective mass, and therefore their position or alignment at initial contact may effect TA magnitude. Denoth [141] and McMahon et al. [142] demonstrated that greater knee flexion angle resulted in smaller effective mass and a reduction in stiffness, leading to greater shock absorption. This concept has been supported with the ‘two-mass’ running model, and its association with vertical GRF–time waveform patterns [143].

With the association between a foot strike pattern and the absence or reduction in GRF vertical impact peak, it was hypothesised that landing with a strike pattern further forward on the foot (e.g. forefoot or midfoot) would reduce peak TA [40,104,114,127]. However, when viewed in relation to foot strike mechanics, the findings are conflicting. Where runners transitioned from a rearfoot strike to either a midfoot or forefoot strike pattern, increases in TA-R [40], and in signal power in the 9-20 Hz frequency range [91], were reported. However, either no change [127], an increase [91,104], or a decrease [114] in TA-A variables were also found (Table 1 & Table 2). Additionally, when non-rearfoot strike runners transitioned to a rearfoot strike pattern they demonstrated a decrease in TA-R [40]. A number of factors could contribute to these conflicting findings, specifically the varied definitions of running kinematics (e.g. forefoot [91] versus non-rearfoot strike [40]), different baseline characteristics [40,91], and the differing intervention durations for retraining habitual patterns [104,127].

To enhance the understanding of the effects of running kinematics on TA, it makes sense to also consider a greater number of segments. While they can’t be determined clinically, lower extremity stiffness and effective mass can also have a meaningful impact on TA. Analysing the discrete kinematic parameters associated with the passive attenuation of both time and frequency domain characteristics, knee flexion velocity at foot-strike was found to be the single regulator of time domain peak TA [144]. While a large proportion of variance and associated mechanisms remain unexplained, this provides some evidence that kinematic parameters can influence TA magnitude during running. When kinematics and stiffness parameters were monitored alongside alterations in decline surface gradient, runners could be classified by their shock attenuation [145]. While all runners demonstrated increased accelerations at the tibia and head with increased decline gradients, Runners with reduced shock attenuation (i.e. relatively higher head accelerations) also demonstrated differences in lower extremity and trunk kinematics at both heel-strike and mid-stance. Specifically, these runners exhibited higher COM displacement, heel-strike velocity, and reduced COM stiffness and damping.

Further evidence supporting an influential relationship between running technique and TA was seen where runners were able to actively modify their kinematics to reduce TA-A, by as much as 50%, after a single session of real-time visual feedback [82,97], or 10% reductions in TA-R in response to real-time audio feedback [39]. Similar changes were noted four weeks post intervention, when runners were screened for high pre-intervention TA-A values and exposed to a more extensive feedback schedule [82,110]. Reductions in TA-A were accompanied by lower instantaneous vertical force loading, as well as increased ankle plantar flexion and decreased heel vertical velocity at initial contact, and changes from a rearfoot strike to a midfoot strike pattern [82,110].

## **2.1.6 Extrinsic factors that can modify tibial accelerations**

### *2.1.6.1 Running Surface*

Owing to their cushioning properties, treadmills typically have a lower compliance compared to tarseal or concrete running surfaces. There is evidence to suggest that TA-A measured overground can be substantially higher than running on some treadmills under comparable conditions [90,119,146], however the relationship between TA-A magnitude and surface compliance is not straightforward. Fu et al. [128] found no differences in TA-A across a wide range of surfaces running at 3.3 m/s, whereas Greenhalgh et al. [117] reported higher magnitudes when participants ran at 5 m/s on concrete compared to a synthetic surface, but again not at a slower velocity (3.3 m/s) (Tables 1 & 2). Conversely, Boey et al. [118] reported lower TA-A when runners ran on a more compliant woodchip trail (compared to concrete or synthetic track), but only when restricted to a slower velocity (3.1 m/s), in comparison to the runners' self-selected pace (average 3.7 m/s).

Experiments using non-running external impacts have suggested that the surface compliance explained less than 10% of the variance of the TA-A, with knee angle and muscle pre-activation explained 25% to 29% and 35% to 48%, respectively [147]. What is clear is that runners rapidly adjust leg stiffness when on different surfaces. By sensing the changes in surface compliance, runners adapt muscle activations and kinematics within a single stride [148]. For surfaces of higher compliance, leg stiffness increases, which serves to keep the path of a runner's COM the same regardless of the surface characteristics [149]. While it has not been examined, it may be that the pre-activation of muscles, and subsequent changes in leg stiffness, is the mechanism runners use to mitigate the effects of surface compliance on lower extremity acceleration.

Negative correlations have been observed between surface gradient, TA-A, TA-ML and TA-R, as well as median frequency [41,145,150]. Hamill et al. [150] reported 30% increases in TA-A on a 8.7% decline gradient, compared to level. Similar, but slightly smaller increases in TA-A magnitude were found by Chu et al. [145], these were accompanied by 51% increases in impact-related frequencies (i.e. power spectral densities within the 12–20 Hz bandwidth). These findings are in contrast to Mizrahi et al. [137] who observed similar magnitude TA-A, and a lower amplitude within the impact frequency range, from runners running on a 7% decline gradient compared to running on the flat.

#### *2.1.6.2 Running Footwear*

Conventional running footwear has been characterised by an ethylene-vinyl acetate (EVA) midsole of approximately 20 mm thickness. Initial research reported substantial reductions in the high frequency components of TA while running in footwear with a midsole, over barefoot conditions. The power spectral density of frequency components above 20 Hz were directly related to shoe midsole hardness [151]. Subsequent studies have shown, that despite some shoes demonstrating significantly reduced cushioning properties when mechanically drop tested [129], no difference in peak TA across various conventional thickness EVA footwear conditions were found [117,129] (Table 2). Tibial acceleration measured in conventional shoes has also been compared to measurements taken running barefoot [103,126], in barefoot-inspired [101,103], and minimalist shoes [98,103,126,127]. In all cases running barefoot produced higher TA magnitudes than in conventional footwear [98,103,126,152]. Additionally, TA magnitude was lower in conventional shoes, compared to the barefoot, barefoot inspired [99,103] or minimalist [98,103,126] footwear conditions.

Recent developments in running footwear have resulted in over-sized lower density midsoles (maximalist shoes), expanded thermoplastic polyurethane midsole, and orthotic inserts claiming to provide additional cushioning and reduced energy loss [101,153]. Findings have indicated that TA-A were actually greater in footwear designed to improve energy return [100]. Additionally, running in a maximalist shoe [101], custom [104,114,154], or over the counter [155] orthotics did not provide further reductions in TA-A than conventional shoes. These findings are less surprising when considered in context of the effects of surface characteristics on stiffness, where runners have been shown to increase their leg stiffness when running on softer surfaces [148].

### **2.1.7 Conclusions and Recommendations**

Clinicians and researchers commonly use tibial acceleration during running as a proxy measurement for the impact forces experienced at the tibia. There is an assumption that this measure corresponds to the acceleration of the bone, and ultimately bone stress and strain, however this is yet to be proven. For users of tibial mounted accelerometers, there are several recommendations that should be adhered to in order to achieve accurate and reproducible results. Devices should be secured firmly to the tibia to limit movement relative to the underlying bone. Differing placements of accelerometers do not necessarily give comparable results; distally attached devices provide higher values, which likely closer represent the accelerations passing through the bone. While the time domain axial tibial acceleration is the only component shown to have construct validity with respect to injury, it is important to quantify the total acceleration passing through the musculo-skeletal system. Where devices of minimal mass can be sourced, triaxial accelerometers should be used to measure all three components of acceleration. Calculating the resultant acceleration can provide a single metric that takes into account all axes, which is independent of accelerometer alignment. Selecting the appropriate filter frequencies are essential, as incorrect filtering can lead to inaccurate interpretation of data. Additional frequency analyses could be useful to provide a more thorough characterisation of the signal.

Tibial acceleration is clearly influenced by running velocity, whereby faster running velocity leads to increased peak tibial acceleration. The extent of tibial acceleration increases are likely dictated by the associated changes to stride rate and stride length. Where substantial stride length increases occur, changes may also occur in knee flexion angle and velocity, heel-strike velocity or subsequent lower extremity stiffness, which are important determinants of impact characteristics. Surface and footwear compliance also have a substantial influence on lower extremity stiffness and tibial acceleration. Runners rapidly adjust to surface compliance, and conditions that are too hard or too soft appear to result in technique modifications and increases in tibial acceleration. There are still considerable gaps in current knowledge, and the interrelationships between muscle pre-activation and fatigue, stiffness, effective mass and tibial acceleration still require further investigation, as well as how changes in these variables impact on injury risk.

## **2.2 Chapter 2 Novel Contributions**

This chapter provides a summary of the current knowledge regarding the use and interpretation of tibial acceleration with runners. While the literature does support tibial acceleration as a proxy for the impact forces experienced at the tibia, there are no clear links between a bone acceleration and actual stress and strain measured from the bone. I have identified considerable gaps in the current knowledge, specifically with regard to the between-session reliability of the measurement of tibial acceleration. Additionally, there is still further work required to understand how tibial acceleration changes with running velocity, as well as the nature of the interrelationships between lower extremity stiffness, effective mass and tibial acceleration.



## **Chapter 3:**

### **The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners.**

*This chapter comprises the following manuscript:*

Sheerin, K., Besier, T., Reid, D., Hume, P.A. (2017). The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners. *Sports Biomechanics*. 17(4) 531-540. <http://doi.org/10.1080/14763141.2017.1371214> (Appendix 1.)

### **3 Preface**

Having identified the gaps in the literature regarding the reliability of the measurement of tibial acceleration, before any intervention studies could be initiated to address tibial acceleration, there is a need to establish the short term (1-week) and long-term (6-month) reliability and variability of 3D tibial acceleration in runners to determine that this is a viable assessment method for use with runners. Running gait re-education studies, some of which have assessed TA, have used follow-up periods of between 1-3 months. However, future studies could include longer-term follow-up measurements to assess the retention of these biomechanical changes. Therefore, the longer 6-month reassessment was chosen based on the need to understand the natural variability in TA of runners over this period when not receiving an intervention. Runners train at a range of speeds, so four different running velocities were chosen to assess reliability and variability outcome measures.

### 3.0 Introduction

Running is a popular form of exercise known to have a positive influence on fitness and health; however, it is also associated with a high incidence of lower extremity overuse injuries [3,4]. When the foot strikes the ground during running, a reaction force is created as the velocity of the foot rapidly decelerates to zero, while the body continues to move forwards. The shock or acceleration due to this impact is transmitted throughout the entire body. Tibial acceleration (TA), measured as peaks in the acceleration time signals, from accelerometers attached to the skin overlying the tibia, are used to represent the acceleration of the bone in runners [105].

Research documenting TA in healthy and injured runners has used running over ground and on treadmills, across a range of speeds and conditions (i.e., different gradients, fatigue states, stride lengths and frequencies, etc.) [67,120,122,125,136]. Tibial acceleration occurs in three dimensions with respect to a local tibial coordinate frame: axial, antero-posterior and medio-lateral [80]. Axial TA magnitude (along the long axis of the tibia) under these conditions has ranged from 3.0 to 24.6 g [75,90].

The repetitive impact loading that occurs with running is thought to play an important role in the pathophysiology of common running injuries, particularly bony fatigue fractures (commonly termed stress fractures) [10,24,62,156,157]. While the development of fatigue fractures is multi-factorial in nature, there are biomechanical risk factors related specifically to the development of tibia fatigue fractures in runners [35]. Axial peak tibial acceleration (PTA) magnitude has been reported as one of a number of biomechanical variables that discriminated female runners with a history of tibial fatigue fractures (PTA:  $6.5 \pm 3.4$  g) from uninjured controls (PTA =  $5.5 \pm 2.5$  g) [31]. Additionally, it has been demonstrated that for every 1 g increase in axial TA, the likelihood of a history of tibial fatigue fracture increased by a factor of 1.4 [31].

After measuring all three planar components of TA at two different running velocities, Lafortune and Hennig [31] reported that the greatest peak magnitudes were found in different planes at different velocities. This lead the authors to conclude that the proportion of TA represented in each plane does not remain constant in different situations, and therefore in order to accurately quantify TA magnitude, it was important to measure all three planes. However, since 1991, only three published studies have reported TA in all three planes of acceleration [39–41]. When measuring TA using a uniaxial accelerometer, there is a need for careful alignment of the device to the tibia [158]. If the correct axial alignment is not achieved, the acceleration will not accurately reflect the axial component of TA. Measuring TA 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 align the device.

If accelerometers are to be used by coaches and clinicians, an appreciation of the reliability of the devices is required. Natural variability in the TA of runners over longer periods of normal training will also help establish sample sizes and statistically meaningful differences for future intervention trials. There is currently an absence of published studies assessing the reliability and variability of TA in runners. Recent running gait re-education studies, some of which have assessed TA, have used follow-up periods of between 1 and 3 months [82,159]. However, future studies should include longer term follow-up measurements to assess the retention of these biomechanical changes [160].

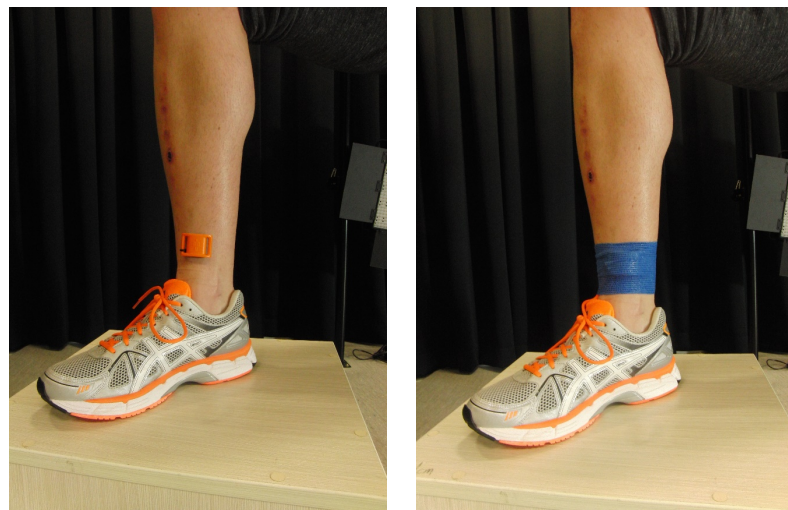
Based on the needs identified, the aims of this study were to determine between-session reliability and variability of resultant TA at four running speeds (at one-week and six-month intervals). It was hypothesised that the resultant TA calculated from accelerometers attached to the tibiae would be repeatable at one week and six months in healthy runners at each of the running speeds assessed. As additional research aims, we sought to investigate differences in the natural variability in TA over a longer period of time (6-months), the smallest worthwhile change in TA and how this related to measurement error. The longer six-month reassessment was chosen based on the need to understand the natural variability in TA of runners over this period when not receiving an intervention.

### **3.1 Methods**

Testing was conducted in a biomechanics laboratory with identical set-up for each session. Procedures were approved by institutional human research ethics (Auckland University of Technology Ethics Committee #1351) and participants signed written informed consent. Fourteen male runners (age =  $33.6 \pm 11.6$  years; height =  $1.77 \pm 0.05$  m; mass =  $75.6 \pm 9.5$  kg) who were injury free at the time of data collection volunteered for this study. They had regularly participated in running for  $8.7 \pm 8.1$  years, and ran on average  $30.3 \pm 25.5$  km over  $4.4 \pm 5.2$  training sessions each week. Data were collected at baseline and at one week for these 14 runners. At six months, three runners had sustained injuries, one had moved overseas and two could not be contacted, leaving eight runners (age =  $37.4 \pm 5.8$  years; height =  $1.75 \pm 0.05$  m; mass =  $72.3 \pm 6.0$  kg) to complete the final reassessment. Runners were injury free at the time of reassessment, and also reported no injuries between the data collection sessions.

### 3.1.1 Equipment

Triaxial accelerometers (IMeasureU, Auckland, New Zealand; mass 12 grams; range  $\pm 16$  g; resolution—16 bit) were securely attached bilaterally over each participant's distal antero-medial tibia bones, 3 cm superior to the crest of the medial malleolus, by a single experienced practitioner. One axis of each accelerometer was aligned with the long axis of the tibia (Figure 3). To minimise oscillations of the accelerometers, they were attached with double-sided tape to the skin, then wrapped in elastic adhesive bandage (Amtech, Auckland, New Zealand) [110]. Runners wore standardised neutral running shoes (Asics Kudrow, Kobe, Japan) for all laboratory tests, and conducted training in their own running shoes. Running trials were conducted on an instrumented treadmill (Bertec, Columbus, OH, USA) that provided minimal shock absorption.



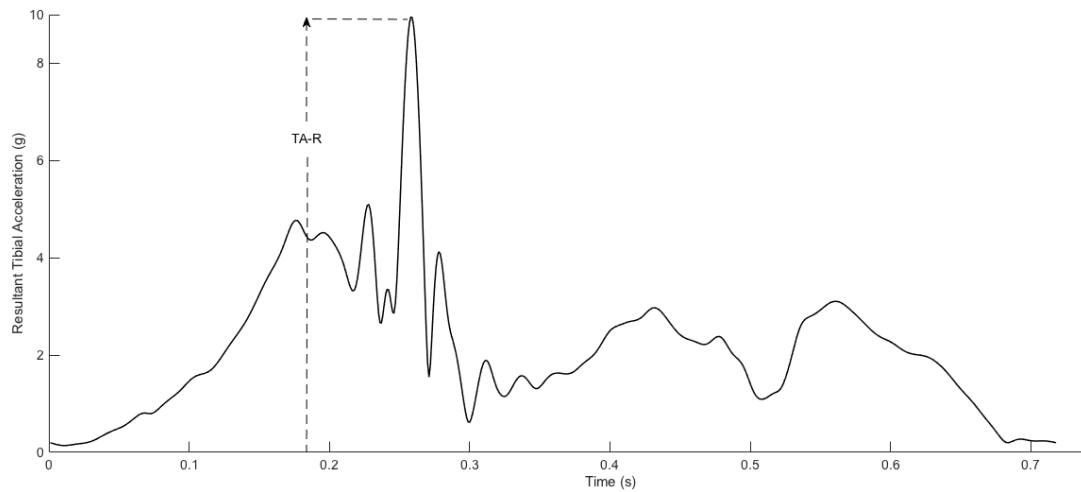
**Figure 3: Accelerometer attachment methods.**

### 3.1.2 Procedures

Height and mass were measured according to standard protocols [161]. Following a 5-minute warm-up and familiarisation at a self-selected pace, runners ran for two minutes each at increasing speeds of 2.7, 3.0, 3.3 and 3.7 m/s (a total of 8 minutes of data collection), with a stationary pause of 30 sec between each speed. Data were logged to the onboard memory of the accelerometers at 1000 Hz for the duration on the running trials and then downloaded after each session for processing. and 10 min and 4 h and 20 min [162], as well as speeds that have been used in previous research [97,110,159]. Data were logged to the onboard memory of the accelerometers at 1,000 Hz for the duration on the running trials, and then downloaded after each session for processing. Runners were free to adopt their natural running technique at each running speed. Kinematics were not controlled, and no feedback was provided to runners.

### 3.1.3 Data Processing

Data processing was conducted using a custom Matlab script (Mathworks, MA, USA). For each running speed, 50 s of data were extracted from the middle of each run. A fourth order, dual pass 60 Hz Butterworth low-pass filter was applied and the resultant accelerations were calculated as  $r = (x^2 + y^2 + z^2)^{0.5}$ . PTA magnitudes for each step were determined from the resultant acceleration for each running speed [110] (Figure 3).



**Figure 4: Resultant tibial acceleration from running at 3.0 m/s**

### 3.1.4 Data Analyses

Group mean and standard deviations (SDs) were calculated for the resultant PTAs across the four running speeds comparing baseline to one week and six months. Data were log transformed to reduce bias arising from non-uniformity of error [163]. Initial analyses revealed no statistically significant differences between legs, and therefore data for the two limbs were condensed for all subsequent analyses. Reliability and variability outcomes were presented as percentage changes. Measurement variability outcomes included intraclass correlation coefficients (ICC) and the typical error of the measurement expressed as a coefficient of variation percentage (CV%). As outlined by Bradshaw et al. [164], the use of a variety of measurement reliability and variability outcomes allows for a robust decision to be made on the appropriateness to use a test measure of interest.

An ICC < 0.70 was indicative of 'poor' agreement and high measurement variability,  $0.7 \leq \text{ICC} \leq 0.80$  represented a questionable outcome and an ICC > 0.8 represented an excellent outcome [163,165]. A CV of <10% was considered small variation. Between-session reliability measures included percentage differences in the means (MDiff%) and Cohen's effect sizes (ES). Effect sizes

were interpreted as trivial (0.0–0.1), small (0.11–0.3), moderate (0.31–0.5), large (0.51–0.7), very large (0.71–0.9) or extremely large (0.91–1.0) [166]. An ES of less than 0.3 is indicative of a minimal change in a variable of interest from one session to the next. Average measurement variability was interpreted as ‘small’ when the ICC was >0.70 and the CV was <10%, ‘moderate’ when ICC was <0.70 or CV was >10% and ‘large’ when ICC < 0.70 and CV > 10%. Average reliability was interpreted as ‘good’ when the MDiff% was less than 5% and the ES was trivial to small. Average reliability was interpreted as ‘moderate’ when the aforementioned criteria for ‘good’ were breached for either the MDiff% or the ES (MDiff > 5% or ES = moderate to large). Average reliability was categorised as ‘poor’ when both the MDiff% and the ES criteria were breached (MDiff > 5% and ES = moderate to large) (Bradshaw et al., 2010). Smallest worthwhile change (SWC%) was calculated as 0.2 multiplied by the between-participant SD across the first two sessions, as a percentage of the mean across two days [164]. A performance/noise ratio was calculated by dividing SWC% by CV%, where a ratio above 1.00 indicated that the performance variability was less than the SWC [164].

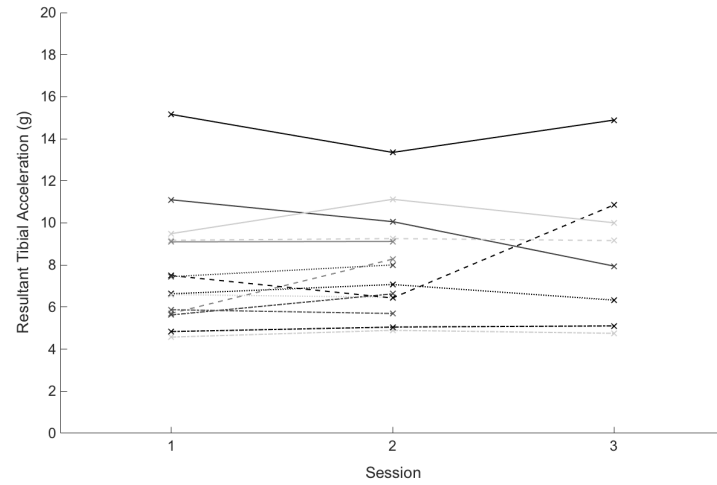
### 3.2 Results

An average of  $61.0 \pm 1.5$  steps were analysed for each runner’s left and right legs for each running speed for all of the three testing sessions. Descriptive, reliability and variability statistics for the resultant PTA were compared for the 14 runners between baseline and one week, and for the eight runners between baseline and six months (Figure 5). The mean resultant PTA values ranged from 7.8 to 12.0 g for the baseline to one-week comparison, and 8.6 to 12.9 g for the baseline to six-month comparison (Table 1). The average mean differences in resultant PTA between the 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. No measures of resultant PTA exceeded an ES of 0.14.

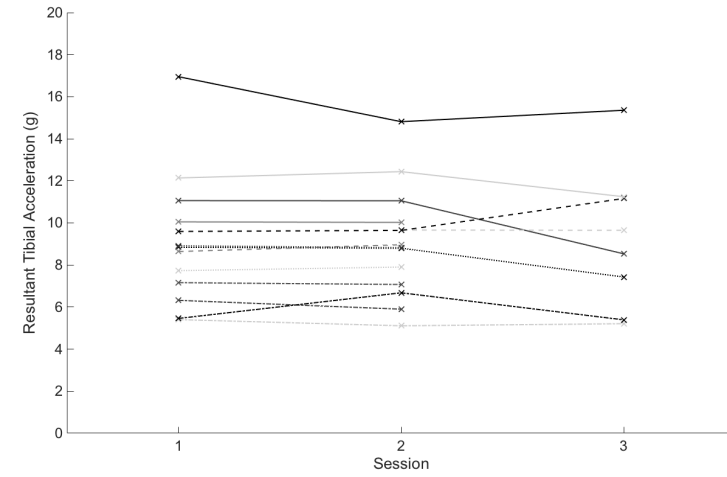
Typical error of the measurement (CV) values ranged from 6.6 to 11.3% for the baseline to one-week comparison, and 8.7–15.1% for the baseline to six-month comparison. Four of the eight comparisons were categorised as ‘small’ variation (<10%). ICC ranged from 0.90 to 0.96 for the baseline to one-week comparison, and 0.89–0.95 for the baseline to six-month comparison, which indicated excellent agreement and low measurement variability between testing sessions. The SWC% provides a statistical estimate of the percentage change in performance required for a significant improvement (such as from a training intervention[164]. A reduction between 6.0 and 6.9% (Table 3) would be required to achieve a statistical improvement in resultant PTA. The

performance/noise ratios indicate that the resultant PTA variation (CV%) when running at all speeds assessed is higher than what is considered a significant training improvement (SWC%).

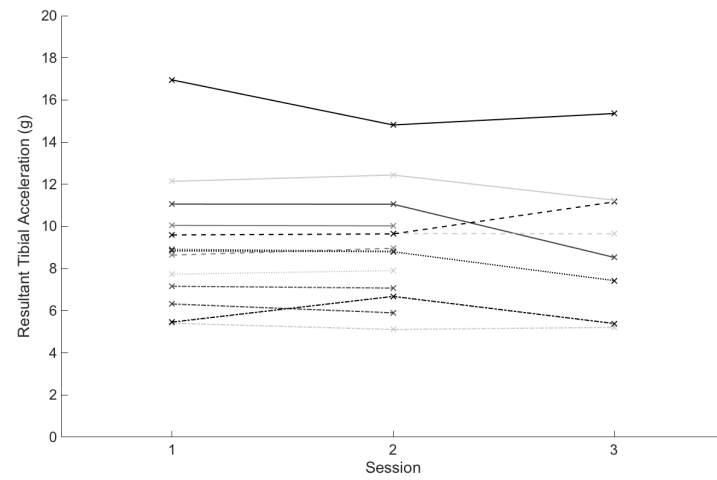
(a)



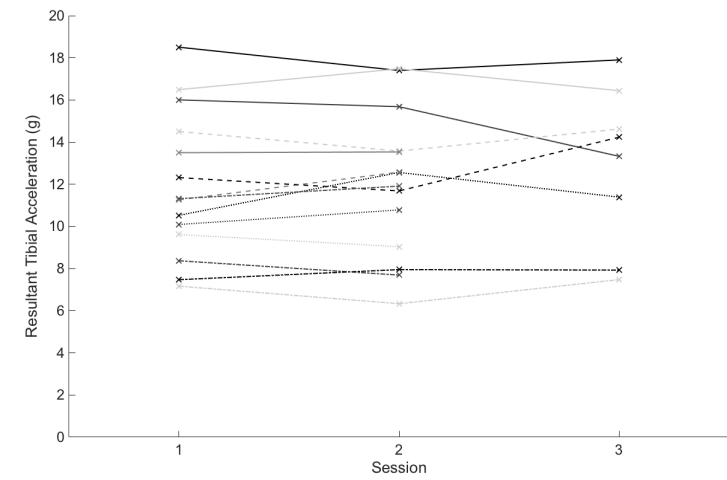
(b)



(c)



(d)



**Figure 5: Individual tibial acceleration results for baseline, one week and six months.**

**(a) velocity 2.7 m/s. (b) velocity 3.0 m/s. (c) velocity 3.3 m/s. (d) velocity 3.7 m/s.**



**Table 3: Peak resultant acceleration between-session variability and reliability for four running speeds.**

Speed (m/s)	2.7	3.0	3.3	3.7
<b>Baseline vs 1-Week (n=14)</b>				
Baseline Mean $\pm$ SD (g)	7.8 $\pm$ 2.9	9.1 $\pm$ 3.1	10.4 $\pm$ 3.4	11.9 $\pm$ 3.6
1-week Mean $\pm$ SD (g)	8.0 $\pm$ 2.5	9.1 $\pm$ 2.7	10.6 $\pm$ 3.1	12.0 $\pm$ 3.5
Mdiff (g)	0.2	0.1	0.1	0.0
CV% (90%CL)	11.3 (9.2-14.9)	6.6 (5.3-8.6)	10.5 (8.5-13.7)	7.6 (6.2-10.0)
ICC (90%CL)	0.90 (0.81-0.95)	0.96 (0.93-0.98)	0.91 (0.83-0.95)	0.95 (0.90-0.97)
Variability rating	Moderate	Small	Moderate	Small
Mdiff% (90%CL)	3.5 (-1.4-8.7)	0.0 (-0.10-0.09)	1.6 (-2.9-6.3)	0.5 (-2.8-3.9)
Effect Size	0.07	0.02	0.04	0.02
Reliability rating	Good	Good	Good	Good
Use	Yes	Yes	Yes	Yes
SWC (%)	6.87	6.35	6.18	5.96
PNR	0.61	0.96	0.59	0.78
<b>Baseline vs 6-Months (n=8)</b>				
Baseline Mean $\pm$ SD (g)	8.6 $\pm$ 3.4	9.7 $\pm$ 3.4	11.7 $\pm$ 3.8	12.9 $\pm$ 4.3
6-months Mean $\pm$ SD (g)	8.6 $\pm$ 3.3	9.2 $\pm$ 3.4	11.1 $\pm$ 3.5	12.9 $\pm$ 3.7
Mdiff (g)	0.1	0.5	0.5	0.0
CV% (90%CL)	15.1 (11.5-22.4)	11.6 (8.9-17.0)	8.7 (6.7-12.8)	9.4 (7.2-13.7)
ICC (90%CL)	0.89 (0.75-0.95)	0.92 (0.83-0.97)	0.95 (0.88-0.98)	0.94 (0.86-0.97)
Variability rating	Moderate	Moderate	Small	Small
Mdiff% (90%CL)	1.4 (-7.0-10.7)	5.3 (-11.5-1.3)	4.0 (-8.8-1.1)	1.7 (-3.8-7.5)
Effect Size	0.02	0.14	0.14	0.01
Reliability rating	Good	Moderate	Good	Good
Use	Yes	Yes	Yes	Yes

### 3.3 Discussion and Implications

The current study is the first to report reliability and variability for triaxial accelerometers placed on the tibiae of runners in a test–retest scenario. Resultant PTA at all speeds for the baseline comparisons with one week and six months showed ‘good’ to ‘moderate’ reliability, and ‘small’ to ‘moderate’ measurement variability. We can therefore be confident that resultant PTA can be used with male runners, running on a treadmill at velocities between 2.7 and 3.7 m/s, to monitor their impacts throughout a six-month intervention. Careful attention to the characteristics and placement of the device, as well as data analysis procedures, outlined in this study is required to uphold the findings.

One benefit of measuring a resultant TA variable is that there is not the same need to carefully align one of the sensitive axes of the accelerometer with the tibia anatomical coordinate frame to obtain valid and reliable results. As a result, the device can be used by coaches, or runners themselves, who do not necessarily have an appreciation for aligning the device to the underlying anatomy.

The mean resultant PTA values for our runners (7.8–12.9 g) were consistent with values reported in three other studies of runners at comparable speeds in a single session [39–41]. The quantitative measures of reliability and variability for the two slower speeds were not as good for the six-month comparison as they were for the one-week comparison. It is possible that some runners were not able to adopt their natural running technique at slower speeds, and therefore demonstrated more variability in their movement patterns [167].

The performance/noise ratio revealed that the between-session variation in resultant PTA would mask significant training effects when determined from calculations of SWC in this group of runners. Therefore, a performance improvement could only be confidently identified if the change was greater than the measured variation. While this study has established the level of measurement reliability and variability that could be expected in the lab for this group of runners, individual measures of SWC% and CV% would need to be undertaken for serial testing of other runners.

One limitation of this study is that we recorded PTA in the laboratory, on a treadmill at constrained running speeds, which are likely different from those measured in the field. We would anticipate greater variability in PTA during field-based assessment due to different terrain and variable running speed [41]. Additional studies need to be conducted using accelerometers in field-testing

sessions to understand this variability. Runners also ran for 2 min at each speed in a progressively faster order. While it is conceivable that a learning or fatigue effect could have influenced the results, adequate warm-up and familiarisation were provided, and the short bouts of running were unlikely to have induced fatigue in this cohort of regular runners.

### **3.4 Conclusion**

Based on our comparisons of baseline to one week and six months for resultant PTA, we can be confident that triaxial accelerometers attached to the tibiae of runners can be used to assess and monitor their impacts throughout a six-month intervention.

### **3.5 Chapter 3 Novel Contributions**

In this chapter the reliability and variability statistics for TA-R collected at four common running speeds on a treadmill are presented. While runners demonstrated marginally lower reliability and higher variability over six months, in all cases TA-R results indicated 'good' to 'moderate' reliability, and 'small' to 'moderate' variability. I am confident, based on the data collected, that TA-R can be used as a reliable metric in both the short- and long-term at a range of running speeds.

## **Chapter 4:**

### **The influence of running velocity on resultant tibial acceleration in runners.**

*This chapter comprises of the following manuscript:*

Sheerin, K., Besier, T., Reid, D. (2018). The influence of running velocity on resultant tibial acceleration in runners. *Sports Biomechanics*. DOI: 10.1080/14763141.2018.1546890 *Published online ahead of print* (Appendix 2.)

#### **4 Preface**

A small number of studies have examined the effects of running velocity on the magnitude of PTA. The findings suggest that running at higher velocities is associated with increased PTA-A. However, there is an absence of PTA-R data, and the inter-relationships between PTA-R and running velocity are still not well understood. The purpose of this study was to investigate the relationship between TA-R and running velocity, and to establish a normative database of tibial acceleration profiles. As a tertiary aim, I also sought to identify runners who might be classified as high-impact based on their TA-R magnitude at each of the running velocities.

## 4.0 Introduction

Running is a popular activity with easy accessibility and relatively low cost, however endurance running is associated with a high incidence of overuse injuries [168]. The transient impact shock that results when the foot impacts the ground surface is thought to play a role in the pathophysiology of many common running injuries [10,24,28]. The axial component of peak tibial acceleration (PTA-A) magnitude has been reported as one of a number of biomechanical variables to discriminate female runners with a history of tibial fatigue fractures from uninjured controls (PTA-A =  $6.5 \pm 3.4$  vs.  $5.5 \pm 2.5$  g, respectively) [31]. Additionally, a PTA-A of 8 g (running velocity of 3.7 m/s) was used as upper threshold to identify runners as high risk of injury and enrol them in gait retraining interventions. This value was one standard deviation (SD) above the mean of 171 uninjured runners [82,169].

The shock waves generated during ground impact are transferred through the foot into the tibia, and can be estimated by measuring the acceleration experienced by the tibia during impact. The PTA-A magnitude has attracted substantial research attention, however it has long been proposed that the other orthogonal directions could also be important [75,81]. More recently, researchers using triaxial accelerometers reported the vector resultant PTA (PTA-R) [39–41]. The PTA-R provides a single metric that takes into account all measurement axes. Additionally, it has shown to be more reliable between measurement sessions, which is most likely due to there being no need to align the sensitive axis of the device to the long axis of the tibia [87].

A small number of studies have examined the effects of running velocity on the magnitude of PTA. The findings suggest that running at higher velocities is associated with increased PTA-A, irrespective of running surface, footwear, running experience, or whether the speed was fixed, or self-selected [117,118,126]. Further to these trends, a linear regression analysis revealed that average PTA-A increased by 34% for each 1.0 m/s increase in running velocity when 10 well-trained runners increased their velocity from 3.4 to 5.4 m/s [108]. The PTA-A measured on these runners demonstrated a linear relationship with running velocity, but this differed between 0.15 and 0.68, suggesting large variation between runners. While the authors claim that the best-fit linear relationship was in 'good agreement' with the experimental data, no statistics were provided to support this [108].

Only one group of researchers have used a triaxial bone-mounted accelerometer to report the different components of PTA, from a single recreational runner, at two velocities (3.5 and 4.7 m/s) [75,170]. They report repeatable patterns of PTA for the two velocities on 0% and 7.5% incline,

with increases in peak magnitudes across all axes. Sheerin, et al. [87] report mean PTA-R, measured from triaxial skin-mounted devices, increased from 7.8 to 12.9 g, as running velocity increased from 2.7 to 3.7 m/s. However, these findings were from a relatively small cohort of 14 runners, and no additional analysis of the relationship between velocity and PTA-R were undertaken.

While changes in a range of kinetic and kinematic variables have been reported with increasing running velocities, there is an absence of PTA-R data, and the inter-relationships between PTA-R and running velocity are still not well understood [171,172]. Despite this, researchers have investigated the effects on PTA-R of differing foot strike patterns and footwear conditions, while running at fixed and self-selected velocities [40,88]. Additionally, PTA-R has been used as augmented feedback to modify mechanics while running at self-selected velocities. The lack of appreciation of differences, or changes, in running velocity may limit the interpretation of findings, or impact on the clinical decision making on which runners are appropriate for interventions. The primary aim of this study was to investigate the impact of increasing running velocity on PTA-R in injury-free runners. It was hypothesised that there would be a linear relationship between PTA-R and running velocity. The secondary aim was to collect a representative database using PTA-R as a key biomechanical variable. As a tertiary aim, we sought to identify runners who might be classified as high-impact based on their PTA-R magnitude.

## **4.1 Methods**

Testing was conducted in a biomechanics laboratory. Runners were invited to participate in this study if they had consistently run three or more times weekly, to a combined distance of at least 20 kilometres, or greater than six months. Procedures were approved by institutional human research ethics (Auckland University of Technology Ethics Committee #15181), and participants signed written informed consent. Sixty-five male and 20 female runners (age =  $39.6 \pm 9.0$  years; height =  $1.76 \pm 0.09$  m; mass =  $73.9 \pm 11.0$  kg), who were injury free at the time of data collection, volunteered for this study. Participants had regularly participated in running for an average of  $8.5 \pm 7.3$  years, and ran  $45.0 \pm 23.2$  km over  $3.9 \pm 1.5$  training sessions each week (Table 4)

**Table 4: Participant characteristics used in Chapter 4.**

	Male (n=65)	Female (n=20)	Combined (n=85)
Age (years)	38.8 ±8.8	41.8 ±9.3	39.6 ±9.0
Height (m)	1.80 ±0.06	1.66 ±0.05	1.76 ±0.09
Mass (kg)	77.9 ±9.0	61.4 ±6.3	73.9 ±11.0
Running history (years)	8.3 ±6.5	9.2 ±9.4	8.5 ±7.3
Weekly running distance (km)	42.6 ±23.2	52.5 ±22.0	45.0 ±23.2
Weekly running frequency (days / week)	3.6 ±1.5	4.8 ±1.4	3.9 ±1.5

#### 4.1.1 Equipment

Triaxial accelerometers (IMeasureU, Auckland, New Zealand; mass 12 grams; range  $\pm 16$  g; resolution - 16 bit) were securely attached bilaterally over each participant's distal antero-medial tibiae, 3 cm superior to the crest of the medial malleolus, by a single experienced practitioner (Figure 3). To minimise oscillations of the accelerometers they were firstly attached with double-sided tape to the skin, then wrapped tightly in elastic adhesive bandage (Amtech, Auckland, New Zealand) [110]. Runners wore standardised neutral running shoes (Asics Kudrow, Kobe, Japan) and ran on an instrumented treadmill (Bertec, Columbus, OH, USA) that provided minimal shock absorption.

#### 4.1.2 Procedures

Height and mass were measured according to standard protocols [161]. Following a 5-minute warm-up and familiarisation at a self-selected pace, runners ran for two minutes each at increasing speeds of 2.7, 3.0, 3.3 and 3.7 m/s (a total of 8 minutes of running), with a stationary pause of 30 s between each speed. These running velocities were chosen based on uniform increases in running pace, being 6.0, 5.5, 5.0 and 4.5 min/km. The total running time for each speed was two minutes (including acceleration and deceleration to and from stationary). To accommodate slight variations in the acceleration time between speeds and runners, data collection was started after one minute of running. Data were logged to the onboard memory of the accelerometers at 1000 Hz for the duration of the running trials, and then downloaded after each session for processing. Runners were free to adopt their natural running technique at each running speed. Kinematics were not controlled, and no feedback was provided to runners.

#### 4.1.3 Data Processing

Data processing was conducted using a custom Matlab script (Mathworks, MA, USA). For each running speed, 60 s was allowed for acceleration and gait stabilisation, and then 50 s of data were



extracted. The final 10 s was excluded to remove any potential gait variation that could occur as runners approached the end of the running period. A fourth order, dual pass 60 Hz Butterworth low-pass filter was applied and the resultant accelerations were calculated as  $PTA-R = (x^2 + y^2 + z^2)^{0.5}$ . Peak tibial acceleration magnitudes for each step were determined from the resultant acceleration for each running speed [110] (Figure 4).

#### 4.1.4 Data Analyses

To determine the similarity of PTA-R values measured from participants' left and right legs, a symmetry angle (SA) was used to quantify the level of bilateral asymmetry within each runner. A SA value of 0% indicated perfect symmetry (equivalence between the values), and a value of 100% reflects two values that are equal and opposite in magnitude [173]. The mean symmetry angle (encompassing left and right sides) for PTA-R was lower than  $0.51 \pm 4.97\%$  for each running velocity, less than that previously reported for uninjured runners, and indicating close symmetry in this cohort of runners [73]. As a result, the left and right leg values were averaged to provide a single PTA-R value for each running velocity, and this value was used for the statistical analyses outlined below.

Statistical analyses were performed using SPSS statistics package version 24 (IBM Corporation, Armonk, NY, USA) with the level of significance set at  $p < 0.05$  throughout. Group mean and SDs were calculated for the PTA-R across the four running velocities. One-way repeated measures analysis of variance (ANOVA) were performed to determine whether there was a significant main effect for running velocity on PTA-R. Where a significant main effect was found, pairwise comparisons with Bonferroni correction post-hoc analyses were performed. To determine the strength of the relationship between running velocity and PTA-R, a Pearson correlation coefficient and multiple linear regression were calculated, with running velocity as the independent variable and PTA-R as the dependent variable. The correlation magnitude was interpreted as trivial (0.0-0.1), small (0.1-0.3), moderate (0.3-0.5), large (0.5-0.7), very large (0.7-0.9), or extremely large (0.9-1.0) [166]. From the resulting regression equation, the predicted increase in PTA-R associated with a one unit increase in running velocity was calculated.

To interpret runners' PTA-R - running velocity relationship individually, additional analyses were carried out. Mean left-right resultant PTA change scores were calculated for subsequent velocities as  $\Delta PTA-R = PTA-R_{Speed2} - PTA-R_{Speed1}$ . The average SD was calculated as  $SD_{AVG} = (SD_{(Speed1)} + SD_{(Speed2)})^{0.5}$ . Clear changes in PTA-R were classified where the change in mean between the two velocities exceeded the average SD for the two velocities. With an increase in

running velocity, a negative change in mean that exceeded the average SD indicated a clear reduction in PTA-R. To provide insight into potential PTA-R responses, 12 individual runners were selected as examples to represent different response patterns. Four runners were selected to represent the category being, 'typical increase' responders, or those that represented changes in PTA-R similar to the mean of the wider group. An additional four runners were 'large increase' responders, whereby the change in PTA-R from one speed to the next was consistently greater than one  $SD_{AVG}$ . The final four runners were examples of 'inconsistent change' responders, who demonstrated a reduction in PTA-R from one speed to the next.

The classification of runners as high-impact was based on the methods of Crowell et al. [82], whereby a threshold PTA-R, corresponding to one SD above the group mean, was defined for each velocity. The number of runners who exceeded these thresholds with either one, or both legs, were identified.

## 4.2 Results

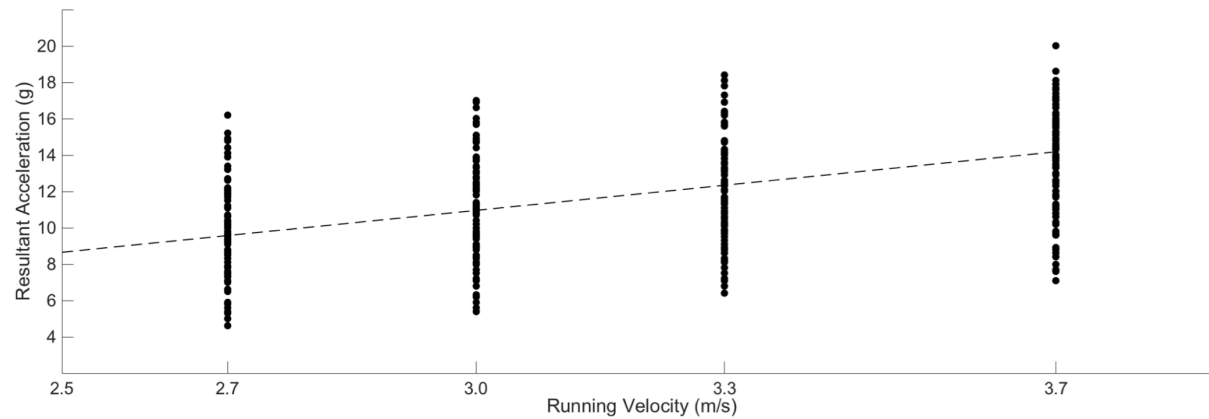
For the 85 runners assessed, the PTA-R ranged from a minimum of 4.5 g to a maximum of 20.6 g. There was a significant effect of running velocity on PTA-R ( $F[3, 83] = 172.2, p < 0.05$ ), with the mean change in PTA-R between speeds ranging from 1.1 to 1.4g (Table 5 and Figure 6). Post-hoc comparisons using the Bonferroni correction indicated that the mean PTA-R measured at each velocity were significantly different from each of the other velocities ( $p < 0.05$ ). Further analysis revealed a statistically significant Pearson correlation ( $r = 0.44, p < 0.05$ ) and regression analysis ( $r^2 = 0.19, p < 0.05$ ). The regression analysis also revealed that for speeds between 2.7 and 3.7 m/s, for every 0.1 m/s increase in running velocity, PTA-R increased by 0.38 g (Figure 6). Between 17 and 20 participants were identified as having a PTA-R result on at least one leg that exceeded the one SD upper threshold, potentially classifying them as high-impact runners.

**Table 5: Resultant tibial acceleration characteristics by running velocity.**

Velocity (m/s)	2.7	3.0	3.3	3.7
PTA-R $\pm$ SD (g)	9.8 $\pm$ 2.7 <sup>b, c, d</sup>	11.0 $\pm$ 3.0 <sup>a, c, d</sup>	12.1 $\pm$ 3.1 <sup>a, b, d</sup>	13.5 $\pm$ 3.1 <sup>a, b, c</sup>
PTA-R range (g)	4.5–19.0	4.9–19.8	5.6–19.7	6.5–20.6
Mean change score (g)		1.2 $\pm$ 1.0	1.1 $\pm$ 0.8	1.4 $\pm$ 0.9
PTA-R 1-SD threshold (g)	12.5	14.0	15.2	16.6
Symmetry Angle (%)	0.22 $\pm$ 5.23	0.28 $\pm$ 5.14	0.51 $\pm$ 4.97	0.42 $\pm$ 5.17
Runners exceeding threshold	17	19	18	21

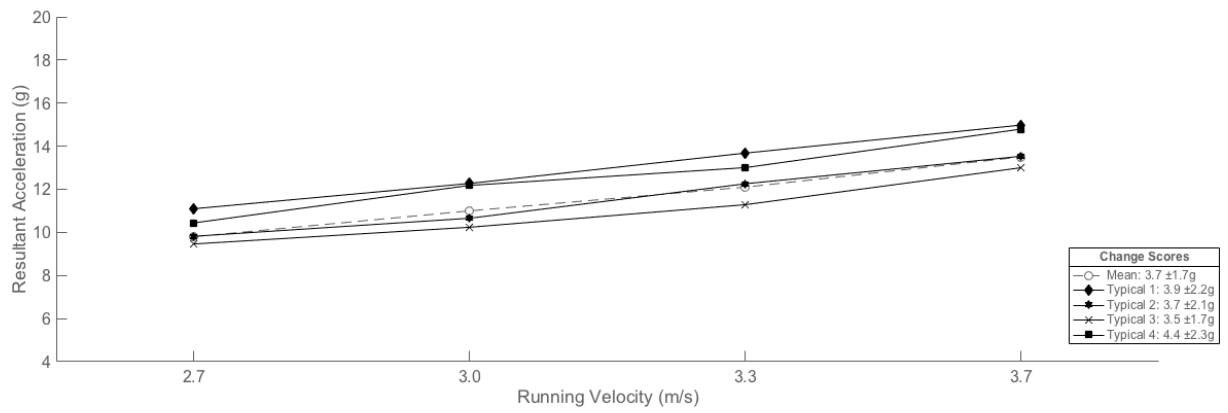
*PTA-R, Resultant peak tibial acceleration; SD, standard deviation; <sup>a</sup> Significantly different ( $p < 0.05$ ) from 2.7 m/s;*

*<sup>b</sup> Sig. different from 3.0 m/s; <sup>c</sup> Sig. different from 3.3 m/s; <sup>d</sup> Sig. different from 3.7 m/s*

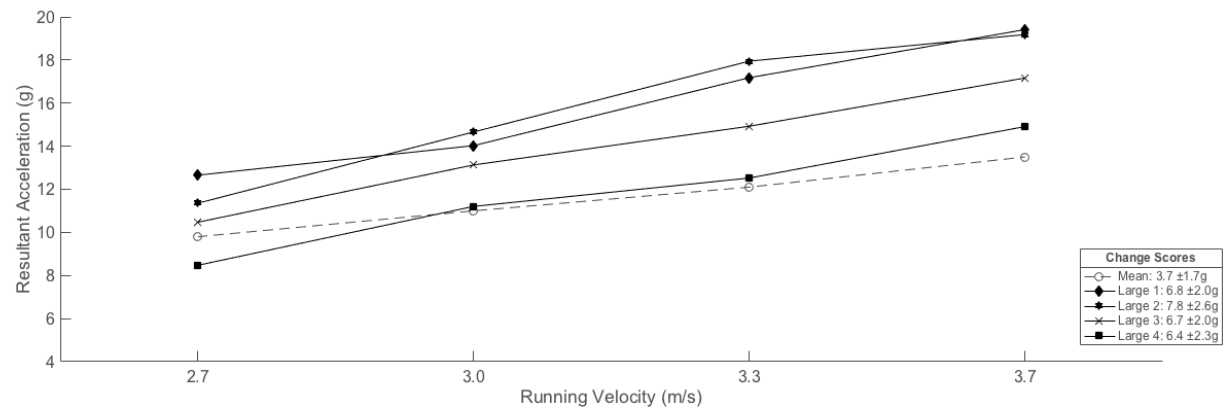


**Figure 6: Individual resultant tibial acceleration observations at 2.7, 3.0, 3.3 and 3.7 m/s and a linear regression line.**

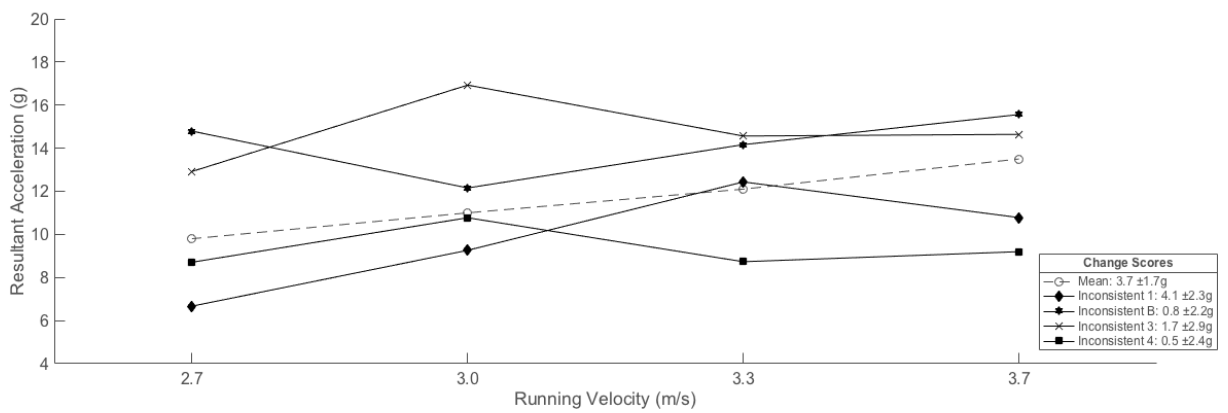
(a)



(b)



(c)



**Figure 7: Example individual resultant tibial accelerations depicting (a) typical increases (a) large increases (a) inconsistent changes, with increasing running velocity.**

### 4.3 Discussion and Implications

Overall, increasing velocity had a statistically significant effect on PTA-R across the four speeds. For the four velocities, there was a statistically significant difference in PTA-R between each of the conditions. The relationship between running velocity and PTA-R was shown to be a moderate positive correlation, however running velocity only accounting for 19% of PTA-R magnitude. While it goes beyond the scope of this study, it is likely that other variables, such as stride length, cadence, joint position, effective mass and muscle function account for large proportions of the remainder [27,120,142]. While differing in participant numbers (85 versus 10), training level (broad range versus well trained), running velocity range (2.7-3.7 m/s versus 3.35-5.36 m/s), and components analysed (resultant versus axial direction only), the results of the multiple linear regression analysis were in line with those reported by Clarke et al. [143]. Therefore, the hypothesis that a linear relationship exists between running velocity and PTA-R, can be accepted.

The linear regression analysis did predict that PTA-R magnitude would be  $-0.18 \pm 1.35g$  at 0 m/s. Owing to gravitational acceleration, the true and correct acceleration at rest is 1 g (9.81 m/s). While this value is included in the spread of data one SD from the mean, it may still suggest that the slope of the regression line (3.8x) is too steep, or that the relationship is not linear at slower velocities. It is likely that at slower, and potentially faster running velocities, that the relationship is non-linear, and there are probably inflection points at different velocities, that are individually defined.

The PTA-R magnitudes measured from this cohort of runners were in line with those demonstrated by Glauberman and Cavanagh [40], but higher than those reported by Wood and Kipp [39]. Several reasons may exist for these differences, with the most likely being the impact shock absorption or cushioning introduced through use of a standard treadmill [39,146]. The treadmill used in this study has minimal shock absorption characteristics, and therefore more closely represents overground running as studied by Glauberman and Cavanagh [40].

One reason for measuring PTA has been to screen runners to determine those who may be at risk of injury [35]. Using a one SD upper threshold, for the cohort of runners assessed as part of this study, 10 runners (12%) were identified with one leg above this value at 2.7 m/s, with an additional seven runners (8%) having both legs above the threshold. The total number of runners crossing the threshold increased at 3.7 m/s, with nine runners (11%) having just one leg exceeding, while an additional 12 (14%) exceeded the threshold with both legs. While this may be a useful statistical method of identifying potentially high-impact runners, it is acknowledged that

those experiencing high impacts may not necessarily go on to develop injury. Due to differing mass qualities and tissue material properties of the tibia as a result of prior loading events, as well as other factors, such as runner caliber and body mass, the tibia will experience force and acceleration differently across runners. It should also be noted that the runners identified as potentially high-impact, are classified as such relative to the other runners in this study only. Additionally, this classification is velocity dependent, such that runners might be classified as high-impact at one velocity and not at other.

The majority of runners demonstrated a positive linear trend in PTA-R, whereby they experienced fairly consistent increases with each increment in running velocity (Figure 7a). There were however a small group of runners that responded in different ways. Some runners demonstrated large changes in their PTA-R (Figure 7b). The velocities used in this study would have been easy for some and hard for other runners. The literature shows us that the sum force of ground impact, thus tibial acceleration, is due to both the motion of the lower leg, and the movement contributions of the rest of the body [143]. Greater exertion introduced with successive increases in velocity, could alter running movement patterns, the ground impact effect of this two-mass approach, and therefore the tibial acceleration magnitudes experienced. In contrast to this, some runners demonstrated a reduction in their PTA-R, from one speed to the next (Figure 7c). While it goes beyond the scope of this analysis, it is possible that these runners have particular speeds that allow for functionally preferred coordination patterns resulting in lower PTA-R, in tune with the theories of dynamical systems theory [174]. These findings do highlight the need for a personalised approach to understanding the response of each athlete.

One limitation of this study is that PTA-R was recorded in a laboratory, on a treadmill at constrained running velocities, which are likely different from those measured in the field. We would anticipate greater variability in PTA-R during field-based assessment due to different terrain and variable running velocity [41]. While current sensors are well placed to be used in the field, additional studies need to be conducted using accelerometers in these environments to better understand this variability. Additionally, participants ran for two minutes at each velocity in a progressively faster order. While it is conceivable that a learning or fatigue effect could have influenced the results, adequate warm-up and familiarisation were provided, and the short bouts of running were unlikely to have induced fatigue in this cohort of regular runners. The four running velocities were chosen to reflect a range of speeds runners would normally train at, however it is acknowledged that runners do run at other velocities, both inside and outside of this range. The relationships identified in this study may not hold true for velocities outside of this range, therefore

additional research needs to expand on the velocities studied, which will undoubtedly strengthen the understanding of the relationship between velocity and PTA-R.

#### **4.4 Conclusion**

Resultant PTA is clearly influenced by running velocity, with average peak magnitudes increasing with faster running velocities, with a moderate correlation between these variables, and 19% of tibial acceleration explained by velocity. While velocity influences tibial acceleration, individual variances to this relationship occur, with some runners demonstrating large increases or inconsistent changes, with increasing running velocity. These findings highlight the need for a personalised approach to understanding the response of each runner.

## **4.5 Chapter 4 Novel Contributions**

In this chapter I outline the average effects that running velocity has on TA-R, but also provide examples of the individual variances that can occur in this relationship. I recommend that it is important to understand these relationships, given that due to their baseline physiological characteristics and training history, runners will likely have a small range of velocities that they're comfortable running at. When working with runners I stress the importance of a personalised approach, and therefore when selecting a running velocity it should be a level that is comfortable for the given runner. The data presented in this chapter can also be used to classify runners as 'high-impact' and potentially at higher risk of injury, and therefore appropriate for a gait retraining intervention.



## **Part B:**

# **The Effectiveness of Real-Time Haptic Feedback for Gait Retraining in Runners.**

### **Overview**

Having provided the background, context and purpose of a gait retraining system for runners at risk of TFF (Sections 1.0.5-1.0.7), the purpose of Part B of this thesis was to assess the effectiveness of a real-time feedback system for gait retraining in runners. Before such a system could be realised, a number of factors needed to be reviewed from a theoretical perspective, and tested from a practical stand-point. As a first step in this sequence, the aim of Chapter 5 was to establish the evidence for the use of augmented feedback to alter lower extremity loading in runners at risk of TFF and to determine an optimal framework to move forward with for the implementation of real-time retraining, of which the effectiveness was assessed in Chapters 6 and 7.

## **Chapter 5:**

### **The effectiveness of augmented feedback to reduce lower extremity loading in runners at risk of tibial fatigue fracture. A systematic review.**

*This chapter is comprised of the following manuscript:*

Sheerin, K., Reid, D., Taylor, D., Besier, T. The effectiveness of augmented feedback to reduce lower extremity loading in runners at risk of tibial fatigue fracture. A systematic review. To be submitted to *Sports Medicine*.

## **5 Preface**

A number of literature reviews have provided summaries of the research evidence for the use of gait retraining as an intervention with runners [47–49,160,175]. Two reviews focussed on running injuries in general [47,49], while Neal et al. [49] analysed research specific to runners with patellofemoral pain. There have been no reviews specifically targeting the kinetic risk factors associated with the development of TFF in runners. The cumulative findings provide a positive foundation of support for the use of this strategy with injured runners. However, for it to be ultimately successful in the rehabilitation or prevention of injuries, the changes need to be persistent, and therefore need to be guided by the principles underlying motor learning. Previous reviews have presented only minimal evidence for the retention of adaptations, and therefore true motor learning [47,48,160].

This Chapter takes with form of a systematic literature review, which has two primary aims. Firstly, to provide an evaluation of the research evidence for the use of augmented feedback to alter lower extremity loading in runners at risk of TFF. Secondly, to provide research-based recommendations for the multitude of factors and variables that can be manipulated during the design and implementation augmented feedback intervention with runners. This systematic review therefore forms the theoretical basis for the intervention applied in Chapters 6 and 7.

## 5.0 Introduction

Running is a popular form of exercise, known to have a positive influence on fitness and health but it is associated with a high incidence of lower extremity overuse injuries [3,4]. Repetitive impact loading is thought to play an important role in the pathophysiology of many common running injuries, especially bony fatigue fractures (commonly termed stress fractures) [10,133,157,176]. Tibial fatigue fractures, which are common in runners [177,178], have been linked to a number of biomechanical risk factors such as elevated tibial acceleration, as well as average and instantaneous vertical force loading rates [11,35,36,177,179,180].

A reduction in training load is a common treatment or prevention approach for fatigue fractures in runners [7,22]. However, if abnormal biomechanical variables are important in fracture development, altering these variables by changing running technique may prevent abnormal bony stress or fracture reoccurrence [181]. Five literature reviews have previously summarised the evidence for altering running technique using gait retraining, with the cumulative findings that gait retraining can result in small to large effects on kinetic, kinematic and spatio-temporal variables [47–49,160,175]. It is also possible to target specific biomechanical variables related to injuries, at least for patello-femoral pain syndrome and chronic exertional compartment syndrome [47–49]. Despite these findings, there is minimal evidence of the retention of adaptations, and therefore true motor learning [47–49]. Even where there has been a focus on augmented feedback, the evidence for particular modalities or methods of feedback for runners is lacking [47].

Motor learning is the process of creating a lasting change in motor performance [50]. In the context of motor learning, feedback is the positive or negative response that informs the learner how well the task was completed. This can be from an internal source, such as proprioceptive feedback, or an external source, such as an instructor or coach [50]. Augmented feedback is a mechanism where information is provided via an external source to supplement internal feedback. The feedback can relate to the individual's performance towards a desired outcome, or instructions used to emphasise certain aspects of the movement [51]. An aim of augmentation in the sport or rehabilitation setting is to enhance complex motor learning by optimising feedback. Augmented feedback has been used with good effect in a range of rehabilitation settings, including participants with both normal [182,183] and compromised [184,185] neurological systems, as well as across a number of different sports [53,186,187].

There are number of complexities around augmented feedback and motor learning that cannot be ignored. Motor learning can be facilitated by the use of an internal focus of attention (focus on the

movements themselves) or with an external focus of attention (focus on the movement effect) [188], with the latter being more suitable for acquisition of complex motor skills like running [189,190]. An external focus of attention accelerates the learning process by facilitating movement automaticity [191] and enhances the production of effective and efficient movement patterns [190,192,193]. The direction of focus can be guided by a combination of the factors associated with the feedback variables selected and the instructions provided [50,190]. The threshold for feedback initiation is also an important parameter. According to the challenge point framework, the gap between an individual's current performance level and the target threshold is regarded as a 'challenge' to bridge [194]. Theoretically, novice or less skilled learners may not improve if the task level is perceived as too challenging [50,194]. There is also evidence from the wider literature to suggest providing learners with some control over a practice session and feedback frequency may enhance learning [190,195]. The advantages of self-controlled feedback are that it focuses on the aspect the learner wants to alter and potentially the promotion of deeper information processing and improved motivation [196]. Even with optimally designed feedback strategy, it is essential that participants are clear on what the desired outcomes of the intervention are. Instructions or cues can not only provide this information but they can also guide the strategies to achieve the desired outcomes [50].

There is currently no review targeting kinetic variables specifically associated with the development of tibial fatigue fractures. The primary aim of this review was to evaluate the evidence for the use of augmented feedback to alter lower extremity loading in runners at risk of tibial fatigue fractures. The secondary aim was to provide recommendations that guide the implementation of augmented feedback interventions.

## **5.1 Methods**

### **5.1.1 Search Parameters and Criteria**

The methods used for this systematic review follow the structure outlined in the guidelines given by the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) statement [197]. The Web of Science, Scopus and MEDLINE, CINAHL and SportDiscus (via EBSCO) databases, to 21 Nov 2017, were searched for the following terms linked with the Boolean operators ('AND', 'OR', 'NOT'): 'run\*', 'feedback', 'retraining', 'biomechanics', 'load\*', 'force\*', 'impact\*', 'acceleration' and 'shock'. The search was limited to English language and peer reviewed academic journals. Additional searches of the reference lists of identified papers and citing reference search using Google Scholar were undertaken. Papers were selected based on title, then abstract and finally text. Only papers that included straight-line, submaximal running,

and specifically measured kinetic variables related to the development of tibial fatigue fractures and incorporated an augmented feedback intervention designed to have a direct effect on lower extremity biomechanical characteristics were included in the data analysis. An augmented feedback intervention was defined where a runner made corrections to their running gait based on externally sourced information regarding their current performance relative to the desired performance for a targeted variable [50].

### **5.1.2 Quality Assessment**

The methodological quality of included articles was analysed using a modified Downs and Black checklist [198]. This checklist has been shown to have good validity and both test-retest and inter-rater reliability for rating the methodological quality of both randomised controlled trials (RCT) and non-RCTs [198,199]. This tool contains 27 items evaluating five subscales: reporting, external validity, bias (intervention and outcome measurement), confounding (cohort selection bias) and power. Items were scored 0 or 1, except for item 5, which scored 0 to 2. The power subscale (question 27) has previously been identified as ambiguous [200] and was therefore modified from a score out of five to a score out of one, with zero points awarded if no power calculation was attempted and one point awarded if a power calculation was described [201], giving a maximum score of 28. Two authors (KS and DR) independently reviewed and scored studies, and after comparing final scores, discussed and reached a consensus on any differences.

Based on quality assessment scores studies were categorised as ‘high quality and providing strong evidence’ ( $\geq 21$ ), ‘moderate quality and level of evidence’ (14–20), ‘low quality and limited evidence’ (7–13), or ‘poor quality and/or conflicting evidence’ ( $< 7$ ) [48,201]. Where possible, similar findings were combined in results tables for various retraining interventions, and the quality of these associated studies subsequently used to determine the level of evidence for each finding based on a modified version of the van Tulder et al. [202] criteria previously used to assess the determinants of lower extremity running injuries in runners (Table 6) [48].

1. Strong = consistent findings among multiple studies including at least three high-quality studies;
2. Moderate = consistent findings among multiple trials, including at least three moderate-quality/high-quality studies or two high-quality studies;
3. Limited consistent findings among multiple low-quality/ moderate-quality studies, or one high-quality study;
4. Very limited findings from one low-quality/moderate- quality study.

There are no consensus definitions on the appropriate timeframes for the retention of biomechanical changes for injury prevention. As such we have defined ‘immediate effects’ as those changes evident at the cessation of the intervention period without a washout period; ‘short-term effects’ as changes evident after a washout period of up to 4-weeks; ‘long-term effects’ as changes remaining after a period exceeding 1-month. These definitions are broadly in agreement with other researchers in the field [97,110,159,203,204], also satisfy the notion that the follow-up period should at least be sufficiently long for the outcome of interest (i.e. tibial fatigue fracture) to appear [205]. In the case of tibial fatigue fractures, this would typically be 1-2 months of running based loading [206].

**Table 6: Levels of evidence.**

<b>Strong evidence</b>	Consistent findings (in $>75\%$ of the studies) among multiple ( $>2$ ) high quality studies.
<b>Moderate evidence</b>	Consistent findings (in $>75\%$ of the studies) among one high quality study and multiple low quality studies.
<b>Limited evidence</b>	Consistent findings (in $>75\%$ of the studies) among multiple low quality studies or one high quality study.
<b>Conflicting evidence</b>	Provided by conflicting findings (fewer than 75% of the studies reported consistent findings).

### **5.1.3 Data Management**

Participant numbers and characteristics, intervention and follow-up details (Table 7), key clinical findings and biomechanical outcomes (Table 8) were extracted from all studies. Biomechanical variables of interest in this review included average vertical loading rate (AVLR), instantaneous vertical loading rate (IVLR), peak vertical ground reaction force (GRF), vertical impact peak (VIP) of the GRF and peak tibial acceleration (PTA). Where data were incomplete authors were contacted and given the opportunity to provide additional data [82,83,97,203,207,208]. To quantify the magnitude of change, mean difference as a percentage and effect sizes (ES) were calculated for the variables of interest [160]. Effect sizes were interpreted as trivial (0.0-0.1), small (0.11-0.3), moderate (0.31-0.5), large (0.51-0.7), very large (0.71-0.9), or extremely large (>0.91) [166].

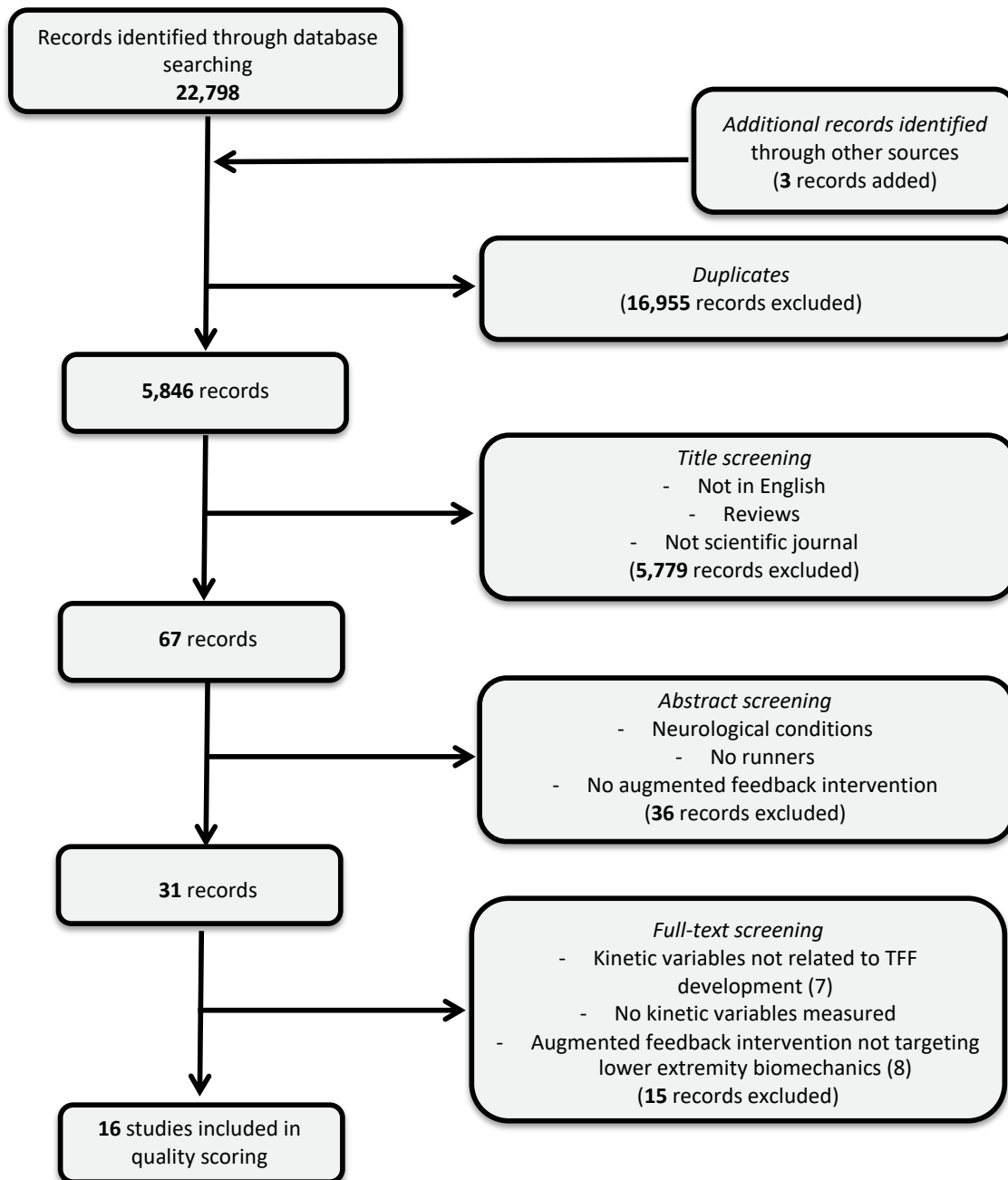
## **5.2 Results**

### **5.2.1 Search Results**

The electronic database search yielded a combined total of 821 citations. After the removal of duplicates and a review of titles, abstracts and full texts based on the inclusion and exclusion criteria, 16 studies were included in the review (Figure 8).

### **5.2.2 Quality Assessment**

From a maximum score of 28, the 16 studies reviewed ranged from 12 to 21 for the modified Downs and Black score (Table 7). One study was high quality [203], 12 were moderate quality [39,82,97,110,204,207–213] and three were low quality [83,159,214]. All studies clearly described their objectives, outcomes, subject characteristics, interventions and main findings, but none described the principle confounders. Two studies reported the adverse events experienced by participants [110,207], and all but two studies provided estimates of the random variability in the data [82,83] and actual probability values [83,159]. With reference to external validity, two studies scored a two out of three [203,209], with all remaining studies scoring a one or zero. Four studies randomised participants to the intervention or control groups [97,110,203,209] and one study blinded participants [203] or assessors [97] to this process. Further detail on the quality scoring criteria and assessment are outlined in Appendix 16.



**Figure 8: Flow diagram of systematic search**



**Table 7: Participant and study characteristics for Chapter 5**

Reference	Quality Score	Participant details Age (years)   Height (cm)   Weight (kg)	Surface / running speed	Augmented FB type	Intervention protocol and follow-up period	Instructions and targeted gait parameter to change	Supplementary intervention
Willy [203]	21	7M & 9F (Int) and 9M & 5F (Con) uninjured recreational runners with elevated IVLR (>85 BWs <sup>-1</sup> ) Int: 21.0 ±1.4   NR   NR Con: 20.7 ±1.3   NR   NR	Treadmill 3.3 m/s	Self-controlled visual FB of cadence on wrist watch. Faded FB: Cadence FB during runs 1–3, 5, and 7 only	Int: 8 sessions of unreported duration (normal training duration). Con: Normal training with pace and distance FB, but not cadence. FU at 4 weeks	Match cadence to the target (7.5% >preferred)	
Chan [207]	19	82M & 84F (Int.) 76M & 78F (Con.) uninjured runners with ALVR >70 BWs <sup>-1</sup> Int: 33.6 ±9.5   166.0 ±1.0   60.0 ±12.6 Con: 34.2 ±9.5   165 ±1.0   61.6 ±12.0	Treadmill Self-selected	Continuous visual FB of the vertical GRF from instrumented treadmill displayed on monitor	8 sessions over 2 weeks. Running period increased from 15-30 min Visual FB progressively withdrawn over final 4 sessions Reassessment immediately post-intervention	Run softer so that VIP is reduced or diminished	Maintain new gait pattern during regular running after the training
Samaan [209]	17	23M & 26F recreational runners with chronic lower extremity injury (all with VIP in GRF trace) 33.7   172.0   66.2	Treadmill Self-selected	Vertical GRF trace displayed on a screen Participants were also required to run barefoot during intervention	Single session of undefined time period	Make the vertical force trace as smooth as possible. Land on the balls of feet. Land as softly as possible.	
Noehren [204]	17	10F recreational runners with PFPS (qualitative video assessment of excessive hip adduction) 23.3 ±5.8   167.0 ±0.05   57.4 ±4.1	Treadmill 3.35 m/s	Visual FB of stance phase hip adduction angle on screen. Continuous FB for initial 4 sessions. Progressive withdrawal over final 4 sessions.	8 sessions over 2 weeks. Running period increased from 15-30 min. FU at 4 weeks	Contract gluteal muscles and attempt to run with knees straight ahead, maintaining a level pelvis, and keeping hip (adduction) angle within target range (±1 SD of normal)	

**Table 7: Continued**

Reference	Quality Score	Participant details Age (years)   Height (cm)   Weight (kg)	Surface / running speed	Augmented FB type	Intervention protocol and follow-up period	Instructions and targeted gait parameter to change	Supplementary intervention
Chen [212]	16	7M & 7F uninjured RFS runners 35.3 ±5.9   170.0 ±0.08   64.0 ±11.2	Treadmill 2.5 m/s	Continuous visual FB of foot landing pattern on screen.	Single session 15 min (5 min baseline; 10 min intervention)	Modify foot strike pattern from RFS to MFS and then FFS	A digital metronome used to stabilise to natural running cadence.
Crowell [82]	16	4M & 6F recreational RFS runners with PTA >8 g 26 ±2.0   172.0 ±0.07   81.5 ±21.0	Treadmill 3.7 m/s	Visual FB of PTA via a screen. Target line indicating >50% of baseline. Continuous FB during initial 4 sessions. Progressive withdrawal over final 4 sessions	8 sessions over 2 weeks. Running period increased from 15-30 min. FU at 4 weeks	Run softer and aim to reduce PTA by 50%	
Wood [39]	19	3M & 6F uninjured recreational runners 20 ±1.5   170.2 ±8.7   59.1 ±8.2	Treadmill Self-selected (3.13 ±2.5 m/s)	Audio FB of PTA when PTA exceeded threshold set 10-15% below baseline PTA. Audio pitch indicated position relative to threshold PTA	Single session of 25 min running	Run without any beeps and keep the pitch of the beeps (PTA) as low as possible	
Tate [210]	15	4M & 10F uninjured runners 23.7 ±2.0   167.0 ±0.8   60.9 ±8.7	Treadmill Self-selected	Visual FB of sound intensity from tablet on treadmill console	Single session of 15 min running	Decrease decibel level by trying to run as quietly as possible	
Baggaley [208]	15	16M & 16F uninjured RFS runners 24.7 ±3.3   NR   NR	Treadmill   Self-selected (mean: 2.9 ±0.3 m/s)	Continuous visual FB - current position and target range identified on screen	Single session of undefined time period	No instructions outlined AVLR	

**Table 7: Continued**

Reference	Quality Score	Participant details Age (years)   Height (cm)   Weight (kg)	Surface / running speed	Augmented FB type	Intervention protocol and follow-up period	Instructions and targeted gait parameter to change	Supplementary intervention
Diebal [213]	15	8M & 2F military personnel (all RFS) with chronic exertional compartment syndrome 20.2 ±1.5   176.6 ±5.1   88.5 ±15.9	Treadmill Self-selected	Post-run video FB of run technique and FFS	18 sessions over 6 weeks. 20 min of running (intervals for distances of 0.25 km with a 2-min walking breaks). The final 3 weeks running speed and endurance (distance) progressed	Run quietly and eliminate heel strike Use hamstrings to pull foot from the ground Increase cadence to 3 steps/s	Specific training drills and exercises designed to teach FFS - based on Pose running technique A digital metronome used to stabilise cadence at 180 steps/min
Clansey [110]	15	10M (Int.) & 12M (Con.) healthy RFS runners with elevated PTA (>9 g) Int.: 33.3 ±9.0   180.0 ±0.1   77.2 ±11 Con.: 33.9 ±11.3   180.0 ±0.1   75.1 ±6.9	Treadmill 3.7 m/s	Simultaneous visual and auditory FB: Visual (traffic symbol) on screen and auditory (pitch) indicating acceptable (<50% of baseline), medium (50-75% of baseline) and high (>75% of baseline) PTA	6 sessions over 3 weeks. 20 min running per session Con = identical training without FB FU at 4 weeks	Find a strategy to run within the acceptable (PTA) shock range	
Creaby [97]	15	11M (clinician) & 11M (visual FB) uninjured runners Clinician guided: 28.1 ±7.8   178.0 ±0.05   76.5 ±7.7 Visual FB: 22.7 ±4.6   179.0 ±0.05   78.8 ±9.2	Treadmill 3.0 m/s	Verbal clinician FB: Cues repeated each min Continuous visual FB of PTA via a screen. Target line indicating >50% of baseline	Single session of 25 min running (5 min without FB, 10 min with FB, and 10 min without FB). FU at 1 week	Clinician FB: Cues to 'run softer' and 'make footfalls quieter' Run softer and aim to reduce PTA by 50%	
Cheung [211]	14	11M & 5F uninjured recreational runners with elevated FA (>10g)	Treadmill Self-selected	Continuous visual (green /red symbol) FB on screen indicating acceptable (<80% of baseline) or high (>80% of baseline) FA. FB withdrawn over final 4 sessions.	8 sessions over 2 weeks. Running period increased from 15-30 min. FU at 1 week	Land softer	A distracting verbal counting task while running during the assessments

**Table 7: Continued**

Reference	Quality Score	Participant details Age (years)   Height (cm)   Weight (kg)	Surface / running speed	Augmented FB type	Intervention protocol and follow-up period	Instructions and targeted gait parameter to change	Supplementary intervention
Phan [214]	13	26M uninjured runners 21.1 ±2.0   179.0 ±0.5   78.3 ±12.2	Barefoot overground running 5.0 ±0.5 m/s	Impact sound of runners' foot captured on microphone	Single session with 5 running bouts	Perform the task as previously but reduce sound when you land	
Crowell [83]	12	5F uninjured recreational runners 26.0 ±2.0   164.0 ±0.06   59.3 ±5.4	Treadmill Self-selected (2.4-2.6 m/s)	Continuous visual FB of PTA via a screen. Target line indicating >50% of baseline.	Single session of 20 min running	Run softer and aim to reduce PTA by 50%	
Cheung [159]	12	3F recreational runners with PFPS 26 - 32   162 - 170   45.0 - 51.2	Treadmill 2.8 m/s	Auditory FB sensor in heel of insole. Continuous FB for initial 4 sessions. FB progressively withdrawn over final 4 sessions.	8 sessions over 2 weeks. Running period increased from 15-30 min. FU at 12 weeks	Eliminate the buzzer noise by shortening stride length and avoid a RFS landing	

AVLR = average vertical loading rate;  $BWs^{-1}$  = body weight per seconds; Con = control F = female; FA = foot acceleration; FB = feedback; FU = follow-up; FFS = forefoot strike; GRF = ground reaction force; ILVR = instantaneous vertical loading rate; Int = intervention; M = male; m/s = meters per second; min = minute; MFS = mid-foot strike; NR = not reported; PFPS = patella-femoral pain syndrome; PTA = peak tibial acceleration; RFS = rearfoot strike; SD = standard deviation; VIP = vertical impact peak

### **5.2.3 Study Characteristics**

Twelve studies were uncontrolled or observational in their design [39,82,83,159,204,208–214]. Three studies included control groups, where participants underwent identical assessments and training regimes, without the specific intervention feedback [110,203,207], while one study randomly allocated participants to a ‘typical treatment’ comparison group, but did not include a ‘no-intervention’ control group [97] (Table 7).

### **5.2.4 Participant Characteristics**

A total of 591 participants were included across the 16 studies (411 cases, 180 controls). Males accounted for 54% of the participants. A key criterion for study inclusion was that kinetic variables related to the development of tibial fatigue fractures were measured during running. While all satisfied this criteria, several studies in the current review included participants diagnosed with other running related injuries such as patella-femoral pain syndrome [159,204], chronic exertional compartment syndrome [213] and non-specific chronic lower extremity injury [209]. The remaining studies included either uninjured runners with no pre-defined injury risk factors [39,83,97,207,210,214], runners demonstrating a rearfoot strike [208,212], or runners with an elevated PTA [82,110,211] or IVLR [159,203] (Table 7).

### **5.2.5 Evidence for Kinetic Biomechanical Effects**

#### *5.2.5.1 Immediate Effects*

Evidence across multiple studies evaluating augmented feedback gait retraining provide moderate evidence for the following immediate effects: Moderate to extremely large reductions in VIP (ES 0.30 to 1.33) [82,204,210,213]; moderate to extremely large reductions in tibial and foot acceleration (ES 0.49 to 1.97) [39,82,97,110,211]; large to extremely large reductions in IVLR (ES 0.68 to 4.26) [82,83,110,159,203,204,207,209–212]; moderate to extremely large reductions in AVLRL (ES 0.47 to 4.15) [82,83,110,159,203,204,207–214].

#### *5.2.5.2 Short- and Long-term Effects*

Evidence from multiple studies provided moderate evidence to support the following short-term effects: Moderate to extremely large reductions in tibial and foot acceleration (ES 0.37 to 1.8) [82,97,110,211]; moderate to extremely large reductions in IVLR (ES 0.35 to 1.58) [82,110,203,207]; small to extremely large short-term reductions in AVLRL (ES 0.27 to 1.51); [82,110,203,207]. Only limited evidence was found to support very large to extremely large short-term reductions in VIP (ES 0.72 to 1.08) [82].

Very limited evidence from a single study supported extremely large long-term reductions in AVLr (ES 1.66) and IVLR (ES 2.13) [159]. While Noehren, et al. [204] conducted a 4-week retention test, the results for kinetic variables were not reported. Only one of the reviewed studies carried out a long-term retention test (3-months post-intervention), however the low subject numbers (n=3) and low quality score (12), do not provide evidence for any long-term effects [159]. Additional information regarding the effects of augmented feedback gait retraining on kinetic biomechanical variables are provided in Table 8.

#### **5.2.6 Feedback Modality**

A range of feedback modalities and combinations of modalities were used (Table 7). The most common feedback modality was visual, used in 12 studies [82,83,97,203,204,207–213]. Four studies used an audio feedback modality [39,97,159,214]. Creaby et al. [97] were the only researchers to compare feedback modalities, whereby one group received feedback via a visual display, while a second group received verbal feedback from a clinician. Clansey et al. [110] were the only researchers to use multi-modal feedback incorporating a simultaneous audio-visual system.

#### **5.2.7 Intervention Length**

Seven studies provide evidence that two- [82,159,203,204,207,211], three- [110] or six-week [213] intervention periods effected changes in kinetic variables as a result of augmented feedback facilitated retraining. Assessments conducted at the cessation of feedback provide a measure for the immediate effect of the intervention.

**Table 8: Key clinical findings and biomechanical outcomes**

Reference	Clinical findings	Biomechanical kinetic results	Mean Difference	Effect Size
Willy [203]	Runners demonstrated significantly increased cadence, and reduced AVLRL, IVLRL, peak hip adduction, and eccentric knee joint work immediately after the intervention. Gait modifications were maintained at 4 weeks.	<p>↓ AVLRL post FB: 75.6 ±7.79 vs. 61.3 ±13.78 BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ AVLRL 4 weeks post FB: 75.6 ±7.79 vs. 60.4 ±13.19 BWs<sup>-1</sup> *(p&lt;0.0001)</p> <p>↓ IVLRL post FB: 101.3±7.79 vs. 83.2 ±20.91 BWs<sup>-1</sup> *(p=0.0001)</p> <p>↓ IVLRL 4 weeks post: 101.3 ±7.79 vs. 83.2 ±19.46 BWs<sup>-1</sup> *(p=0.001)</p>	<p>19%</p> <p>20%</p> <p>18%</p> <p>17%</p>	<p>1.33</p> <p>1.44</p> <p>1.18</p> <p>1.21</p>
Chan [207]	At 12-month follow-up, the occurrence of running-related injury was 16% and 38% in the gait retraining and control groups respectively. Hazard ratio between gait retraining and control groups was 0.38 (95%C.I.=0.25-0.59), indicating a 62% lower injury occurrence in gait retrained runners, compared with controls.	<p>↓ AVLRL post FB (2.2 m/s): 65.95 ±9.90 vs. 54.82 ±11.04 * BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ AVLRL post FB (3.3 m/s): 81.28 ±13.59 vs. 66.65 ±12.53 BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ IVLRL post FB (2.2 m/s): 90.69 ±13.90 vs. 5.02 ±17.09 BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ IVLRL post FB (3.3 m/s): 111.87 ±14.51 vs. 4.75 ±19.61 BWs<sup>-1</sup> *(p&lt;0.001)</p>	<p>17%</p> <p>18%</p> <p>17%</p> <p>15%</p>	<p>1.06</p> <p>1.12</p> <p>1.01</p> <p>1.00</p>
Samaan [209]	A combined barefoot running and visual FB intervention resulted in significant reductions in AVLRL, IVLRL, vertical stiffness, peak medial and lateral GRFs, and vertical, medial and lateral force impulses.	<p>↓ AVLRL post FB: 71.60 ±26.60 vs. 26.70 ±7.80 BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ IVLRL post FB: 83.90 ±29.00 vs. 36.50 ±9.20 BWs<sup>-1</sup> *(p&lt;0.001)</p>	<p>62%</p> <p>57%</p>	<p>2.61</p> <p>2.48</p>
Noehren [204]	Following 2 weeks of FB retraining, runners with PFPS demonstrated reductions in AVLRL, IVLRL, hip adduction angle, and contralateral pelvic drop and improvements in pain. Runners were able to maintain their biomechanical and pain improvements at 4 weeks.	<p>↓ VIP post FB: 1.43 ±0.20 vs. 1.28 ±0.10 BW</p> <p>↓ AVLRL post FB: 55.0 ±13.00 vs. 44.0 ±7.0 BWs<sup>-1</sup></p> <p>↓ IVLRL post FB: 65.0 ±15.0 vs. 63.0 ±7.1 BWs<sup>-1</sup></p>	<p>10%</p> <p>20%</p> <p>18%</p>	<p>1.00</p> <p>1.10</p> <p>1.09</p>
Chen [212]	vGRF loading rates were reduced when the runners landed with MFS or FFS, compared to RFS. Changes to foot strike did not lower peak tibial strains.	<p>↓ AVLRL with MFS: 65.65 ±7.94 vs. 40.06 ±9.10 BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ AVLRL with FFS: 65.65 ±7.94 vs. 35.00 ±6.84 BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ IVLRL with MFS: 91.16 ±10.21 vs. 60.75 ±12.08 BWs<sup>-1</sup> *(p&lt;0.001)</p> <p>↓ IVLRL with FFS: 91.16 ±10.21 vs. 51.07 ±8.60 BWs<sup>-1</sup> *(p&lt;0.001)</p>	<p>39%</p> <p>47%</p> <p>33%</p> <p>40%</p>	<p>3.00</p> <p>4.15</p> <p>2.73</p> <p>4.26</p>
Crowell [82]	Reductions in PTA, VIP, AVLRL and IVLRL were demonstrated immediately following, and 4 weeks following, visual FB training in runners with elevated baseline PTA.	<p>↓ VIP post FB: 1.67 ±0.26 vs. 1.37 ±0.27 BW *(p&lt;0.05)</p> <p>↓ VIP 1 month post FB: 1.67 ±0.26 vs. 1.35 ±0.33 BW *(p&lt;0.05)</p> <p>↓ AVLRL post FB: 75.2 ±15.7 vs. 50.2 ±10.3 BWs<sup>-1</sup> *(p&lt;0.05)</p> <p>↓ AVLRL 1 month post FB: 75.2 ±15.7 vs. 55.1 ±10.8 BWs<sup>-1</sup> *(p&lt;0.05)</p> <p>↓ IVLRL post FB: 90.4 ±18.5 vs. 59.2 ±10.8 BWs<sup>-1</sup> *(p&lt;0.05)</p> <p>↓ IVLRL 1 month post FB: 90.4 ±18.5 vs. 63.1 ±15.97 BWs<sup>-1</sup> *(p&lt;0.05)</p> <p>↓ PTA post FB: 8.2 ±2.5 vs. 4.3 ±1.5 g *(p&lt;0.05)</p> <p>↓ PTA 1 month post FB: 8.2 ±2.5 vs. 4.6 ±1.5 g *(p&lt;0.05)</p>	<p>18%</p> <p>19%</p> <p>33%</p> <p>27%</p> <p>34%</p> <p>30%</p> <p>48%</p> <p>44%</p>	<p>1.33</p> <p>1.08</p> <p>1.92</p> <p>1.51</p> <p>2.12</p> <p>1.58</p> <p>1.95</p> <p>1.8</p>

Table 8: Continued

Reference	Clinical findings	Biomechanical kinetic results	Mean Difference	Effect Size
Wood [39]	Runners were able to reduce their PTA after audio FB intervention.	↓ PTA post FB 1: $5.90 \pm 0.70$ vs. $5.30 \pm 0.80$ g; $^*(p=0.015)$ No significant difference in PTA post no FB 1: $5.90 \pm 0.70$ vs. $5.60 \pm 1.10$ g ( $p>0.05$ ) ↓ PTA post FB 2: $5.90 \pm 0.70$ vs. $5.20 \pm 0.60$ g $^*(p=0.021)$ ↓ PTA post no 2: $5.90 \pm 0.70$ vs. $5.40 \pm 0.70$ g $^*(p=0.033)$	10% 5% 12% 8%	0.80 0.33 1.08 0.71
Tate [210]	Runners were able to immediately reduce loading rates and impact forces after a sound-intensity intervention. Transfer of effects to overground running was also demonstrated.	↓ VIP post FB: $1.56 \pm 0.31$ vs. $1.13 \pm 0.34$ BW $^*(p<0.001)$ ↓ AVLR post FB: $69.09 \pm 20.15$ vs. $43.91 \pm 16.14$ BWs <sup>-1</sup> $^*(p<0.001)$ ↓ IVLR post FB: $95.48 \pm 27.41$ vs. $62.79 \pm 22.35$ BWs <sup>-1</sup> $^*(p=0.001)$	28% 26% 35%	1.33 1.39 1.31
Baggaley [208]	All interventions were effective at reducing AVLR. FFS was the most effective, but resulted in a shorter step length (which the other interventions did not).	↓ AVLR post step rate FB training: $56.59 \pm 19.81$ vs. $48.18 \pm 16.12$ BWs <sup>-1</sup> $^*(p=0.004)$ ↓ AVLR post AVLR FB training: $56.59 \pm 19.81$ vs. $41.67 \pm 13.9$ BWs <sup>-1</sup> $^*(p<0.001)$ ↓ AVLR post FFS FB training: $56.59 \pm 19.81$ vs. $30.17 \pm 7.83$ BWs <sup>-1</sup> $^*(p<0.001)$	15% 26% 47%	0.47 0.89 1.91
Diebal [213]	FFS run training resulted in reductions in VIP, AVLR, post-run anterior compartment pressures and pain. 1-year post-intervention run times were faster than pre-intervention, and no runners required surgery.	↓ VIP post FB: Mean difference = $2.40 \pm 0.18$ vs. $2.34 \pm 0.22$ BW $^*(p<0.05)$ ↓ AVLR post FB: Mean difference = $29.66 \pm 5.54$ vs. $26.43 \pm 5.71$ BWs <sup>-1</sup> $^*(p<0.05)$	3% 10%	0.30 0.57
Clansey [110]	The retraining group demonstrated significant reductions in PTA, AVLR and IVLR, with no changes in the control group. Only PTA changes remained at 4 weeks.	No significant difference in VIP post FB: $2.72 \pm 0.17$ vs. $2.62 \pm 0.16$ BW ( $p=0.228$ ) No significant difference in VIP at 4 weeks: $2.72 \pm 0.17$ vs. $2.59 \pm 0.19$ BW ( $p=0.228$ ) ↓ AVLR post FB: $66.54 \pm 17.45$ vs. $54.62 \pm 15.41$ BWs <sup>-1</sup> $^*(p=0.005)$ ↓ AVLR at 4 weeks: $66.54 \pm 17.45$ vs. $61.38 \pm 20.15$ BWs <sup>-1</sup> $^*(p=0.005)$ ↓ IVLR post FB: $113.87 \pm 33.01$ vs. $92.10 \pm 27.06$ BWs <sup>-1</sup> $^*(p=0.001)$ ↓ IVLR at 4 weeks: $113.87 \pm 33.01$ vs. $101.63 \pm 26.73$ BWs <sup>-1</sup> $^*(p=0.001)$ ↓ PTA post FB: $10.67 \pm 1.85$ vs. $7.39 \pm 1.48$ g ( $p<0.001$ ) ↓ PTA at 4 weeks: $10.67 \pm 1.85$ vs. $8.30 \pm 1.82$ g ( $p<0.001$ )	4% 5% 18% 8% 19% 10% 31% 22%	0.62 0.72 0.75 0.27 0.72 0.35 1.97 1.29
Creaby [97]	Both clinician and PTA FB were effective in reducing PTA immediately following FB and after 1-week. There was no difference between the two FB interventions in the reduction in PTA at FU.	↓ PTA during clinical FB: $5.74 \pm 2.25$ vs. $4.37 \pm 1.86$ g $^*(p=0.006)$ ↓ PTA post clinical FB: $5.74 \pm 2.25$ vs. $4.13 \pm 1.82$ g $^*(p=0.008)$ ↓ PTA 7-days post clinical FB: $5.74 \pm 2.25$ vs. $4.48 \pm 1.53$ g $^*(p=0.029)$ ↓ PTA during accelerometer FB: $5.34 \pm 1.93$ vs. $3.81 \pm 1.36$ g $^*(p=0.006)$ ↓ PTA post accelerometer FB: $5.34 \pm 1.93$ vs. $4.32 \pm 2.20$ g $^*(p=0.008)$ ↓ PTA 7-days post accelerometer FB: $5.34 \pm 1.93$ vs. $4.20 \pm 1.54$ g ( $p=0.029$ )	23% 28% 22% 28% 20% 21%	0.67 0.79 0.67 0.93 0.49 0.66



**Table 8: Continued**

Reference	Clinical findings	Biomechanical kinetic results	Mean Difference	Effect Size
Cheung [211]	FA, AVLRL and IVLR during running were significantly reduced after retraining, even under distraction.	↓ AVLRL post FB: 45.01 ±11.48 vs. 37.26 ±11.28 BWs <sup>-1</sup> *(p<0.001)	17%	0.69
		↓ AVLRL 2 weeks post FB: 34.02 ±11.18 vs. 30.03 ±11.17 BWs <sup>-1</sup> *(p=0.002)	12%	0.41
		↓ IVLR post FB: 57.8 ±7.74 vs. 50.29 ±10.04 BWs <sup>-1</sup> *(p<0.001)	13%	0.68
		↓ IVLR 2 weeks post FB: 47.43 ±9.25 vs. 43.82 ±10.14 BWs <sup>-1</sup> *(p=0.001)	8%	0.36
		↓ FA post FB: 16.05 ±7.03 vs. 11.92 ±4.9 g *(p=0.001)	26%	0.84
		↓ FA post 2 weeks FB: 9.46 ±4.03 vs. 8.07 ±2.73 g *(p=0.001)	15%	0.37
Phan [214]	A quieter impact sound was not directly associated with lower peak vGRF or force loading rates. However, runners were able to reduce their peak vGRF, AVLRL and sound amplitude when instructed to run quietly.	↓ Peak vGRF post FB: 2.7 ±0.38 vs. 2.53 ±0.28 BW; *(p=0.001)	6%	0.52
		↓ AVLRL post FB: 390.17 ±214.14 vs. 115.04 ±125.89 BWs <sup>-1</sup> ; *(p<0.015)	71%	1.62
Crowell [83]	All runners demonstrated reductions in VIP, AVLRL and IVLR, and 4/5 subjects showed reductions in their PTA post FB.	↓ VIP post FB: 1.28 ±0.05 vs. 1.11 ±0.04 BW	13%	1.46
		↓ VIP 10 min post FB: 1.28 ±0.05 vs. 1.07 ±0.05 BW	16%	1.72
		↓ AVLRL post FB: 53.8 ±10.9 vs. 43.6 ±10.3 BWs <sup>-1</sup>	19%	0.97
		↓ AVLRL 10 min post FB: 53.8 ±10.9 vs. 39.1 ±9.7 BWs <sup>-1</sup>	27%	1.43
		↓ IVLR post FB: 60.4 ±12.0 vs. 48.6 ±11.3 BWs <sup>-1</sup>	20%	1.02
		↓ IVLR 10 min post FB: 60.4 ±12.0 vs. 43.7 ±10.9 BWs <sup>-1</sup>	28%	1.46
		↓ PTA post FB: 9.0 ±1.6 vs. 7.2 ±4.9 g	20%	0.60
		↓ PTA 10 min post: 9.0 ±1.6 vs. 7.2 ±4.9 g	30%	0.79
Cheung [159]	Runners successfully changed from a RFS to a MFS as a result of FB, and maintained changes for 3 months. VIP and loading rates were reduced, and PFPS and functional limitations improved.	No difference in VIP post FB: 1.57 ±0.13 vs. 1.33 ±0.10 BW	15%	2.00
		No difference in VIP at 12 weeks: 1.57 ±0.13 vs. 1.33 ±0.17 BW	15%	1.56
		↓ AVLRL post FB: 48.47 ±7.67 vs. 37.20 ±8.20 BWs <sup>-1</sup>	23%	1.42
		↓ AVLRL at 12 weeks: 48.47 ±7.67 vs. 35.53 ±7.93 BWs <sup>-1</sup>	27%	1.66
		↓ IVLR post FB: 67.57 ±7.67 vs. 51.57 ±8.03 BWs <sup>-1</sup>	24%	2.04
		↓ IVLR at 12 weeks: 67.57 ±7.67 vs. 50.70 ±8.20 BWs <sup>-1</sup>	25%	2.13

AVLRL = average vertical loading rate; BWs<sup>-1</sup> = body weight per seconds; FA = foot acceleration; FB = feedback; FFS = forefoot strike; GRF = ground reaction force; ILVR = instantaneous vertical loading rate; m/s = meters per second; MFS = mid-foot strike; PFPS = patella-femoral pain syndrome; PTA = peak tibial acceleration; RFS = rearfoot strike; VIP = vertical impact peak; vGRF = vertical ground reaction force

## **5.3 Discussion**

This systematic review synthesises the current evidence for the use of augmented feedback to facilitate running retraining to reduce biomechanical risk factors for tibial fatigue fracture occurrence. The initial parts of the discussion outline the evidence for augmented feedback to facilitate changes to running kinetic variables, while the later portions review motor learning principles important for the success of practical interventions.

### **5.3.1 Study Characteristics**

While the focus of this review was to summarise the evidence for the effectiveness of augmented feedback in runners with TFFs, there were no studies that specifically recruited participants that fitted this criteria. As such, studies that addressed the kinetic risk factors related to TFFs were selected. As such, studies with healthy participants and other disorders (such as patellofemoral pain syndrome) were also included. Only a small number of studies included control [110,207], or typical treatment comparison groups [97]. This is likely due to the difficulty in creating a true control for many interventions in this field. For example, it is not possible to have a sham feedback intervention [215]. Experimental approaches, such as single-subject A-B-A interventions, or designs where each subject trial are considered independent, present potential alternatives without the need for control groups [56,216,217]. However, in the field of motor learning, where often the goal is sustained changes, such approaches are challenging [215,218].

While group effects are typically closely reported, clinicians and coaches are interested in individual changes [215,218]. As Crowell et al. [83] highlight, even within a small cohort of runners, there can be non-responders. While it is speculated that this might be due to different rates of learning, the authors provide no further analysis of individuals in their subsequent and larger intervention study [82]. To provide a robust appraisal of the effectiveness of gait retraining studies, randomly allocated control groups should be used where possible [219], but designs to facilitate the analysis of individuals could be more appropriate [56,216,217]. It should also be noted that methodological and statistical requirements for the valid assessment of interactive effects, including subject-by-training interaction, may differ from those for main effects [215,218].

### **5.3.2 Participant Characteristics**

The selection of appropriate participants is important for the success of any intervention [203]. Despite this point, five studies recruited uninjured runners for their intervention, with no initial screening of biomechanical injury risk factors [39,83,97,210,214]. If these runners exhibited typical readings for these variables, the likelihood of meaningful changes as a result of the intervention would be minimal.

A small number of researchers were selective with their recruitment, targeting runners diagnosed with running related injuries [159,204]. While these are obviously targeted to specific injuries, they may provide strategies to use as injury prevention for other runners. Specific quantitative criteria have been used to pre-screen runners for injury, including elevated AVLR [207], IVLR [203] and tibial or foot acceleration [39,110,159]. While these metrics may be accurate and reliable, some require the use of a force plate, presenting considerable challenge for clinicians without access to this equipment. Other researchers have used video assessment to qualitatively categorise runners with excessive hip adduction [204] or a rearfoot strike pattern [208,212]. This is a more clinically accessible method, however in the case of foot strike pattern, there are no known links to running injuries [220–222], so the value of this as a selection method is questionable.

### **5.3.3 Evidence for Kinetic Biomechanical Effects**

Based on study quality, changes scores and ESs, there is limited to moderate strength evidence to support moderate to extremely large immediate reductions in kinetic variables related to tibial fatigue fractures [82,97,110,204,207,211,213]. There is also limited evidence supporting very large to extremely large short-term reductions in these variables [82,97,110]. While there is no doubt these changes are statistically substantive, it is difficult to know what these relate to in terms of clinically meaningful findings. While immediate effects are important for proving the initial worthiness of an intervention, for the purposes of injury prevention, they will only be worthwhile implementing if they translate to changes that persist.

### **5.3.4 Feedback Modality**

#### *5.3.4.1 Visual Feedback*

Visual feedback has been the most frequently investigated feedback modality in motor learning [50]. Many possibilities exist for visualising kinematic or kinetic variables, ranging from abstract simple plots or numbers, to more realistic virtual environments. The specific design of a visual feedback may have a significant impact on the outcome [50].

Diebal, et al. [213] used a basic visual feedback in the form of a post-run video of running technique and foot strike pattern. While relatively low cost and accessible from a range of devices, the feedback is delayed until after task completion and therefore cannot always be associated with the internal sensory information at the time of motor execution [223]. A live feed of information, whether video [212] or numerically-based data [203] is relatively easy to set up, as it may require no additional processing. However, the ‘processing’ demand is placed on the runner to interpret meaning, from what could be quite fast-moving images or data. Running related kinetic variables often considered in gait analysis lend themselves to abstract

representation [50], where only the key features are presented [82,83,97,110,204,207–210]. The addition of target threshold information has the benefit of informing the learner in which direction they need to change and how far they are from the targeted zone [82,83,97,204,208]. In some cases, over simplifying the data (i.e. a traffic light system) can actually remove some of this useful information [110].

Historically, an inherent downside of most visual feedback was the requirement for a fixed screen, thereby limiting retraining to a lab or clinic environment. Willy et al. [203] made use of advancing technology, displaying numerical cadence data on a wrist watch while runners trained in their normal environment. Not only is this intervention more ecologically valid, it also has the potential to counter the boredom created in static environments [50]. There is moderate evidence that visual feedback can be successfully used with runners, at least short-term [82,110,203,204,211], however comparisons between feedback designs have not been reported and therefore specific conclusions on how visualisations should be designed cannot be drawn.

#### *5.3.4.2 Auditory Feedback*

There are a number of different forms of auditory feedback, which are typically categorised as either an auditory alarm or sonification. Natural audio feedback is where natural sounds, without any manipulation, are amplified and fed back. An ‘alarm’ is defined as a unmodulated sound, played as long as a related movement variable exceeds a predefined threshold. Whereas, sonification is where non-speech audio changes in magnitude to represent changes in the targeted variables [50].

Phan, et al. [214] adopted a basic sonification approach by amplifying the sound of runners’ feet, and asking them to make quieter landing sounds. This approach was successful in reducing loading variables after 6 weeks of feedback (ES: VIP 0.30; AVLr 0.57). Cheung et al. [159] used a warning beep linked to a force transducer on the heel of runners. To promote a forefoot strike pattern, a buzzer sounded with sensor-ground contact. These researchers reported positive results, with runners able maintain reductions in all kinetic variables for three months, although this study was of low quality and therefore the evidence is questionable. One method for improving auditory feedback could be to make runners aware of their deviation with respect to a reference point, rather than just alerting them to an error via an alarm [50]. Using a ‘sonified alarm’, Wood et al. [39] presented runners with a signal when impact accelerations exceeded a threshold. This was supplemented by scaling the pitch of the beep to the acceleration magnitude, such that the frequency of the beep would increase with larger accelerations. While this appears to be effective at the motor performance level, no follow-up was used to assess motor learning.

In the same way that visual display design can impact on learning, the success of auditory feedback interventions can depend on the accurate interpretation of sound functions used. While the 'foot impact' feedback provided by Phan, et al. [214] was novel and effective, it would be difficult to implement in the real-world environment where other ambient sounds also attract runners attention. Whereas alarms are simple to interpret, and if the runners can appropriately interpret the required technique correction, they appear an adequate form of feedback for motor learning in running. The limitation of standard alarm systems to represent the extent of the required movement changes has been previously identified. Given the limited evidence with runners, it is still unclear whether appropriately designed auditory feedback is as effective as visual feedback, or whether it can provoke a better linkage to kinaesthetic information.

#### *5.3.4.3 Haptic Feedback*

Tactile and kinesthetic perception conveyed through vibrations or pressure on the skin, or via receptors in muscles and tendons, that allow us to feel the position of our body is known as haptic feedback [50]. Vibration units have been used with good effect in a range of sports [54,224], however evidence for motor learning is limited to two rowing studies [53,54]. Haptic feedback has been used more widely in walking gait retraining to reduce knee adductor moments in healthy individuals and those with osteoarthritis [60,225]. Haptic feedback has also been tuned to provide independent feedback about multiple gait parameters within a single session [52,55]. No studies have used a haptic feedback modality in running gait retraining.

Like auditory alarms, haptic feedback is typically initiated based on the parameter error signals, meaning that learners receive intermittent, rather than continuous feedback, which could reduce attentional demand on learners. An additional motivation for the further development of wearable haptic devices is that they lend themselves to use in normal training environments. They also leave vision and hearing free for other tasks, which is particularly important for safety. To date, many haptic feedback systems are still in their early phase, and while there appears to be large potential, detailed studies focusing on design, long-term learning and practicability in running are needed to determine its effectiveness.

#### *5.3.4.4 Multimodal Feedback*

Multimodal feedback is where more than one form of feedback is received concurrently [50]. In daily life, multimodal, rather than unimodal, stimuli are present. Not only are humans used to simultaneously processing stimuli from different modalities, being presented with varied stimuli about an event could benefit our situational interpretation [50]. The only study to provide multimodal feedback to runners, used simultaneous audio-visual stimuli in the form of a traffic light symbol, with an associated variable pitched auditory alarm indicating acceptable, medium

and high PTA [110]. While this feedback appears logical and in line with stimuli we would typically receive in the natural environment, the short-term retention was less favourable than in other studies using single stimulus feedback [82,203,204]. To date there is no evidence to support the use of multimodal feedback to impact kinetic variables in runners.

### **5.3.5 Intervention Length**

While evidence exists that interventions ranging from two to six weeks in length can effect immediate changes in kinetic variables [82,97,110,159,203,204,211,213], no comparative studies have been conducted between differing intervention lengths and how this may impact on immediate changes, as well as the long-term persistence of these changes.

### **5.3.6 Motor Learning Considerations**

#### *5.3.6.1 Retraining Environment*

The retraining environment is intrinsically linked to the feedback modality; and is an important consideration that ultimately dictates the ecological validity of an intervention. The retraining environment can be classified in three ways: a laboratory setting with highly specialised equipment, a medical/rehabilitation clinic, or the runners' natural training environment. The advanced biomechanical metrics used by some researchers require technology only available in laboratory settings, making them inaccessible to most [204,207–209,211,214]. However, with more readily accessible metrics, such as sound intensity, cadence and tibial acceleration, interventions can take place in clinical settings, using relatively cheap equipment [82,83,97,110,159,210,212,213]. Despite the enhanced accessibility, many of these interventions still require a treadmill, which induce running technique changes [226].

The proliferation of mobile devices, small wireless inertial measurement units and accelerometers has enabled the measurement of many biomechanical metrics. The main advantages of this technology are that a therapist or coach may not be required for setup or feedback [97], and retraining programmes are available in a runners' normal environment [203]. While the research investigating retraining in a normal environment is still minimal [203], there is now the opportunity to establish whether this change to a natural environment, along with a runners ability to self-manage their retraining, increases the long-term success of the retraining programme.

#### *5.3.6.2 Instructions*

All but one of the reviewed studies [208] included instructions for a strategy, or way to run. In line with the internally orientated target variables, some researchers provided very directed, movement specific instructions (e.g. 'modify your foot strike pattern from a rearfoot to a forefoot strike' or 'run with your knee pointing straight ahead, while maintaining a level pelvis')

[204,212,213]. Most studies provided instructions that promoted an external focus (e.g. 'land softer' or 'make the vertical force trace line as smooth as possible') [39,82,83,110,207,210,211,214,227]. Two research groups did however seemingly alter the initial externally orientated focus of attention created by their selected target variable, by providing 'suggestive cues' (e.g. 'eliminate the buzzer noise by avoiding a rearfoot strike landing') [159,209].

It is thought that externally orientated attentional focus may enable runners to experiment with different running mechanics (e.g. foot-strike pattern, lower extremity compliance, etc.) to alter the target variable and ultimately select a gait modification method that best suits them [207,210]. It is therefore important that the instructions are carefully crafted to ensure they're not leading towards a particular technique change. There are clearly a number of factors related to target variables and instructions, and it is not possible to determine which are optimal for motor learning in runners. It has been suggested that individual response to cues is likely, meaning clinicians and researchers need to be able to adapt and tailor cues to ensure desired biomechanical changes, while limiting these to one or two at a time to avoid cognitive overload [48]. A comprehensive compendium of potential cues and instructions has been catalogued by previous authors [48].

### **5.3.7 Feedback Protocol Design**

#### *5.3.7.1 Training Schedule*

Feedback interventions are typically designed in accordance with a schedule, which defines the frequency, duration and volume of exercise and feedback over a defined period of time. A large proportion of studies conducted a single session of feedback only and therefore would best be classified as 'proof of concept', rather than true motor learning studies [39,83,97,208–210,212,214]. From the studies that did report multiple training sessions, five prescribed an identical schedule, with eight sessions over two weeks, with running time progressively increasing from 15 to 30 minutes [82,159,204,207,211]. Willy, et al. [203] were less prescriptive and precise in the reporting of their feedback training, while the most comprehensive training schedule was prescribed by Diebal et al. [213], where participants undertook 18 feedback sessions over six weeks. There is moderate evidence to suggest that eight sessions of 15 to 30 minutes of running over a two-week period is an appropriate dose of feedback stimulus to result in short-term changes [82,203,204,211].

#### *5.3.7.2 Feedback Timing and Withdrawal*

The amount or frequency of feedback presented in a session, and how and when it is withdrawn, are essential considerations for motor learning. While many studies reviewed provided constant feedback throughout the intervention [83,97,110,208–210,212–214], there

is evidence from motor learning literature that providing feedback as a constant stream of data may not be optimal due to the high levels of attentional demand required [228]. Previous studies have shown that subjects who receive intermittent or delayed feedback have performed better in the long-term, when compared to a continuous feedback schedule [229]. As highlighted in Section 5.5.7.1, five studies used a combined approach whereby participants were presented with continuous feedback for half of the feedback schedule, and intermittent feedback for the remaining sessions [82,159,204,207,211]. In this way, the feedback is progressively withdrawn or faded out. It has been suggested that this type of design can facilitate the internalisation of new movement patterns, thereby improving their persistence [230].

Willy et al. [203] used a feedback schedule in which runners self-determined how often they would receive feedback. With the inherent nature of this, it is impossible to know how much feedback each participant engaged in, however the short-term changes reported (ES: AVLRL 1.44; IVLR 1.21) provide some limited evidence that this may be appropriate with runners. Clansey et al. [110] report less substantial short-term changes at four weeks (ES: AVLRL 0.27; IVLR 0.35), while all those that used a faded withdrawal of feedback reported much larger reductions in these variables in the short-term (ES: AVLRL 1.1-1.51; IVLR 1.09-1.58) [82,203,204] and at three months (ES: AVLRL 1.66; IVLR 2.13) [159]. This provides moderate evidence that feedback withdrawal may be important to reduce the reliance on feedback [230].

### **5.3.8 Limitations**

One of the limitations of this review was that there were no studies that provided specifically included participants with tibial fatigue fractures, therefore the evidence from the included studies can only be presumed to relate to this particular cohort of runners. The interventions reviewed varied on many levels, with respect to augmented feedback type, targeted variables, instructions and in some cases, supplementary activities or interventions. It was thus impossible to determine to what degree changes in runners' performance are due to the augmented feedback itself, or the influence of other factors. This review focused on studies targeting kinetic variables identified as biomechanical risk factors for tibial fatigue fracture. This does not mean these variables are the most important clinically or biomechanically, these may simply be variables that are easier to manipulate or measure. There may also be other biomechanical variables related to the mechanism of fracture that have either not been targeted or have yet to be identified.

Motor learning and the assessment of long-term retention posed another limitation. Only a single low powered and poor quality study carried out a long-term retention test (3-months post-intervention) [159]. As a result, it is unclear what effects augmented feedback retraining



might have after there has been time for changes to be embedded within habitual running. It should also be acknowledged that due to the particular equipment used many interventions researched to date may not translate well to clinical practice. Additionally, the 'one size fits all' approach of interventions, where each participant is prescribed identical retraining parameters is not how individualised treatment typically occurs, nor is it likely the most effective.

## **5.4 Conclusions**

The results of this review found moderate evidence to support short-term reductions (four weeks) in vertical force loading rates as well as foot and tibial acceleration resulting from running gait retraining interventions facilitated by augmented feedback. Very limited evidence supported long-term changes (12 weeks) in force loading rates only. There was also moderate evidence to suggest that eight sessions over two weeks is an appropriate stimulus dose to result in short-term changes and that feedback withdrawal may be important to reduce the reliance on feedback. Visually augmented feedback is the only modality where there is enough research to provide moderate evidence for use with runners. Additional high-quality studies, with longer duration follow-up assessments are warranted to establish the optimal training parameters required for long-term habituation of gait modifications. Studies should be designed to facilitate individual analysis, so that responders and non-responders can be identified and individual motor learning styles accommodated. Further research also needs to be undertaken to determine the optimal variables to target and how these are aligned with cues and instructions. Additionally, where thresholds should be set to provide a challenging, yet achievable target.

## **5.5 Practical Recommendations**

To optimise the effectiveness of augmented feedback interventions with runners there are a number of key elements to consider.

- Runners should be screened to ensure they possess appropriate potential for change in the biomechanical variables being targeted.
- The feedback intervention interface should be designed in such a way as to prevent misinterpretation or high levels of cognitive demand.
- Retraining interventions would ideally be designed with user-friendly technology that facilitates use in clinics and the normal training environment.
- Feedback schedules should have a faded withdrawal of feedback to avoid overreliance and to encourage internalisation of gait changes.
- Finally, to ensure motor learning has occurred, long-term retention should be assessed after feedback withdrawal.

## **5.6 Chapter 5 Novel Contributions**

In this chapter I have reviewed and analysed all studies that have used augmented feedback to facilitate running gait retraining, but unlike previous reviews, the focus was interventions that had specifically targeted the kinetic risk factors associated with the development of tibial fatigue fractures. This review highlighted that there was evidence to support short-term reductions in tibial acceleration, as well as associated vertical ground reaction force loading rates, but only very limited evidence to support longer term changes and only in force loading rates. Almost all the studies reviewed analysed changes at the group level only, and therefore there was considerable masking of how many of the individual runners who engaged in the intervention were actually responders. To date, researchers have used visual or auditory feedback, or a combination of these two modalities. This review has provided clear support for the effectiveness of augmented feedback for modifying kinetic risk factors associated with TFFs in runners, it has also highlighted the considerable gaps in the current knowledge, and the need for additional research to address these.

## **Chapter 6:**

### **The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners.**

*This chapter is comprised of the following manuscript:*

Sheerin, K., Reid, D., Taylor, D., Besier, T. (2020). The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners. *Physical Therapy in Sport* (43), 173-180. DOI:10.1016/j.ptsp.2020.03.001 (Appendix 4.)

## **6 Preface**

As highlighted in Chapter 5, a number of studies have used augmented feedback gait retraining interventions to successfully modify kinetic risk factors for TFF. However, with a few exceptions, these interventions have used measurement tools (e.g. instrumented treadmills or motion capture systems) or feedback modalities (e.g. computer displays or TV screens) that are restrictive to a laboratory setting. Interventions such as these typically require runners to attend a clinic or laboratory, therefore placing time and location constraints on runners wanting to engage with them, while also lacking ecological validity. I adapted a real-time haptic feedback system that was previously successfully used to help modify walking gait. This system is lightweight and wireless, and therefore has the potential to be used in a normal training environment, but before this can take place it needs to be validated in a controlled setting. This chapter takes the form of an intervention study designed to assess the effectiveness of this novel system to reduce tibial acceleration in runners. The design of the wider retraining framework takes into account the findings and recommendations of Chapter 5.

## 6.0 Introduction

There are clear health and fitness benefits that come from regular participation in running [1,2], however running is also associated with a higher rate of overuse injury than other aerobic training [4,231,232]. Bony fatigue fractures (frequently termed stress fractures) are a common running injury [6], with the tibia being the most common site [177,178]. The causal mechanism of fatigue fractures is related to an imbalance between mechanical loading and an accumulation of damage of the tissue and the remodeling of the bone [233]. Various mechanical loading variables have been correlated to the chronicity and recurrence of tibial fatigue fractures, including surrogates such as tibial acceleration (TA) and instantaneous vertical force loading rates [35,180,233,234]. Previous studies have demonstrated that with appropriate feedback and guidance, running biomechanics can be successfully altered to reduce biomechanical injury risk factors in relatively short periods of time [82,97,110,203,204]. There is also some evidence that these changes can translate to injury reductions, at least in novice runners [207]. Within the context of changing injury-related biomechanical risk factors, it is not sufficient for runners to demonstrate a change in their running performance; they should be able to consistently reproduce an altered technique in the long-term, such that they have relearned this motor skill [235].

Several gait retraining interventions have used biofeedback from tibial mounted accelerometers to modify lower extremity biomechanics [39,82,83,97,110]. Compared to force plates, the benefit of using this surrogate measure of impact is that it enables measurement outside the laboratory [41]. Additionally, accelerometers are also considerably less expensive, and therefore more accessible for clinical applications [236].

To date, many real-time feedback systems used with runners have been designed in a way that lack ecological validity, and ultimately restrict the potential for their use in a normal training environment. For example, while visual feedback displays have been successfully used [82,208], this approach requires runners to remain indoors on a treadmill. While audio feedback, presented as either amplified the impact sound [214] or different audible pitches [39,237], are options that offers improved mobility, audio feedback has important safety considerations specifically when headphone use by pedestrians has been implicated in serious injury and death [238,239]. Tactile and kinesthetic feedback conveyed through vibrations on the skin (haptic feedback) is an option that has not yet been explored in this area [57]. Motivation for the use of wearable haptic devices is that they lend themselves to use in normal training environments, while also leaving vision and hearing free for other tasks [54]. Haptic feedback has been used more widely in walking gait retraining to reduce knee adductor moments in healthy individuals, those with osteoarthritis [60,225], and to provide independent feedback about multiple gait parameters within a single session [52,55]. However, its use with

runners is limited to providing feedback on directional information [240], rhythmic breathing patterns [241], and TA where haptic stimuli was paired with simultaneous visual and auditory feedback [56].

In sports medicine and biomechanics it is rare that a one-size-fits-all intervention is successful for each individual, however this is only occasionally acknowledged and not often discussed in running gait retraining literature [83,242]. In order to appropriately determine the effectiveness of an intervention, the acknowledgement and analysis of non-responders must occur alongside those who positively respond. Additionally, within the context of motor learning, consideration should also be given not only to whether participants immediately changed their performance after receiving feedback, but also the retention of these changes over a period with no feedback, as well as the ability to transfer changes to different environments [230]. The aim of this study was to examine the effectiveness of a real-time haptic feedback gait retraining system on reducing resultant tibial acceleration with individual runners. A secondary aim was to assess the retention of changes over four weeks, and the transfer of learning to over ground running.

## **6.1 Methods**

### **6.1.1 Participants**

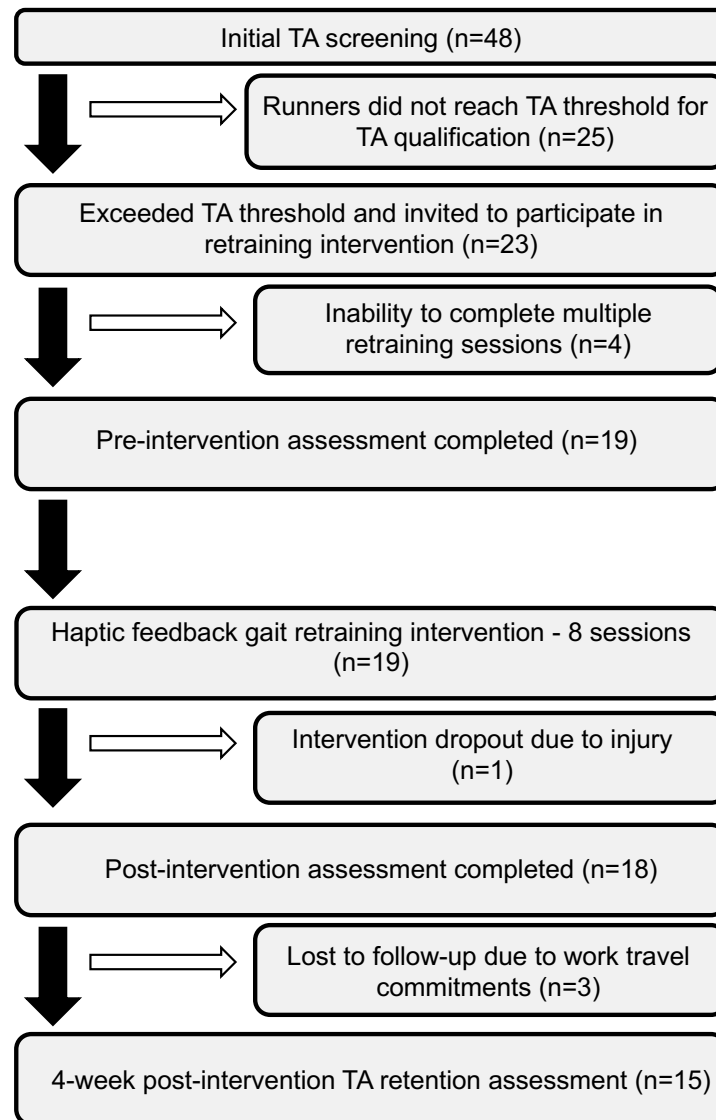
Runners between the ages of 18 and 65 years of age were invited to participate in this study if they were injury free, and had consistently run three or more times, to a combined weekly distance of at least 20 km for greater than six months. Procedures were approved by institutional human research ethics (AUTEC #15181), and participants signed written informed consent.

Using TA data from Zhang, et al. [242] an *a priori* power analysis ( $\alpha=0.05$ ,  $\text{power}=0.8$ ) was conducted with G\*Power (v3.1.9.3). Fifteen subjects were needed to adequately power the study. To account for potential drop-out, 19 runners were enrolled in the study.

### **6.1.2 Study Design**

To avoid a floor effect of the retraining intervention, participants were screened to determine their baseline resultant tibial acceleration (TA-R) (Figure 9). Runners were invited to participate in the intervention where the TA-R from either limb exceeded the threshold corresponding to their nominated training pace. Runners who accepted the invitation to participate in the intervention underwent a 3-week 'control' period to confirm the stability of their baseline TA-R. During this period runners continued with their typical training, before returning for the pre-intervention assessment. TA-R data measured during both treadmill and over ground running was recollected at this time point, as well as immediately post-intervention, and 4-weeks post-

intervention. During the four weeks that followed the intervention, runners reverted to their typical training. They received no feedback, guidance or oversight from the research team during this period. Injuries and adverse incidents were also monitored for the duration of the study.



**Figure 9: Flow chart describing the study recruitment screening and study design.**

### 6.1.3 Procedures

Acceleration data were collected via inertial measurement units (IMeasureU, Auckland, New Zealand) securely attached to the antero-medial tibiae, at a sample rate of 1000 Hz. In all instances, 30s of TA data were extracted for processing. TA-R data were processed and analysed according to previously described procedures [87]. Runners wore standardised neutral running shoes (Asics Kudrow, Kobe, Japan) and ran on an instrumented treadmill (Bertec, Columbus, OH, USA) that provided minimal shock absorption.

Following a 5-minute warm-up at a self-selected pace, participants ran at the velocities of 2.7, 3.0, 3.3 and 3.7 m/s, during which time their TA was recorded. Runners were free to adopt their natural running technique and no feedback was provided. After the screening assessment each runner nominated a speed from the four options, which matched closest to their comfortable 10km training pace. Owing to the relationship between faster running velocity and increased TA-R, different threshold levels were set for each velocity, corresponding to  $>1SD$  of a healthy group of runners at this speed (Table 9) [87]. Only runners who exceeded the threshold TA-R at their nominated training velocity were invited to participate in the intervention. If both limbs exceeded the threshold, the limb with the higher TA-R was used for the retraining portion of this study (hereafter referred to as the ‘target limb’).

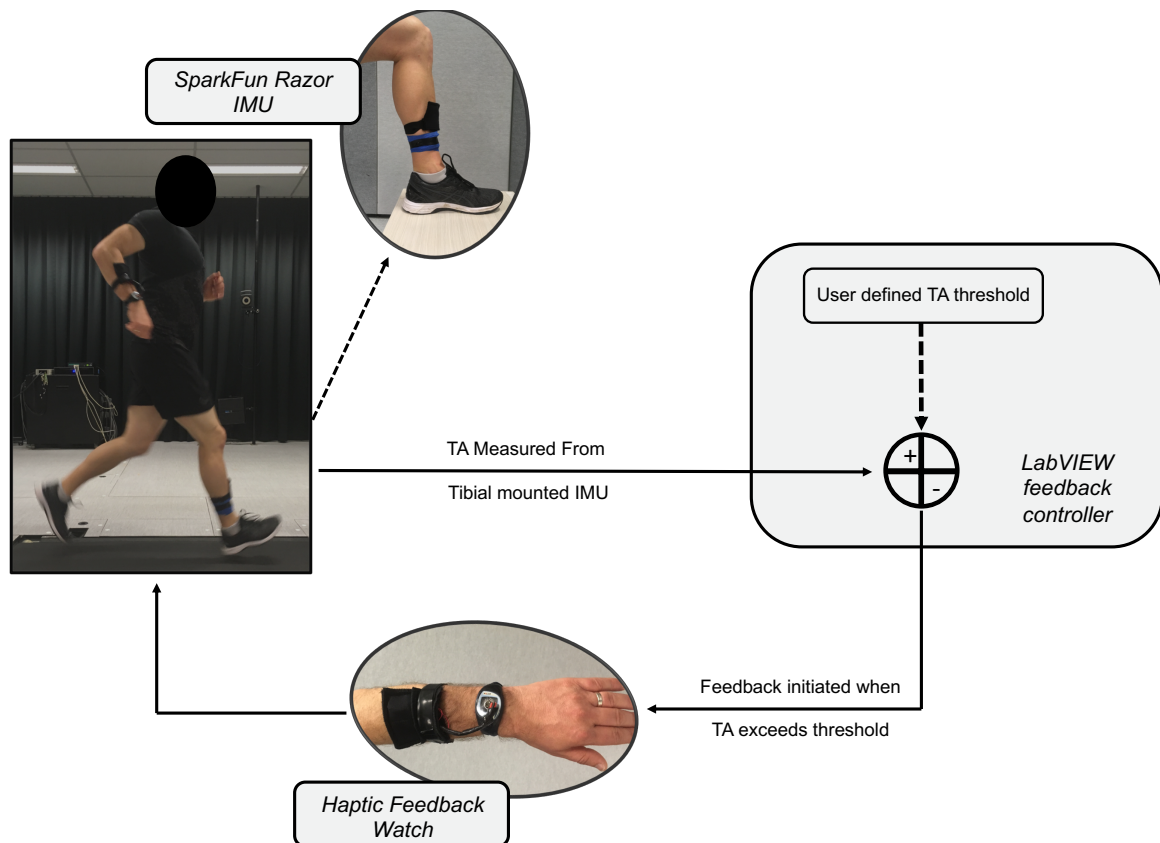
With tibial accelerometers still in place, additional over ground running assessments were conducted on standard athletics track during the pre-intervention, post-intervention and 4-weeks post-intervention assessments. Participants ran two complete full laps (corresponding to 800m) at a self-nominated pace, as close as possible to their nominated training velocity. Pace was recorded via a GPS watch (Garmin Forerunner 235, Garmin Ltd, Schaffhausen, Switzerland). Overground running pace during the post-intervention and 4-weeks post-intervention assessments were deemed acceptable when they were within  $\pm 5\%$  of the pre-intervention pace [110].

**Table 9: Resultant tibial acceleration screening thresholds**

Running velocity (m/s)	2.7	3.0	3.3	3.7
Resultant tibial acceleration (g)	12.5	14.0	15.2	16.6

#### **6.1.4 Gait retraining intervention**

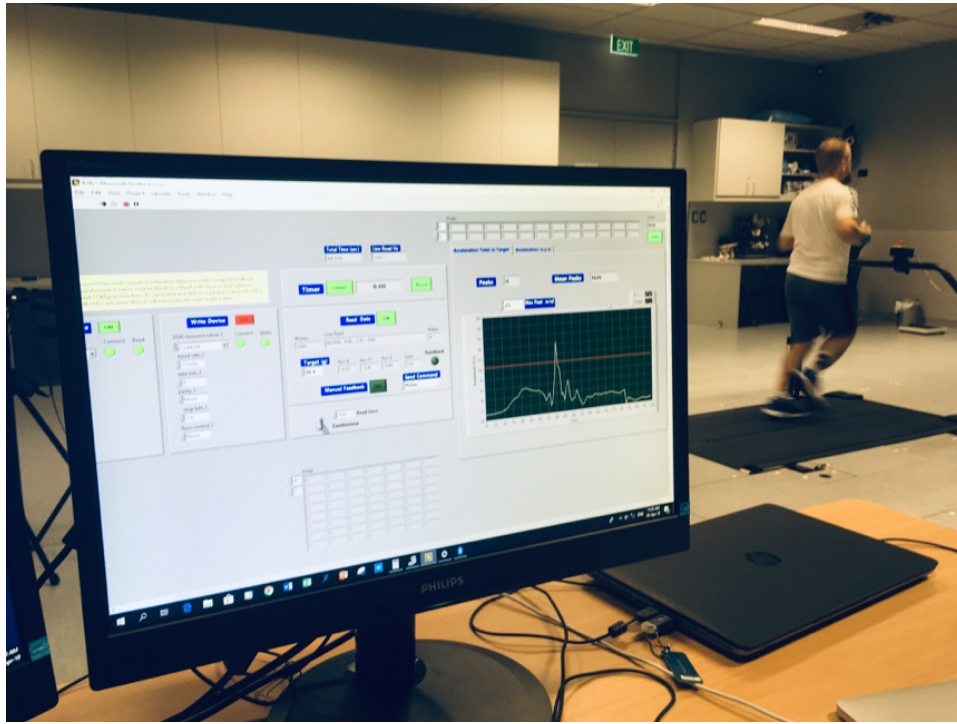
Prior to each retraining session a custom accelerometer (SparkFun 9DoF Razor IMU, SparkFun Electronics, CO, USA) was securely attached to the target limb while the controlling electronics were attached more proximally on the lower calf (Figure 10). A haptic feedback device previously described by Chen, Haller & Besier [57] was fitted into a watch, and attached to the wrist corresponding to the side of the target limb. Vibration actuators were located in the watch, while the controlling electronics were attached more proximally on the forearm. An integrated Arduino MICRO development board served as a controller driving four eccentric rotating mass motors to illicit a vibration.



**Figure 10: Block diagram of real-time haptic feedback system.**

The acceleration measured from a SparkFun IMU sensor was transmitted via Bluetooth to a computer running a custom LabVIEW programme (National Instruments Corp.; Austin, Texas, USA) which captured the raw signal and calculated TA-R (Figure 11). Where peaks in the TA-R (corresponding to foot-ground impact) exceeded the target threshold, an output signal was transmitted via Bluetooth to the haptic feedback watch. The feedback system was set without a delay on the haptic feedback, and the stimulus duration was set at 200ms. Within the LabVIEW programme was a function to switch off the feedback watch, so that users could measure from the accelerometer without providing feedback to runners. Additional detail on the real-time haptic feedback system and LabVIEW programme are provided in Appendix 15.

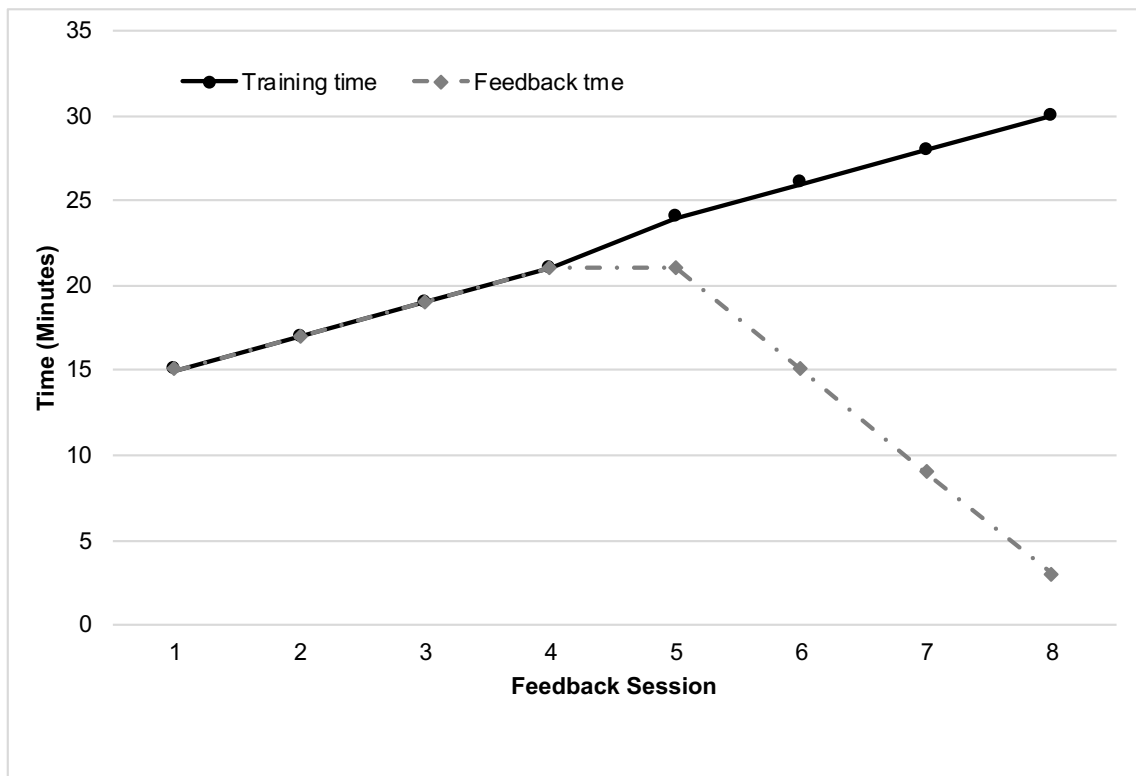




**Figure 11: Typical participant running on the treadmill while receiving real-time haptic feedback via custom software.**

In accordance with the challenge point framework [192], to ensure a challenging, yet achievable target, runners' TA-R was re-assessed prior to each retraining session. The target threshold was then set 10% below the session baseline. This proportional target has previously been successfully used with an audio intervention [39], and was re-confirmed through pilot testing with haptic feedback. During the intervention period, step-by-step peak TA-R were calculated, and a pulse was provided to the wrist by the feedback watch when the runner exceeded the session target threshold. Runners were instructed to make their footfalls softer and eliminate the vibration feedback [169].

A faded feedback design was used to encourage internalisation of the new running pattern, and the retraining schedule was as described by other research groups [82,207]. Runners participated in eight feedback sessions, which were carried out on a treadmill at the runner's nominated training velocity. The duration of each session increased step-wise from 15-minutes for the first session, to 30-minutes in the final session (Figure 12). Participants received feedback for 100% of the time during the first four sessions, and it was progressively withdrawn over the final four sessions, such that only three minutes of feedback was provided in the final session, with one minute at the start, middle and end of the session. Participants had a minimum of one day off after two consecutive days of running, and they were required to complete all running retraining sessions within a three-week period. During this time, participants were allowed to continue with their normal training.



**Figure 12: Gait retraining schedule**

### 6.1.5 Statistical Analysis

Statistical analyses were performed using SPSS statistics package version 24 (IBM Corporation, Armonk, NY, USA) with the level of significance set at  $p < 0.05$  throughout. Data normality and variance homogeneity were tested using the Shapiro-Wilk test. Group and individual statistics including medians (Md) and interquartile range (IQR) were calculated. The difference between TA-R measured pre- and post-intervention, as well as pre-intervention and 4-week follow-up, were analysed using a Wilcoxon Signed-Rank Test. Probability of Superiority Effect Sizes (ES) were calculated as:  $ES = ZScore / \sqrt{n}$ , and ranged from 0 to 1. An ES of 0.5 indicated a 50% probability that there was no difference, 1.0 indicated absolute certainty there was a difference, between the scores being compared. An ES of 0.9 indicate almost certainty, 0.7-0.8 likely probably, and 0.6 only slightly greater than an even chance of a difference between scores [243]. A multi-point criterion was used to define runners who positively responded, and therefore the intervention was only considered successful where an individual runner achieved each of the three criteria outlined in Table 10.

**Table 10: Multi-point criteria to define positive responders to intervention.**

- 
1. Target threshold TA-R for the selected training velocity is achieved during treadmill running.
  2. A statistically significant reduction ( $p < 0.05$ ), and a likely probable ES ( $> 0.7$ ) when pre-intervention TA-R was compared to both post-, and 4-week post-intervention during treadmill running.
  3. A statistically significant reduction ( $p < 0.05$ ), and a likely probable ES ( $> 0.7$ ) when pre-intervention TA-R was compared to both post-, and 4-week post-intervention during overground running.
- 

## 6.2 Results

Forty-eight runners were screened for eligibility (Figure 9), of which 23 demonstrated TA-R measures that exceeded the pre-defined threshold (Table 9). Nineteen runners agreed to participate in the gait retraining intervention and therefore completed the pre-intervention assessment. One participant dropped out of the study after the first intervention session due to an injury unrelated to running. The remaining 18 participants (7 female, 11 male; age:  $35.2 \pm 9.6$  years) completed all intervention sessions and the post-intervention assessment (Table 11). Participants had regularly participated in running for an average of  $8.5 \pm 9.5$  years and ran  $42.4 \pm 22.2$  km over  $4.2 \pm 1.2$  weekly training sessions. Three participants were lost to follow up due to being overseas with work commitments ( $n=2$ ) or moving from the study region ( $n=1$ ). All 18 runners who completed the haptic feedback gait retraining, and the runners who returned for the 4-week follow-up, did so without any injuries or adverse effects.

**Table 11: Participant characteristics used in Chapter 6.**

	Male (n=11)	Female (n=7)	Combined (n=18)
Age (years)	$38.3 \pm 8.2$	$30.4 \pm 10.4$	$35.2 \pm 9.6$
Height (m)	$1.81 \pm 0.07$	$1.67 \pm 0.06$	$1.75 \pm 0.10$
Mass (kg)	$82.5 \pm 10.2$	$58.7 \pm 4.5$	$73.3 \pm 14.5$
Running history (years)	$6.8 \pm 8.9$	$11.2 \pm 10.4$	$8.5 \pm 9.5$
Weekly running distance (km)	$36.7 \pm 18.1$	$51.4 \pm 26.5$	$42.4 \pm 22.2$
Weekly running frequency (days / week)	$4.0 \pm 1.3$	$4.6 \pm 0.9$	$4.2 \pm 1.2$

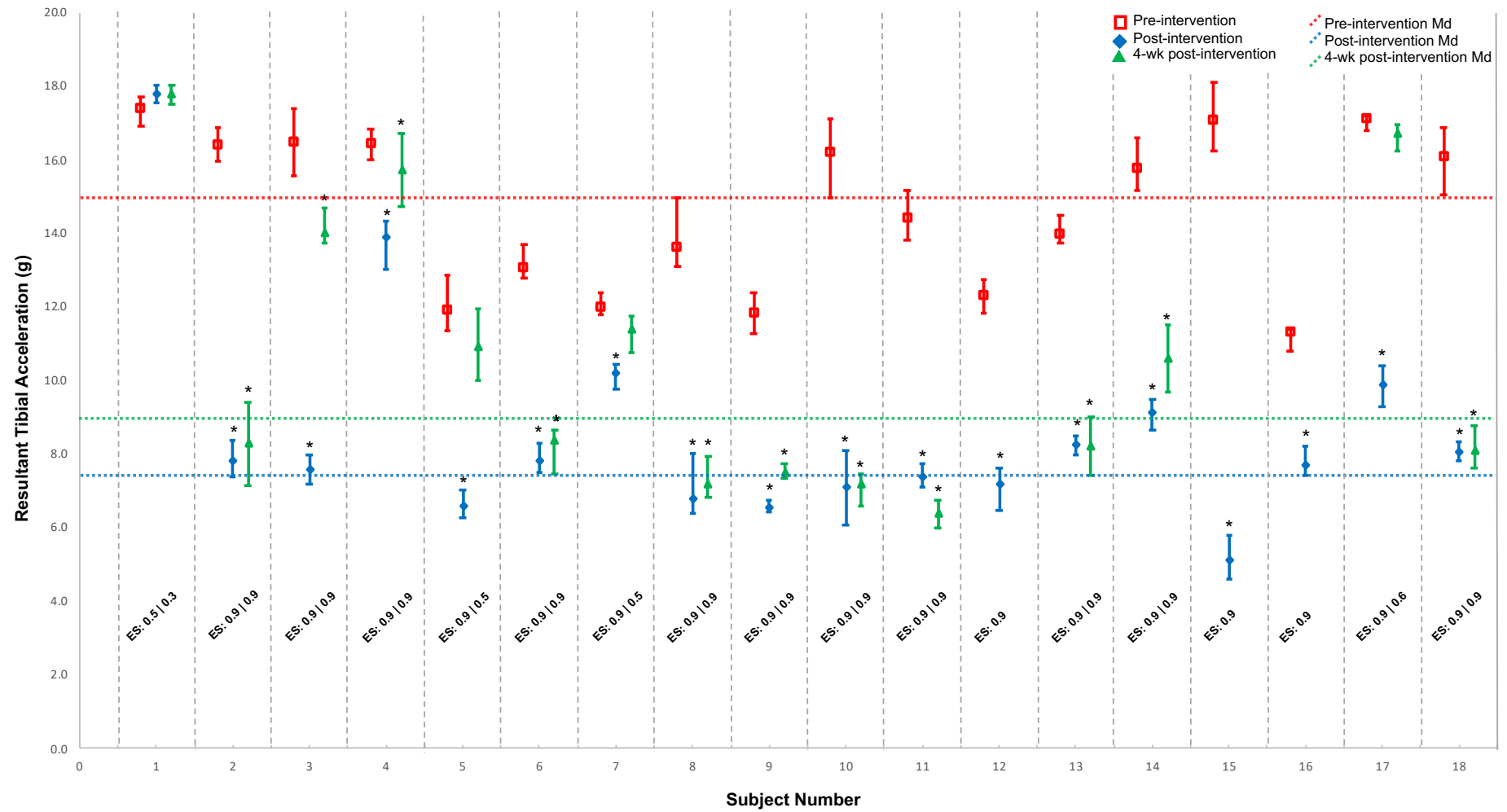
The Shapiro-Wilk test showed that the data were not normally distributed and therefore non-parametric statistics were used for analysis. Comparing pre- to post-intervention TA-R measures from treadmill running, the median decreased by 50% (Table 12). A Wilcoxon Signed-Rank Test revealed a reduction in TA-R following the intervention ( $z = -18.2$ ,  $p < .001$ ), with an ES indicating almost certainty of a change ( $ES = 0.9$ ). Comparing TA-R measures at these same time points from over ground running, the median decreased by 28% ( $z = -13.2$ ,  $p < .001$ ).

**Table 12: Group-based resultant tibial acceleration results for treadmill and over ground running.**

Condition	Assessment Time Points	TA-R (g) Md (IQR)	Md Diff (%Diff)	Z	p	ES
Treadmill	Pre-intervention	14.8 (12.5, 16.6)				
	Post-intervention	7.4 (6.1, 9.2)	-7.4 (-50%)	-18.2	<0.001	0.9
	4-wk post-intervention	8.7 (7.4, 13.7)	-6.2 (-41%)	-12.9	<0.001	0.8
Overground	Pre-intervention	19.4 (17.3, 20.6)				
	Post-intervention	14.0 (11.8, 17.7)	-5.4 (-28%)	-13.3	<0.001	0.7
	4-wk post-intervention	16.1 (10.5, 18.6)	-3.3 (-17%)	-11.3	<0.001	0.7

*For each condition, results from post-intervention and 4-wk post-intervention assessments were compared to the pre-intervention assessment.*

When TA-R from treadmill running were compared from pre-intervention to 4-weeks post-intervention, the median TA-R decreased by 41% (Table 12). A Wilcoxon Signed-Rank Test confirmed a reduction in TA-R 4-weeks post-intervention ( $z=-12.9$ ,  $p<.001$ ), with an ES indicating a likely probability of a true reduction ( $ES=0.8$ ). The median reduction in TA-R between these same time points from the over ground running was 17%. All but two runners responded initially positively to the intervention, meeting the criteria outlined in Table 9. This reduced to 11 runners at the 4-week post-intervention follow-up (Figures 13 and 14).



**Figure 13: Resultant tibial acceleration during treadmill running at pre-intervention, post-intervention and 4-weeks post-intervention assessments.**

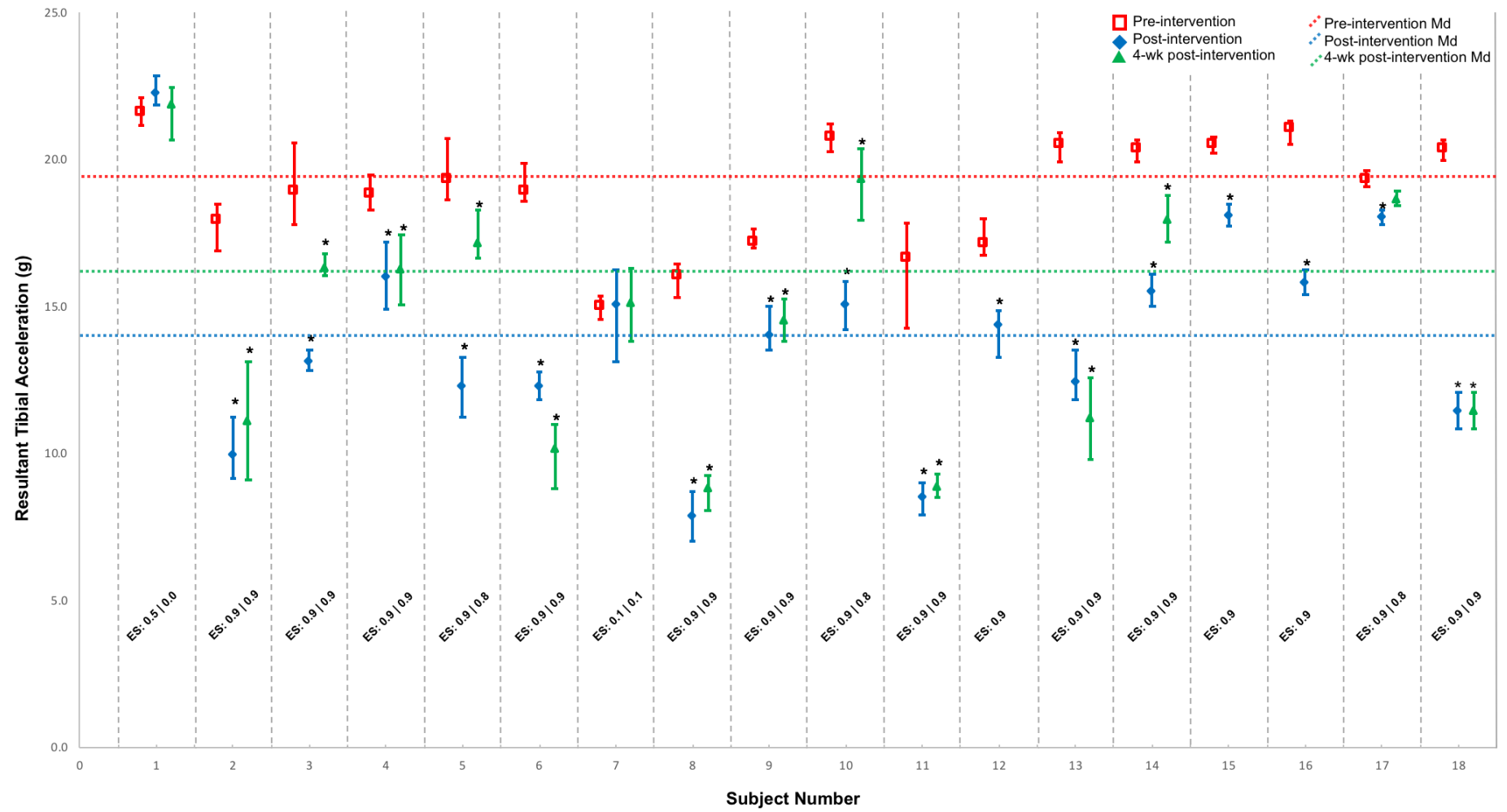


Figure 14: Resultant tibial acceleration during overground running at pre-intervention, post-intervention and 4-weeks post-intervention assessments.

### 6.3 Discussion

The primary aim of this study was to examine the effectiveness of a real-time haptic feedback gait retraining system for reducing resultant tibial acceleration with individual runners, with a secondary aim to assess the retention and the transfer of learning to over ground running. Across the cohort of runners studied, this intervention was associated with a 50% reduction in TA-R at the cessation of the intervention, and a 41% reduction 4-weeks later. The magnitude of these changes are amongst the highest reported when compared similar designed studies that have measured axial TA [39,82,83,97,110]. When overground running, tibial acceleration were in all cases higher in magnitude than treadmill running. Reductions in TA-R were measured at the post- and 4-week post-intervention time points, with these being somewhat smaller than the changes during treadmill running (28% and 17% respectively). This is a consistent trend observed in previous studies [83,110,242]. The differences when compared to treadmill running are likely due to a greater inconsistency in running pace and reduced attentional focus due to inherent distractions of a normal training environment.

While the obvious difference in design between this intervention and those previously published is the use of a haptic feedback device, there were other notable differences. With regard to participant recruitment, while most studies conducted screening based on a target biomechanical variables (e.g. TA or AVLr), a number of studies specifically targeted rearfoot strike runners [82,110,208,212,213]. Additionally, other researchers used a single velocity (commonly 3.7 m/s) to screen all potential participants [82,97,110,203,212,214]. Both of these factors could have biased the study recruitment, and as a result, recruitment of participants for this study was not restricted to any other biomechanical criteria other than high TA-R. Screening was also conducted at four different velocities, and runners chose the velocity which most closely matched their 10km training pace. Participants were only recruited for the intervention where they exceeded the pre-defined TA-R threshold at their individually selected velocity. Subsequent gait assessments and retraining were all carried out at this velocity. These measures ensured that each participant was undertaking the intervention at a comfortable running pace which had the potential to optimise engagement with the feedback.

At the individual level, the ultimate success of the intervention was determined by a multi-point criterion for both treadmill and over ground running. Based on these criteria, 61% of runners who completed the intervention were classified as positive responders. At the post-intervention time point this success rate was similar to that reported when a visual feedback intervention was used [242]. A degree of drop-off in retention was noted at the 4-week post-intervention mark, however, previous studies have not assessed changes at the individual level for this length in time so no direct comparisons can be made [83,242].

It has previously been highlighted that flexibility in feedback protocols is important to optimise individual motor learning [194,242], and therefore as part of this study, rather than having a set proportional target threshold that remained constant for the duration of the intervention, a 10% target threshold was reset before each intervention session to ensure that runners had a challenging, yet achievable goal. While the reaction from participants to the haptic feedback intervention was largely positive, many runners struggled during initial sessions determining their current TA-R level in relation to the threshold. With the simple feedback design used, runners received the same feedback stimuli if they just exceeded, or greatly exceeded the target threshold. While there is substantial evidence to support the 8 session, faded feedback protocol design can be effective [82,159,204,207,211], there is no evidence that this was the optimal design for each individual in this cohort. Superior results may have been achieved with greater or lesser number of sessions, of a shorter or longer duration. These are two areas where there is clearly a need for additional design optimisation and research validation with runners. While we support the identification of non-responders, there is further work required to improve the early identification of these individuals, so that the time and effort of clinicians and runners is not needlessly expended.

While this study presents positive findings, at least for a large proportion of high tibial load runners, it is not without limitations. Due to the lack of a control group it was not possible to establish causation in relation to the intervention. Also, only a relatively small sample of healthy participants were recruited and hence our findings may not be generalised to injured runners, or those with differing characteristics. Additionally, the follow-up timeframe assessment timeframe was relatively short, and the incidence of subsequent injuries were not recorded or analysed. There is some evidence that changes in biomechanical risk factors from similar interventions can reduce the incidence of injuries in novice runners [207], but these findings need to be replicated for a more experienced cohort as in this study. There were also a number of other elements that were not controlled within the study, this included the participants' running training outside the feedback protocol, as well as the footwear the runners chose to use during feedback sessions. Both of these factors could have impacted on the training, and the translation of the learning effects.

## **6.4 Conclusions**

After a 4-week follow up, 61% of high tibial load runners reduced their tibial acceleration both on a treadmill and overground after undertaking a real-time haptic feedback gait retraining intervention. This intervention appears to be as effective, but less invasive and expensive, compared to other more established modalities, such as visual feedback. This new approach to movement retraining has the potential to revolutionise the way runners engage in gait retraining.



## **6.5 Chapter 6 Novel Contributions**

This is the first study to exclusively use a haptic feedback intervention to alter the biomechanics of runners. At the group level I observed amongst the largest reductions in tibial acceleration reported in the literature. While the intervention used in this study was in a laboratory based setting, it appears to be as effective as other visual and auditory modalities, while addressing the accessibility and safety concerns that have existed with many alternative approaches.

It is widely acknowledged that in sports medicine and biomechanics that it's unlikely that a single intervention will be successful for all individual cases. Therefore, as part of this study I also assessed the effectiveness of the real-time haptic feedback intervention at the individual level, which has been rarely undertaken previous research. In an attempt to classify individual responders, I established a criteria which encompassed assessing the magnitude of change in tibial acceleration immediately after, as well as 4-weeks after the cessation of the intervention, while runners ran on both a treadmill and overground. This is clinically-relevant to assess the success of an intervention, and will ultimately help to identify those runners that will benefit from this type of approach.

## **Chapter 7:**

### **The biomechanical effects of real-time haptic feedback gait retraining on runners with high tibial acceleration.**

*This chapter is comprised of the following manuscript:*

Sheerin, K., Reid, D., Taylor, D., Besier, T. The biomechanical effects of real-time haptic feedback gait retraining on runners with high tibial acceleration. Submitted to *Journal of Applied Biomechanics*.

## **7 Preface**

Traditional methods for correcting movement, which are widely used in practice, has been to identify an error and use instructions to correct it. An alternative approach is the promotion of exploration of various strategies to facilitate the development of individual movement patterns that still achieve the outcome goals. In Chapter 7 I outlined the success of our real-time haptic feedback intervention to reduce tibial acceleration, both at a group level and with individual runners. With a view to enhancing individual motor learning, it is important to appreciate not only which runners responded to the intervention, but also how they responded. With this level of information practitioners will be able to guide those using similar haptic feedback interventions on what self-selected strategies used by runners to achieve changed in tibial acceleration.

## 7.0 Introduction

Despite the notion that individuals adopt a 'preferred movement path' when running [244], there is evidence that the biomechanics adopted by runners can actually be a contributing factor to the development of running related injuries [31,35,245,246]. Targeted gait retraining, or the deliberate modification of running technique, has been advocated as a strategy for the prevention [207], and rehabilitation [49] of running related injuries. The focus of many recent gait retraining studies has been to provide real-time feedback with the goal of altering biomechanical variables in order to prevent [82,97,110,203,207,247], or treat [213,248] running injuries.

It has been suggested that within the context of changing injury-related biomechanical risk factors, it is not sufficient for runners to demonstrate a change in their running performance; rather a need to consistently reproduce an altered technique in the long-term, therefore demonstrating true motor learning [45,50]. There is, however, some evidence, such as the feedback protocol design, the retraining environment, and the feedback timing and withdrawal that suggests research in this area has often not been based on best practice of how individuals actually learn [47–50,249]. Traditional methods for teaching or correcting movement has been to identify an error and use instructions to correct it [249]. An alternative approach is the promotion of decision making and self-organising behaviours to facilitate the emergence of functional movement patterns [249,250]. Adopting this approach, while also providing augmented movement feedback, potentially reduces the need to provide verbal instructions [249].

While it has previously been highlighted that flexibility in feedback protocols is important to optimise individual motor learning with runners [194,242], across the different gait retraining interventions that have been examined, almost all have analysed the effect of the intervention at a group level [82,110,203,210,251], with individual performance rarely mentioned [208,242,251]. However, in sports medicine and biomechanics it is rare that a one-size-fits-all intervention is successful for all. With a view to individual motor learning, a crucial area to measure and understand is how changes in human movement are achieved during goal-directed activity [49,251,252]. This will provide information to coaches, clinicians and scientists on what mechanical strategies runners are using during gait retraining to further guide the development of this approach [49,251]. The primary aim of this study was to determine the self-selected strategies used by runners to achieve a reduction in tibial acceleration (TA). Many researchers have targeted changes in cadence [203,208,253–257], or modification of foot strike angle [208,255,257] as successful strategies to reduce lower limb loading variables in runners. Due to their potential importance to these approaches, the secondary aims were to investigate whether there were differences in strategies used between runners who presented

at baseline with a rearfoot strike (RFS) or non-RFS landing pattern, as well as runners who were categorised as having either a low or high cadence at baseline. It was hypothesised that runners would be very individualised with their change in biomechanics. An additional hypothesis was that runners who presented with a low baseline cadence at baseline would increase their cadence, while those who presented with a RFS would adopt a non-RFS pattern.

## **7.1 Methods**

### **7.1.1 Participants**

Runners between the ages of 18 and 65 years of age were invited to participate in this study if they were injury free, and consistently ran at least three times per week for a minimum combined distance of 20 km, weekly for greater than six months. The procedures were approved by institutional human research ethics (AUTEC #15181), and participants signed written informed consent.

### **7.1.2 Procedures**

#### *7.1.2.1 Screening*

The initial screening of runners was part of a larger study [116,258], where TA data were collected at the the velocities of 2.7, 3.0, 3.3 and 3.7 m/s. Acceleration data were collected via inertial measurement units (IMUs) (Blue Thunder, IMeasureU, Auckland, New Zealand) securely attached to the antero-medial tibiae, at a sample rate of 1000 Hz. In all instances, 30s of TA data were extracted for processing. Peak resultant tibial acceleration (TA-R) data were processed and analysed according to previously described procedures [87]. Runners ran on an instrumented treadmill (Bertec, Columbus, OH, USA) that provided minimal shock absorption, and wore standardised neutral running shoes (Asics Kudrow, Kobe, Japan) for the pre- and post-intervention assessments, but their own running shoes for retraining sessions.

#### *7.1.2.2 Gait retraining intervention*

Runners were invited to participate in the retraining intervention when their baseline TA-R from either limb exceeded a pre-defined threshold at their nominated running velocity, corresponding to a comfortable 10km training pace [121] (Table 9). An eight session gait retraining protocol was used over a 3-week period, as described by other research groups [82,207]. The duration of each session increased step-wise from 15-minutes for the first session, to 30-minutes in the final session. A faded design was used whereby participants received feedback for 100% of the time during the first four sessions, and feedback was progressively withdrawn over the final four sessions, such that only three minutes of feedback was provided in the final session. Runners were permitted to continue with their normal training during the intervention period.

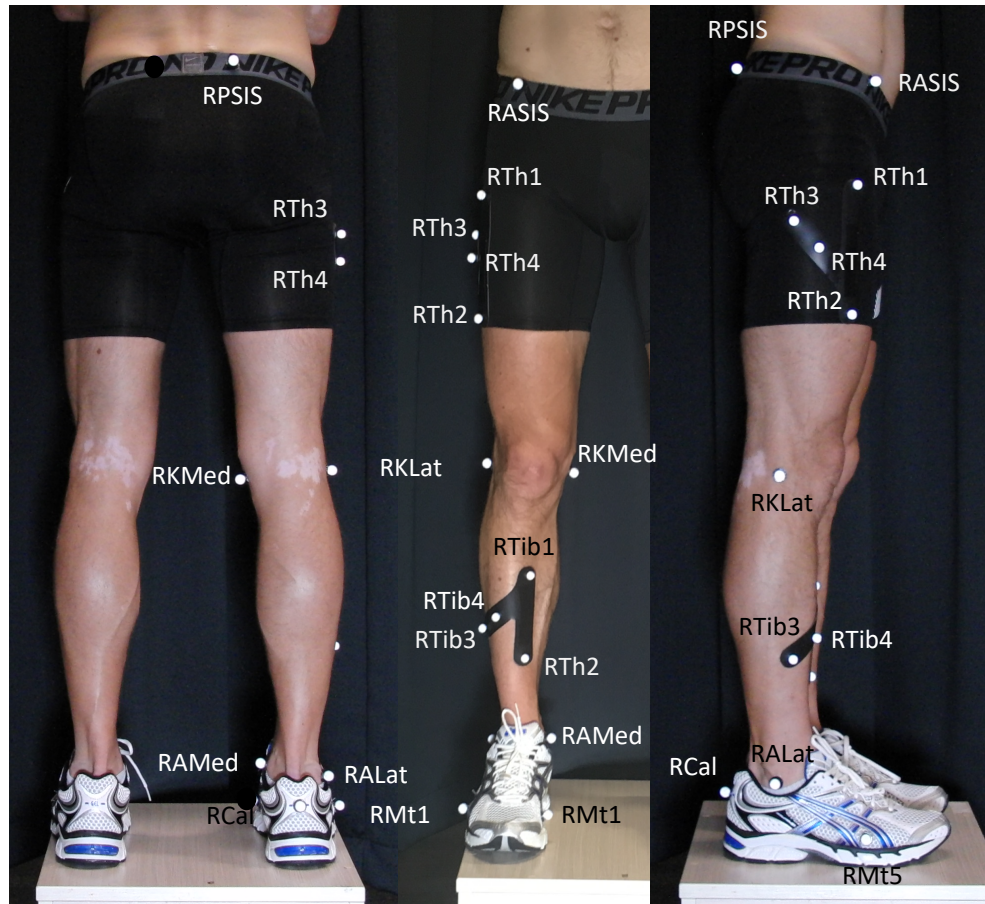
The limb with the highest TA-R was used throughout as the target retraining limb. A custom-made haptic feedback system, consisting of an IMU (SparkFun 9DoF Razor IMU, SparkFun Electronics, CO, USA) and a vibrating feedback watch, were securely attached to the distal medial aspect tibia of target limb, and the participant's wrist on the same side (Figure 10). The acceleration measured from the tibia was transmitted via Bluetooth to a computer running a custom LabVIEW programme (National Instruments Corp.; Austin, Texas, USA) which captured the raw signal and calculated TA-R. Where peaks in the TA-R (corresponding to foot-ground impact) exceeded the target threshold, an output signal was transmitted via Bluetooth to the haptic feedback watch. The feedback system was set without a delay on the haptic feedback, and the stimulus duration was set at 200 ms. Within the LabVIEW programme was a function to switch off the feedback watch, so that users could measure from the accelerometer without providing feedback to runners. For each retraining session, after the 5-min warm-up, TA-R was measured for 1-min. The average TA-R from this period was defined as the 'session baseline', and 10% below this figure was used as the target threshold for the following retraining session. During the intervention period, step-by-step peak TA-R were calculated, and a pulse was provided by the feedback watch when the runner exceeded the target threshold. Runners were instructed to make their footfalls 'softer' and 'eliminate the vibration feedback' [169]. Runners were free to continue with their normal frequency and volume of running training outside of the study protocol, and were encouraged to practise the strategies they had found to be successful during feedback sessions.

#### *7.1.2.3 Pre- and post-intervention gait analysis*

Prior to, and immediately following, the intervention, runners underwent a full three-dimensional (3D) gait analysis at their nominated running velocity. During the pre-intervention running assessment runners were not provided with any instructions relating to technique. During the post-intervention assessment runners were instructed to run as they had during the feedback sessions. Tibial acceleration was collected using the methods outlined for the screening. Kinematic data were collected via an eight-camera Vicon motion analysis system (Vicon, Oxford, UK) sampling at 200 Hz. Kinetic data were sampled at 1000 Hz from the instrumented treadmill.

To create a 3D model for analysis, a modified version of the University of Western Australia lower-body model was used [259]. Clusters of four markers, on thermo-moulded plastic shells were attached to anterior-lateral thigh (distal and lateral to avoid the bulk of muscle) and anterior shank (along the tibia). Anatomical markers were attached bilaterally to the anterior superior iliac spine, posterior superior iliac spine, medial and lateral femoral condyle, medial and lateral malleoli, posterior calcaneus, and head of the first and fifth metatarsals (Figure 15). Static calibration and functional joint dynamic trials were recorded, before anatomical markers

on the knees and ankles were removed. Data processing and modelling occurred within Vicon Nexus (v2.7), including the calculation of functional hip and knee joint centres, using the in-built Score and Sara functions [259]. Following a 5-minute warm-up, 30s of data were collected while participants ran at their nominated running velocity. Runners carried out an warm-up of progressively increasing velocity for 5-minute warm-up, after which 30s of data were collected while participants ran at their nominated running velocity.



**Figure 15: Lower body kinematic marker locations**

### 7.1.3 Data processing

The primary outcome variable was the peak resultant tibial acceleration. Additional variables of interest were chosen due to their contribution to higher TA based on previous studies [27,40,104,110,114,120,127,141,142,203,253,256,260,261], and their potential for use to cue lower TA in a clinical setting [48]. Marker, force and acceleration data were filtered at 8 Hz, 50 Hz, and 60 Hz with a fourth order zero lag Butterworth low pass filters, respectively [82,87]. Joint angles, velocities and moments were calculated via inverse dynamics using a Vicon BodyBuilder script, before being exported to Matlab (Mathworks, MA, USA) for further processing.

Data from the target feedback limb from each runner were extracted for further analysis. Cadence was defined as the number of left and right ground contacts in 60s and was calculated by multiplying the figure recorded in 30 s by two. The start and end of the ground contact phase were identified where the vertical ground reaction force (GRF) exceeded a 20 N threshold [256]. The vertical impact peak (VIP) was defined as the distinct impact transient in the vertical force generated after foot contact. In case of an absence of a VIP, the force value at 13% of stance phase was used, since this has been reported to be the average location of the VIP when one is present [262,263]. The loading phase was defined as the period of time between foot contact and VIP [209,264]. An impulse-momentum modelling approach was used to calculate effective mass, which was calculated as:  $\text{Effective mass} = \text{vertical GRF integral} / (\Delta \text{foot velocity} + g * \Delta \text{time})$  [264,265]. The change in foot velocity was quantified as the average vertical foot velocity of the ankle joint centre (as determined by Vicon Nexus) across the loading phase [143,265].

Contact angles for the hip, knee and ankle were determined as the sagittal plane joint angle at the time of vertical force contact. The footstrike angle was calculated by subtracting the ankle contact angle from the ankle angle during static standing calibration [266]. Footstrike patterns were classified according to established criteria, whereby a RFS was deemed when the footstrike angle was greater than 8° [266]. Sagittal plane joint excursion was then calculated for the loading phase. Leg stiffness during loading was calculated by dividing VIP magnitude by the vertical excursion of the centre of mass (COM) over this phase. The COM excursion was estimated from the vertical displacement of the sacral virtual marker (as determined by Vicon Nexus) [267,268]. Joint stiffness for the hip, knee and ankle joints were calculated by dividing the joint excursion by the change in joint moment during loading [269,270]. Effective mass, stiffness and joint moment variables were normalised to body mass [35,208,265,267,270,271].

#### **7.1.4 Statistical analysis**

Data normality and variance homogeneity were tested using the Shapiro-Wilk test. Descriptive statistics including medians (Md) and interquartile range (IQR) were calculated. Initial analyses were conducted on the entire group (n=18). The effect of haptic feedback retraining was also examined on runners with either a RFS or a non-RFS, as well as those with low (<170 steps/min) or high cadence (>170 steps/min), at baseline [272]. The difference between pre- and post-intervention scores for key biomechanical variables were assessed using a Wilcoxon Signed-Rank Test. Probability of Superiority Effect Sizes (ES) were calculated as:  $ES = Z\text{Score} / \sqrt{n}$ , and ranged from 0 to 1. An ES of 0.5 indicated a 50% probability that there was no difference, 1.0 indicated absolute certainty there was a difference between the scores being compared. An ES of 0.9 indicate almost certainty, 0.7-0.8 likely probably, and 0.6 only slightly

greater than an even chance of a difference between scores [243]. Statistical analyses were performed using SPSS statistics package version 24 (IBM Corporation, Armonk, NY, USA) with the level of significance set at  $p < 0.05$  throughout. At the group level, difference in pre- and post-intervention scores were indicated where statistically significant change was associated with an ES indicating greater than likely probability of change. To be classified as a positive responder to the feedback, individual runners also needed to demonstrate a reduction in TA-R such that it was equal to or below the TA-R threshold corresponding to their nominated running velocity.

## 7.2 Results

Forty-eight runners were screened for eligibility (Figure 9), of which 23 demonstrated TA-R measures which exceeded the pre-defined threshold (Table 9). Nineteen runners agreed to participate in the gait retraining intervention and therefore completed the pre-intervention assessment. One participant dropped out of the study after the first intervention session due to an injury unrelated to running. The remaining 18 participants (7 female, 11 male; age:  $35.2 \pm 9.6$  years) completed all intervention sessions and the post-intervention assessment (Table 13). Participants had regularly participated in running for an average of  $8.5 \pm 9.5$  years, and ran  $42.4 \pm 22.2$  km over  $4.2 \pm 1.2$  weekly training sessions.

**Table 13: Participant characteristics used in Chapter 7.**

	Male (n=11)	Female (n=7)	Combined (n=18)
Age (years)	$38.3 \pm 8.2$	$30.4 \pm 10.4$	$35.2 \pm 9.6$
Height (m)	$1.81 \pm 0.07$	$1.67 \pm 0.06$	$1.75 \pm 0.10$
Mass (kg)	$82.5 \pm 10.2$	$58.7 \pm 4.5$	$73.3 \pm 14.5$
Running history (years)	$6.8 \pm 8.9$	$11.2 \pm 10.4$	$8.5 \pm 9.5$
Weekly running distance (km)	$36.7 \pm 18.1$	$51.4 \pm 26.5$	$42.4 \pm 22.2$
Weekly running frequency (days / week)	$4.0 \pm 1.3$	$4.6 \pm 0.9$	$4.2 \pm 1.2$

The Shapiro-Wilk test showed that data were not normally distributed and therefore non-parametric statistics were used for analysis. Group based descriptive statistics and outcomes of the Wilcoxon Signed-Rank test for key biomechanical variables are presented in Table 14. Seventeen of the 18 runners that completed the intervention positively responded to the intervention (Figure 16a). At the group level, runners demonstrated a 50% reduction in TA-R, with an ES indicating almost certainty of a change ( $z = -18.2$ ,  $ES = 0.9$ ). Three variables showed consistent changes across all runners. Reductions were seen in both the change in foot velocity at impact ( $z = -16.0$ ,  $ES = 0.8$ ) (Figure 16b) and leg stiffness ( $z = -14.2$ ,  $ES = 0.7$ ) (Figure 16c) after the intervention, in addition to increases in cadence ( $z = -14.6$ ,  $ES = 0.7$ ) (Figure 16d) across the group. No consistent changes were observed in effective mass (Figure 16e), COM



excursion (Figure 16f), or the contact angles, joint excursions or stiffness at the hip, knee and ankle joints post-intervention.

**Table 14: Biomechanical variables before and after the gait retraining intervention.**

Variable	Pre-intervention (Md and IQR)	Post-intervention (Md and IQR)	Md Difference (%Diff)	Z	p	ES
TA-R (g)	14.8 (12.5, 16.6)	7.4 (6.1, 9.2)	-7.4 (-50%)	-18.2	<0.01	0.9
Cadence (steps/min)	172.8 (170.9, 178.1)	183.5 (175.6, 190.1)	10.7 (6%)	-14.6	<0.01	0.7
COM excursion (cm)	3.3 (3.0, 3.7)	3.0 (2.7, 3.5)	-0.2 (-7%)	-6.7	<0.01	0.3
Effective mass (%BW)	3.1 (2.4, 3.9)	3.7 (2.8, 4.6)	0.7 (23%)	-10.9	<0.01	0.5
Foot impact velocity (m/s)	5.2 (4.2, 6.1)	3.6 (2.3, 5.3)	-1.6 (-31%)	-16.0	<0.01	0.8
Leg stiffness (N/m/kg)	435.5 (376.1, 525.7)	368.1 (318.1, 450.5)	-67.4 (-16%)	-14.2	<0.01	0.7
<b>HIP</b>						
Contact angle (°)	32.5 (26.8, 39.5)	31.8 (20.5, 38.2)	-0.7 (-2%)	-6.7	<0.01	0.3
Joint excursion (°)	2.1 (1.1, 2.9)	2.3 (1.4, 3.4)	0.2 (10%)	-5.2	<0.01	0.2
Joint stiffness (Nm/°/kg)	17.7 (11.4, 31.1)	15.7 (8.8, 28.0)	-2.0 (-11%)	-3.6	<0.01	0.2
<b>KNEE</b>						
Contact angle (°)	15.5 (12.2, 19.7)	16.9 (13.5, 20.7)	1.4 (9%)	-6.5	<0.01	0.3
Joint excursion (°)	11.3 (10.2, 12.9)	11.0 (9.7, 13.2)	-0.3 (3%)	-0.2	0.82	0.0
Joint stiffness (Nm/°/kg)	2.1 (0.9, 2.5)	2.0 (1.3, 2.5)	-0.1 (-7%)	-0.6	0.53	0.0
<b>ANKLE</b>						
Foot strike angle (°)	2.0 (-1.9, 4.9)	-0.7 (-3.6, 8.1)	-2.7 (-137%)	-0.9	0.38	0.0
Joint excursion (°)	5.7 (2.7, 9.7)	7.3 (3.6, 9.7)	1.6 (29%)	-5.1	<0.01	0.2
Joint stiffness (Nm/°/kg)	35.9 (10.3, 78.0)	9.9 (4.5, 91.9)	-26.0 (-72%)	-3.7	<0.01	0.2

*BW = body weight; COM = centre of mass; TA-R = peak resultant tibial acceleration*

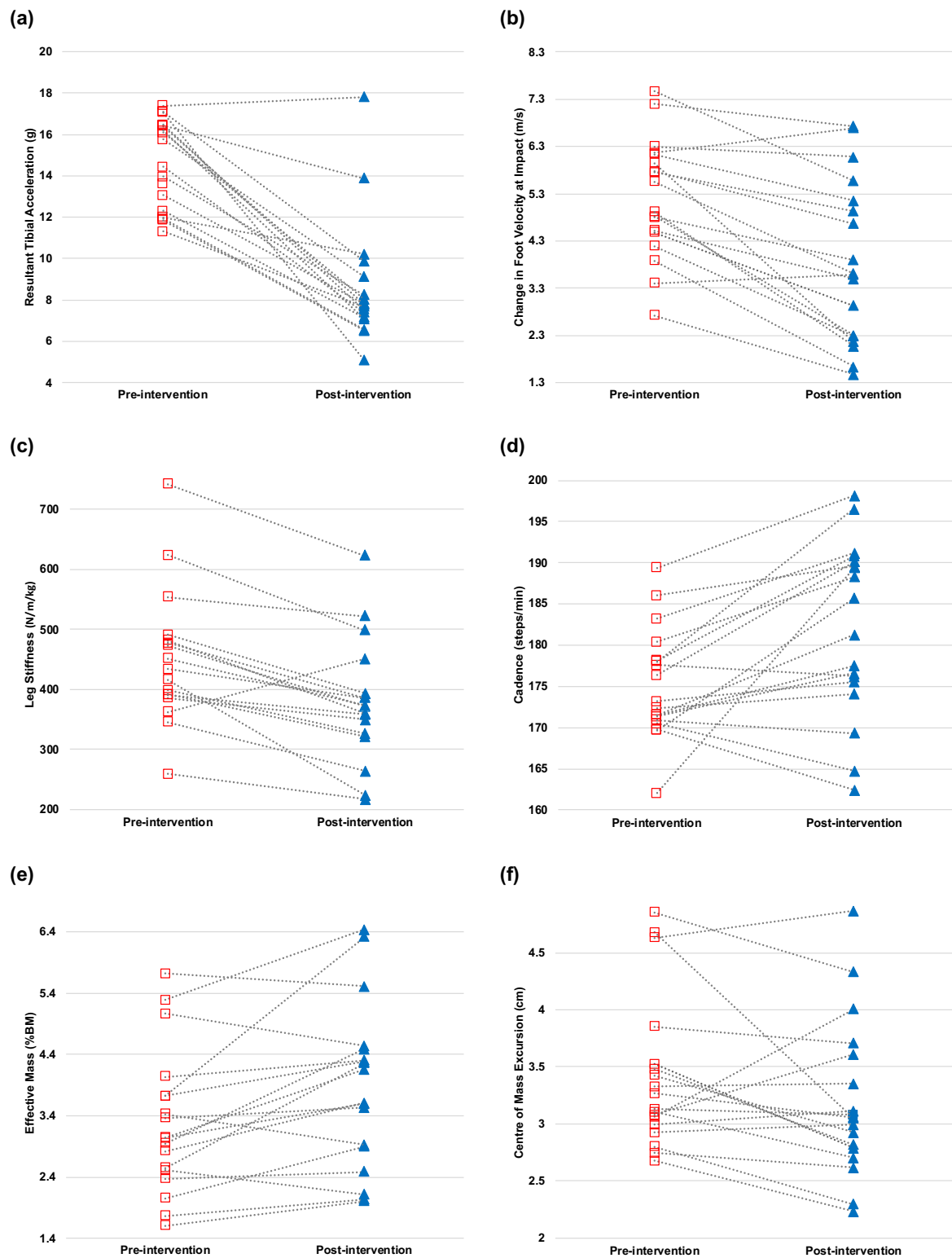
Runners that had a higher cadence at baseline demonstrated a 22% reduction in leg stiffness ( $z=-14.2$ ,  $ES=0.7$ ), a 38% reduction in foot velocity at impact ( $z=-16.0$ ,  $ES=0.8$ ) and a 7% increase in cadence ( $z=-13.4$ ,  $ES=0.8$ ) (Table 15). No consistent changes were demonstrated in secondary biomechanical variables in runners who had a lower baseline cadence. No changes were observed in the contact angle, joint excursion or stiffness at the hip, knee and ankle joints post-intervention for either the high or low cadence sub-groups.

**Table 15: Step rate sub-group analysis of biomechanical variables before and after the gait retraining intervention.**

Variable	Pre-intervention (Md and IQR)	Post-intervention (Md and IQR)	Md Difference (%Diff)	Z	p	ES
<b>Low Cadence (n=8)</b>						
TA-R (g)	14.6 (12.7, 16.1)	8.1 (7.1, 8.9)	-6.5 (-44%)	-12.3	<0.01	0.9
Cadence (steps/min)	170.7 (169.7, 171.5)	177.0 (169.3, 184.6)	6.4 (4%)	-6.6	<0.01	0.5
COM excursion (cm)	3.4 (3.0, 4.5)	3.2 (2.9, 4.0)	-0.2 (-6%)	-3.1	<.001	0.2
Effective mass (%BW)	3.4 (2.6, 4.5)	3.8 (2.7, 5.7)	0.4 (11%)	-6.3	<.001	0.4
Foot impact velocity (m/s)	5.3 (4.4, 6.3)	3.8 (2.8, 5.8)	-1.5 (-28%)	-8.4	<.001	0.6
Leg stiffness (N/m/kg)	416.3 (372.8, 464.2)	373.5 (325.3, 432.6)	-42.8 (-10%)	-8.0	<.001	0.6
<b>High Cadence (n=10)</b>						
TA-R (g)	15.2 (12.3, 16.9)	7.8 (6.7, 10.1)	-7.4 (-49%)	-13.4	<0.01	0.8
Cadence (steps/min)	178.1 (176.3, 183.2)	189.8 (176.2, 191.1)	11.8 (7%)	-13.4	<0.01	0.8
COM excursion (cm)	3.2 (2.9, 3.5)	2.9 (2.5, 3.4)	-0.2 (-7%)	-6.7	<0.01	0.4
Effective mass (%BW)	2.9 (2.4, 3.7)	3.7 (2.9, 4.3)	0.9 (30%)	-9.1	<0.01	0.6
Foot impact velocity (m/s)	5.2 (4.1, 6.1)	3.2 (2.2, 5.2)	-2.0 (-38%)	-13.1	<0.01	0.8
Leg stiffness (N/m/kg)	462.9 (386.9, 546.0)	363.2 (310.2, 474.9)	-99.8 (-22%)	-11.8	<0.01	0.7

*BW = body weight; COM = centre of mass; TA-R = peak resultant tibial acceleration*

Runners who had a non-RFS at baseline demonstrated a 7% increase in cadence ( $z=-11.9$ ,  $ES=0.7$ ), a 40% reduction in foot velocity at impact ( $z=-13.1$ ,  $ES=0.8$ ), whereas those runners that had a RFS at baseline demonstrated a 16% reduction in leg stiffness only ( $z=-10.5$ ,  $ES=0.7$ ) (Table 16). No changes were observed in the contact angle, joint excursion or stiffness at the hip, knee and ankle joints post-intervention for either the RFS or non-RFS sub-groups.



**Figure 16: Individual pre- and post-intervention changes in (a) resultant tibial acceleration (b) change in foot velocity at impact (c) leg stiffness (d) cadence (e) effective mass (f) centre of mass excursion**

**Table 16: Footstrike angle sub-group analysis of biomechanical variables before and after the gait retraining intervention.**

Variable	Pre-intervention (Md and IQR)	Post-intervention (Md and IQR)	Md Difference (%Diff)	Z	p	ES
<b>Non-RFS (n=10)</b>						
TA-R (g)	15.7 (12.8, 16.8)	7.9 (6.8, 9.1)	-7.7 (-49%)	-13.7	<0.01	0.9
Cadence (steps/min)	174.4 (171.5, 180.4)	187.0 (177.5, 190.1)	12.6 (7%)	-11.9	<0.01	0.7
COM excursion (cm)	3.2 (3.0, 3.6)	2.9 (2.6, 3.4)	-0.3 (-11%)	-9.7	<0.01	0.6
Effective mass (%BW)	3.2 (2.8, 4.1)	3.9 (3.3, 4.6)	0.7 (23%)	-8.3	<0.01	-0.5
Foot impact velocity (m/s)	4.4 (3.8, 5.0)	2.6 (1.9, 3.5)	-1.8 (-40%)	-13.1	<0.01	0.8
Leg stiffness (N/m/kg)	400.1 (351.7, 448.9)	344.2 (262.5, 397.3)	-55.9 (-14%)	-9.7	<0.01	0.6
<b>RFS (n=8)</b>						
TA-R (g)	13.9 (12.3, 15.7)	8.1 (6.9, 12.0)	-5.7 (-41%)	-11.9	<0.01	0.8
Cadence (steps/min)	172.2 (170.6, 177.9)	176.4 (170.9, 190.7)	4.2 (2%)	-7.3	<0.01	0.5
COM excursion (cm)	3.3 (2.9, 4.0)	3.2 (2.9, 3.9)	-0.1 (-2%)	-0.4	.671	0.0
Effective mass (%BW)	2.8 (2.2, 3.7)	3.2 (2.3, 4.5)	0.4 (16%)	-7.3	<.001	0.5
Foot impact velocity (m/s)	6.3 (5.7, 6.9)	5.5 (4.7, 6.3)	-0.9 (-14%)	-8.8	<.001	0.6
Leg stiffness (N/m/kg)	517.8 (425.6, 620.3)	435.5 (340.5, 558.7)	-82.3 (-16%)	-10.5	<.001	0.7

*BW = body weight; COM = centre of mass; TA-R = peak resultant tibial acceleration*

### 7.3 Discussion

The goal of the haptic feedback intervention used in this study was to promote the reduction in TA-R in high tibial load runners, and to this end it was largely successful, and associated with a 50% reduction in TA-R across the group, and all but one of the 18 runners demonstrating a positive response post-intervention. There is however a need for enhanced understanding on how changes are executed during the actual execution of running [49,241,242], therefore the primary aim of this study was to determine the self-selected strategies used by runners to achieve a reduction in TA-R. Runners were instructed to reduce their TA-R while running on the treadmill at a fixed velocity, and across the group demonstrated consistent changes in only three of the biomechanical variables measured. Specifically, runners increased their cadence, while simultaneously minimising their foot impact velocity and reducing their leg stiffness during the initial phase of stance. These functional adaptations demonstrate how runners were able to successfully adapt their coordination patterns to achieve defined goals.

With regards to cadence, the findings from this cohort of runners were consistent with what has been reported from other cohorts of high-load runners, where cadence has increased alongside a reduction in vertical force loading rates (both average and instantaneous) irrespective of whether cadence was the target feedback variable, or whether increased cadence was a secondary

outcome of the intervention [203,208]. Centre of mass excursion has also been associated with peak knee extensor moment, peak vertical GRF and braking impulse [273], and therefore a reduction in COM excursion could be a strategy used by runners to minimise TA-R. However, no change in COM excursion was observed from this cohort post-intervention. Inconsistent findings with regard to changes in COM excursion have also been reported in other gait retraining studies, even when a reduction in COM excursion has specifically been targeted [256,261]. Despite not observing any changes in COM excursion, there were consistent reductions in leg stiffness (16%), which have in this cohort come about due to reductions in VIP. It appears that these runners, when allowed to self-select a change in technique, were able to reduce their TA-R and leg stiffness while maintaining a degree of stability in the oscillation of their COM. While speculative, it may be that this a level of self organisation to prevent increases in the physiological cost of running [274].

The reductions in foot velocity at impact were consistent with the findings of Clansey et al. [110] who used a TA based audio-visual feedback approach. However, contradictory to their findings, no change in foot strike angle or ankle joint excursion were found in this cohort of runners. Differences in findings may be due to Clansey et al. [110] specifically selecting runners with a RFS at baseline, and therefore an increase in ankle joint excursion and a change in foot strike angle away from the RFS was likely a logical movement solution from this homogenous group. The evidence linking changes in TA with effective mass is contradictory, with both increases [31,35] and decreases [134] in TA associated with increases in effective mass. Despite observing clear reductions in foot velocity at impact, which is a component of effective mass, no consistent change in effective mass were observed from this cohort of runners. As a result, these findings do not add any further clarity to this relationship.

The secondary aims of this study were to investigate potential alternative strategies used by runners who presented at baseline with a RFS, compared to a non-RFS, as well as runners who were categorised as either low or high cadence at baseline. At the group level, runners demonstrated a 6% increase (11 steps/min) in cadence after undertaking the feedback intervention, which is comparable to the results reported from other retraining studies where runners self-selected their strategy [208,254,256]. The changes observed by high cadence runners were closely representative of the wider group, with a 49% reduction in TA-R and a further 7% increase in cadence, in addition to reductions in foot velocity at impact and leg stiffness. While low cadence runners still demonstrated large reductions in TA-R (44%), the mechanisms of how they achieved these changes are less obvious. In contrast to the high cadence runners, the low cadence sub-group did not demonstrate a consistent increase in cadence, or in any other variable assessed. While these findings appear somewhat counterintuitive, the relatively low subject

numbers and high variability in measures indicate that they are likely not actually homogenous based on low cadence grouping. It does appear that manipulating cadence is a key part of the strategy adopted by some of the runners in this study, with over half increasing their cadence by 5% or more. However, the remainder of runners achieved a reduction in TA-R with minimal change, or in some cases a decrease, in cadence (Figure 16d), indicating that a cadence based feedback intervention, similar that reported by Willy et al. [203] may not have been successful with this sub-group. Further research is required to provide insight into the self-selected retraining strategies of low cadence when presented with haptic feedback.

Runners classified as non-RFS at baseline exhibited a 49% reduction in TA-R, compared to the 41% of RFS runners. These reductions in TA-R were accompanied by an increase in cadence and a decrease in foot velocity at impact in the non-RFS sub-group, but only a reduction in leg stiffness for RFS runners. Surprisingly, irrespective of whether the runners were classified as RFS or non-RFS at baseline, there were no clear changes in foot strike angle after the intervention, whether the entire cohort is analysed together or within the sub-groups classified on foot strike angle at baseline. Based on the results of previous studies [159,208,255,257] it would have been reasonable to think that there would be a clear difference in foot strike angle changes between the two sub-groups, and that the RFS sub-group would adjust their foot strike angle in a way that it would more closely represent the non-RFS sub-group. However, this was not the case for this cohort of runners, as foot strike pattern appeared to have little association with changes in TA-R.

While almost all runners were able to demonstrate consistent reductions in TA-R post-intervention, there was high inter-individual variability observed in how they achieved these changes (Figure 16). These findings provides evidence of the presence of redundancy being present at several different levels in the system, whereby there can be multiple joint configurations and muscle activations to produce the same outcome goal, in this case a reduction in TA-R [275]. Where individuals are not fixed into a rigidly stable solution, but can adapt movement coordination patterns as a function of system degeneracy [252], would generally be perceived as a positive outcome. This point does highlight the difficulties with group-based, or technique-specific running retraining, where runners are directed to specifically change their foot strike pattern, or knee angle at contact [48,261]. It also reinforces the notion that while having a clear picture of the patterns that can occur at a group level, an understanding of individual strategies and motor learning [249] and a data-driven approach to retraining is essential and warranted. Despite the desire for individually specific strategies, few studies have presented individual performance results [83,159,257].

The findings of this study should be considered alongside several limitations. Only a small sample of healthy participants were recruited and hence our findings may not be generalised to injured runners, or those with differing characteristics. It may be that there were biomechanical strategies used by runners that either were not investigated (e.g. trunk inclination, heel to COM distance), or that were masked by the limited statistical power of this study. The study was powered specifically for a change in TA-R, and therefore with this sample size, it is likely that it was underpowered for detecting changes in other variables. Additionally, the biomechanical variables presented were predominantly calculated over the loading phase of stance, to provide close alignment with the time point of peak TA-R, however there is potential that changes outside this interval are important to the strategies adopted. There were also a number of variables that were not controlled in the study, this included the participants' running training outside the feedback protocol, as well as the footwear the runners chose to use during feedback sessions. These factors could have influenced the training, and the translation of the learning effects. Finally, the feedback system used for this study did require a computer, limiting its mobility. With this initial proof of concept, subsequent systems can be developed using mobile technology.

## **7.4 Conclusions**

After completion of a real-time haptic feedback gait retraining intervention, all but one of the high tibial load runners were able to reduce their resultant tibial acceleration. For those working with runners there is a need for a deeper understanding of the mechanics runners use to achieve these reductions. At the group level, runners adopted a higher cadence, while reducing their foot impact velocity and leg stiffness during initial stance. However, there was a high degree of individual variability in the mechanical strategies used, highlighting the need for a personalised, data-driven approach to understanding the response of each runner. Haptic feedback provides a useful tool to facilitate runners self-selecting movement strategies that most suit them as an individual.

## **7.5 Chapter 7 Novel Contributions**

While the success of an intervention is often judged based on the magnitude of changes observed in the target variable, what is also important is understanding how these changes are effected. While the findings in this Chapter are specific to the runners studied and can't necessarily be generalised to others, what I have determined is that when runners are encouraged to self-select the changes in their technique to achieve movement related targets there is large inter-individual variability in how they achieve this. While previous researchers have reported that increased cadence and changing foot strike angle away from a rearfoot strike were important strategies for reducing TA in high load runners, these were not always found to be associated with a reduction in TA in this cohort.



## **Chapter 8:**

### **Discussion and Conclusions.**

#### **8 Context and Framework**

Running is a very popular form sport and exercise. From a global event participation perspective running has grown by 58% (from 5 to 7.9 million participants) over the last 10 years [276]. There is also evidence to show that running can have beneficial impact on cardiovascular function, body composition, as well as the prevention of chronic disease and premature mortality regardless of sex and age [1,277]. The low cost and easy accessibility of running, means it is an activity that potentially promotes physical activity from those that otherwise may not be as active [277]. Runners participating in events are on average getting older, increasing from 35 years of age in 1986, to 39.3 years of age in 2018 [276]. Additionally, female participation has risen from under 20% in 1986 to over 50% in 2018 [276]. Despite the potential health benefits, running is associated with a particularly high annual injury incidence [3,4] ranging between 19% and 79%, depending on the study population and definition of injury [4], most of which are classified as overuse in nature [6]. Injuries inevitably result in time away from running and activity in general, potentially having a subsequent negative impact on an individual's health, as well as a cost to the healthcare system [278]. It has also been shown that, in novice runners at least, sustaining a running injury is the primary reason they cease the activity [279].

Bony fatigue fractures are one of the common injuries sustained in runners, with the tibia being one of the most common sites [6]. The aetiology of TFFs is known to be multi-factorial [11], but from a mechanical perspective, the development of TFFs is due to the accumulation of bone microdamage [15] (Figure 1). The modification of repetitive mechanical load via real-time augmented feedback was used as the theoretical framework for this thesis. Whilst the majority of research conducted with runners to date has been driven by technology that constrained to a laboratory environment [39,82,83,97,110,204,207–214], there has been no research investigating interventions incorporating mobile haptic stimuli as part of an unobstructed and intuitive feedback system aimed at decreasing tibial impact load and the risk of TFFs.

#### **8.0 Summary and Main Findings**

The overarching aim of this thesis was to examine the effectiveness of real-time haptic feedback gait retraining to reduce tibial acceleration in high tibial-load runners. Six studies were conducted to; 1) establish the methodology for the measurement, as well as the processing and interpretation of tibial acceleration data. 2) determine the between-session reliability and variability of resultant

tibial acceleration at different running velocities. 3) investigate the effect of increasing running velocity on resultant tibial acceleration and establish a representative database using this variable. 4) identify runners who might be classified as high-impact based on their resultant tibial acceleration magnitude. 5) to evaluate the evidence for the use of augmented feedback to alter lower extremity loading in runners at risk of tibial fatigue fractures. 6) to undertake an intervention to attempt to reduce tibial acceleration in high tibial-load runners and to determine the self-selected strategies used by runners to achieve a reduction in resultant tibial acceleration. These studies are summarised and the main findings are discussed below.

#### **8.0.1 Chapter 2: The Measurement of Tibial Acceleration in Runners. A Review of the Factors That Can Affect Tibial Acceleration During Running and Evidence-Based Guidelines for its Use.**

Clinicians and researchers commonly use tibial acceleration during running as a proxy measurement for the impact forces experienced at the tibia. There are a number of important considerations that ultimately dictate the accuracy of data collected from runners, as well as a number of contributing factors that influence the interpretation of these metrics. Chapter 2 therefore served as an update of current knowledge of the measurement of tibial acceleration in runners and consolidation the factors that can impact on interpretation.

This chapter provided a comprehensive summary of the current knowledge regarding the use and interpretation of tibial acceleration with runners. What is clear is that for the accurate measurement and interpretation of tibial acceleration, the device must be lightweight, secured firmly to the tibia [79]. The device needs to have an appropriate range and sampling frequency to measure the accelerations associated with running, and the data filtered at an appropriate range (typically between 40 and 100 Hz) [74,105]. It has also been suggested that in order to quantify the total acceleration passing through the musculo-skeletal system all axes of acceleration should be measured [38]. However, further research is needed to understand the importance of data extracted from these additional axes.

A number of gaps in the current evidence were identified. Firstly, evidence to support the between-session reliability of TA in runners is minimal and limited to TA-A [43]. Secondly, tibial acceleration appears to be positively affected by running velocity [108,117,118,126], but data pertaining to axes other than TA-A is extremely limited [38,280], and the relationship between PTA-R and running velocity are not well understood [171,172]. Given runners train at range of paces, being able to interpret TA at across a ranges of velocities is important.

Thirdly, the use of running shoes, as well as their design and material properties, clearly affects tibial acceleration. Specifically, running barefoot produced higher TA magnitudes than in shoes [98,99,103,126]. Tibial acceleration was lower in conventional shoes, compared to shoes with a reduced midsole height [99,103,126], but higher in footwear designed to improve energy return [281] as well as those with lower density mid-soles [101]. Given these variances, it would be necessary to ensure the consistency of footwear when comparing across multiple sessions with a single runner, or between different runners. Finally, it appears likely that other variables, such as joint position, lower extremity stiffness, effective mass and muscle function have a substantial influence on TA [27,120,141,142], but there is still further work required to better understand the nature of the interrelationships between these variables.

### **8.0.2 Chapter 3: The 1-Week and 6-Month Reliability and Variability of Three-Dimensional Tibial Acceleration in Runners.**

As highlighted in Chapter 2, there are few published studies assessing the reliability and variability of TA in runners, and limited to TA-A [43]. If accelerometers are to be used to influence decisions made by coaches and clinicians, an appreciation of the reliability of the devices is required. Natural variability in the TA of runners over months of normal training will also help to establish sample sizes and statistically meaningful differences for future intervention trials. Therefore, the aims of Chapter 3 were to determine between-session reliability and variability of TA-R across both 1-week and 6-month time intervals.

A test-retest experimental experiment was designed, and TA-R was measured using triaxial accelerometers from 14 runners at baseline, and at 1-week, and eight of the runners again at 6-months while they ran on a treadmill at speeds of 2.7, 3.0, 3.3 and 3.7 m/s. CV% and ICCs to assess measurement variability, and MDiff% and ESs to measure reliability. CV% ranged from 6.6% to 11.3% for the 1-week comparison, and 8.7 % to 15.1 % for the 6-month comparison. ICCs ranged from 0.90 to 0.96 for the 1-week comparison, and 0.89 to 0.95 for the 6-month comparison. The MDiff% in TA-R at 1-week ranged from 0% to 3.5%, and from 0% to 5.3% after 6-months, and no ESs exceeded 0.14. Using the criteria outlined by Bradshaw et al. [164] comparisons in TA-R from baseline to 1-week and baseline to 6-months showed 'good' to 'moderate' reliability, and 'small' to 'moderate' measurement variability at all the speeds assessed. It was therefore confirmed that TA-R could be confidently used to monitor runners in a laboratory setting under these conditions.

### **8.0.3 Chapter 4: The Influence of Running Velocity on Resultant Tibial Acceleration in Runners.**

A small number of studies have examined the effects of running velocity on the magnitude of TA. The findings suggest that running at higher velocities is associated with increased PTA-A. However, there is an absence of TA-R data, and the inter-relationships between PTA-R and running velocity are still not well understood. The purpose of this study was to investigate the relationship between TA-R and running velocity, and to establish a normative database of tibial acceleration profiles. As a tertiary aim, we also sought to identify runners who might be classified as high-impact based on their TA-R magnitude at each of the running velocities.

Resultant tibial acceleration was measured from 85 experience uninjured runners as they ran at four velocities ranging from 2.7 to 3.7 m/s. Values ranged from 4.5 g to a maximum of 20.6 g across these velocities. A significant positive correlation was revealed between TA-R and running velocity and a regression analysis revealed that on average, every 0.1 m/s increase in running velocity, TA-R increased by 0.38 g.

One reason for measuring TA has been to screen runners to determine those who may be at risk of injury [35]. Using a one SD upper threshold, for the cohort of runners assessed as part of this study, across the speeds between 17 and 20 participants were identified as having a TA-R result on at least one leg that exceeded the one SD upper threshold, potentially classifying them as high-impact runners. While it is acknowledged that those experiencing high impacts may not necessarily go on to develop injury, this may still be a useful statistical method of identifying potentially high-impact runners.

The majority of runners demonstrated a positive linear trend in TA-R, whereby they experienced fairly consistent increases with each increment in running velocity, there were however a small group of runners that responded in different ways. Some runners demonstrated large changes in their TA-R with each subsequent increase in running velocity. In contrast, some runners actually demonstrated a reduction in their TA-R, from one speed to the next. It is possible that these runners have particular speeds that allow for functionally preferred coordination patterns resulting in lower TA-R, in tune with the theories of dynamical systems theory [174]. These findings highlight the need for a personalised approach to understanding the response of each athlete.

#### **8.0.4 Chapter 5: The Effectiveness of Augmented Feedback to Reduce Lower Extremity Loading in Runners at Risk of Tibial Fatigue Fracture. A Systematic Review.**

A number of literature reviews have provided summaries of the research evidence for the use of gait retraining as an intervention with runners [47–49,160,175], but there have been no reviews specifically targeting the kinetic risk factors associated with the development of TFF in runners. Chapter 5 provided a synthesis of the current evidence for the use of augmented feedback to facilitate running retraining to reduce biomechanical risk factors for developing TFFs and therefore provided the theoretical basis for the intervention applied in Chapters 6 and 7.

There was limited to moderate strength evidence to support moderate to extremely large immediate reductions in kinetic variables related to TFFs [82,83,110,159,203,204,207–214]. There was also limited evidence supporting very large to extremely large short-term reductions in these variables [82,97,110]. Only a single low powered and poor quality study carried out a long-term retention test (3-months post-intervention) [159]. As a result, it is unclear what effects augmented feedback retraining might have after there has been time for changes to be embedded within habitual running.

There was moderate evidence that visual feedback modality can be successfully used with runners [82,110,203,204,211], at least short-term, and limited evidence for use of audio feedback modalities [39,97,159,214]. While haptic feedback has been used in walking gait retraining [60,225], no studies have exclusively used this form of feedback modality with runners. There was moderate evidence to suggest that eight sessions of 15 to 30 minutes of running over a two-week period is an appropriate dose of feedback stimulus to result in short-term changes [82,203,204,211], although no studies have specifically analysed this to provide an optimised dose for individual runners. It appears that a faded withdrawal of feedback is important to avoid overreliance and to encourage internalisation of gait changes [82,159,204,207,211], but once again this has not been specifically researched within the context of running gait retraining.

One of the limitations of Chapter 5 was that the interventions reviewed varied on many levels, with respect to augmented feedback type, targeted variables, instructions and in some cases, supplementary activities or interventions. This hasn't made it possible to determine to what degree changes in runners' performance are due to the augmented feedback itself, or the influence of other factors.

### **8.0.5 Chapter 6: The Effectiveness of Real-Time Haptic Feedback Gait Retraining for Reducing Resultant Tibial Acceleration with Runners.**

As part of Chapter 4 a number of runners were identified as 'high-impact' based on their TA, which has previously been identified as a risk factor for the development of TFF [35]. The aim of this study was to determine the effectiveness of using an adapted real-time haptic feedback system that was previously successfully used to help modify walking gait, to assist runners reduce their TA. A secondary aim was to assess the retention of changes over 4-weeks, and the transfer of learning to overground running.

After screening 48 runners, 23 were identified as having high tibial acceleration at the running velocity selected from four options, which they deemed to be closest to their comfortable 10 Km training pace. Nineteen runners agreed to participate in the gait retraining intervention, with one suffering an unrelated injury after the first session. The remaining 18 runners completed an 8-session feedback intervention over three weeks, with 15 returning for a follow-up assessment after 4-weeks of normal running without feedback. A faded feedback design was used to encourage internalisation of the new running pattern, with the duration of sessions progressively increased from 15- to 30-minutes, and the feedback withdrawn over the final four sessions.

Across the cohort of runners studied, this intervention was associated with a 50% reduction in TA-R at the cessation of the intervention, and a 41% reduction 4-weeks later. The magnitude of these changes are amongst the highest reported when compared similar designed studies that have measured axial TA [39,82,83,97,110]. The reductions in TA-R at the same timepoints during overground running, were still substantial, albeit slightly less than those demonstrated during treadmill running (28% and 17% respectively).

Runners who responded to the intervention were defined as those who achieved the target TA-R threshold during treadmill running after the eight feedback sessions, while also demonstrating a statistically significant, and a likely probable ES ( $>0.7$ ) reduction in TA-R when post-intervention measures were compared to pre-intervention during treadmill and overground. 61% of runners who completed the intervention achieved all these criteria, which is a success rate was similar to that previously reported with a visual feedback intervention [242]. While these findings are positive, it is acknowledged that they were from a relatively small sample of healthy participants, with a short follow-up period. Further work is required to monitor these in a larger number of runners with differing characteristics over a longer period of time while monitoring injury incidence.

#### **8.0.6 Chapter 7: The Biomechanical Effects of Real-Time Haptic Feedback Gait Retraining on Runners with High Tibial Acceleration.**

Chapter 7 outlined the success of a real-time haptic feedback intervention based on TA-R data. To further guide the development of this approach, and to provide information to coaches, clinicians and scientists on what mechanical strategies runners are used during gait retraining, the primary aim of this study was to determine biomechanical strategies used by runners to achieve a reduction in TA-R. It has been identified that many researchers have targeted changes in cadence [203,208,253–257], or modification of foot strike angle [208,255,257] as part of their strategies to reduce lower limb loading variables in runners. Therefore, the secondary aims of this study were to investigate whether there were differences in strategies used between runners who presented at baseline with a RFS or non-RFS landing pattern, as well as runners who were categorised as having either a low or high cadence at baseline.

This study is an examination of additional data collected from the 18 participants who completed the 8-session retraining intervention outlined in Chapter 6. Additional to TA-R data previously presented, prior to, and immediately following the intervention, runners underwent a 3D gait analysis. Kinematic data were collected from an 8-camera Vicon motion analysis system, and this was combined with kinetic data from an instrumented treadmill. The additional variables calculated from these data were chosen due to their contribution to higher TA based on previous studies [27,40,104,110,114,120,127,141,142,203,253,254,256,261], and their potential for use to cue lower TA in a clinical setting [48].

Seventeen of the 18 runners that completed the intervention positively responded to the intervention. Alongside the 50% reduction in TA-R post-intervention previously presented, reductions were seen in both the change in foot velocity at impact (31%) and leg stiffness (16%), as well as increased cadence (6%). The changes in cadence [203,208] and in foot velocity at impact [110] were consistent with what has been reported from similar studies targeting other cohorts of high-load runners. However conversely to other research, no change in foot strike angle or ankle joint excursion were found in this cohort of runners. The differences may be due to differing baseline characteristics, with other researchers specifically targeting RFS runners [110].

When further sub-group analysis was carried out it was determined that the changes observed in high cadence runners were closely representative of the wider group, with a 49% reduction in TA-R and a further 7% increase in cadence, in addition to reductions in foot velocity at impact and leg stiffness. However, the low cadence runners did not demonstrate a consistent change in cadence, or any other variable assessed. Based on previous research, it would have been reasonable to

predict that low-cadence runners would have increased their cadence, and high-cadence runners would have adopted an alternative strategy, however this is clearly not the case for this cohort of runners.

Runners classified as non-RFS at baseline exhibited a 49% reduction in TA-R, compared to the 41% of RFS runners. These reductions in TA-R were accompanied by an increase in cadence and a decrease in foot velocity at impact in the non-RFS sub-group, but only a reduction in leg stiffness for RFS runners. Surprisingly, irrespective of whether the runners were classified as RFS or non-RFS at baseline, there were no clear changes in foot strike angle after the intervention, whether the entire cohort is analysed together or within the sub-groups classified on foot strike angle at baseline. Based on the results of previous studies [159,208,255,257] it would have been reasonable to think that there would be a clear difference in foot strike angle changes between the two sub-groups, and that the RFS sub-group would adjust their foot strike angle in a way that it would more closely represent the non-RFS sub-group. However, this was not the case for this cohort of runners, as foot strike pattern appeared to have little association with changes in TA-R.

While almost all runners were able to demonstrate consistent reductions in TA-R post-intervention, it did appear that when runners were allowed to self-select the changes in their technique, this promoted large inter-individual variability in how runners achieved these changes. Potentially each runner did so in a way that was most effective for them, however this was not specifically assessed. What these findings do highlight is that when working with individual runners to modify their technique, a personalised, data-driven approach is required.

#### **8.0.7 Specific Limitations of the Thesis**

Whilst general limitations of the thesis were outlined in Section 1.3, there are some specific limitations that should be outlined. Firstly, it has been suggested by previous researchers that in order to accurately quantify the total acceleration passing through the musculo-skeletal system, it is important to measure all three components of acceleration [38]. Fundamentally, this is achieved by using a triaxial accelerometer. While each of the three axes could be processed and interpreted separately, calculating the vector sum, or resultant acceleration, incorporates the three orthogonal orientated axes. It is acknowledged that only axial tibial acceleration has demonstrated construct validity with respect to TFF in runners [35], despite this resultant tibial acceleration was the metric computed for the series of research studies presented in this thesis. While this could be considered a potential limitation, this decision was justified as being able provide a single metric that takes into account all elements of acceleration. Additionally, this metric is independent of



accelerometer alignment, it is able to be measured with a higher level of reliability than any single axis [115]. This is an important consideration for the development and validation of a system to be used independently by runners in their normal training environment. This is the ultimate goal for the real-time feedback system used in this thesis.

Secondly, tibial acceleration values were measured at a relatively narrow range of treadmill velocities in the series of studies included in this thesis. The range of running velocities were chosen to reflect a range of speeds runners would normally train at, as well as speeds that have been used in previous studies. However, it is acknowledged that runners do run at other velocities both inside and outside the selected range, and the restricted range is a limitation of this research.

Thirdly, the feedback schedule used in Chapter 7 was based on that originally outlined by Crowell, et al. [82], where feedback is provided during eight sessions of 15 to 30 minutes of running. In line with the wider motor learning literature the feedback is progressively withdrawn or faded out over the final four sessions. There was moderate evidence for this approach to facilitate short-term changes in running gait biomechanics [82,203,204,211], however it is acknowledged that individuals learn in different ways and at different rates. Some runners may have responded more positively had the schedule been shorter, longer, or arranged differently.

Fourthly, a number of published studies have recruited participants who were willing to refrain from running outside of their training sessions in the lab [82,110,204]. Runners who volunteered for the intervention study (Chapter 7) were allowed to continue with their normal training alongside the feedback sessions and were specifically instructed to maintain their typical weekly training regime. While it is acknowledged that this removes an element of control from the experiment, it has also been allowed in other similar research [207].

Finally, the haptic feedback system design used in the intervention study (Chapter 7) was relatively simple in design, both from a hardware and software perspective. This design was a basic adaptation of the technology successfully used in the retraining of walking gait [57] whereby a single vibration pulse was provided immediately after each step that exceeded the predefined tibial acceleration threshold. Outside initial pilot testing with runners, no additional research was conducted to further enhance or optimise the design of the feedback system. One specific limiting feature of the system, which is inherent in many forms of haptic feedback, was that runners could not determine where they were at any given time with reference to the threshold. For example, those runners who exceeded the threshold by only a small degree received the same feedback as those runners who substantially higher.

## 8.1 Future Research Directions

In the light of the findings of this thesis and the acknowledged limitations, several recommendations can be made for future research. The studies carried out in this thesis were largely laboratory based, offering a high level of standardisation and control, but low ecological validity. Given the ultimate goal for the real-time haptic feedback system is to be used independently by runners in their normal training environment, additional validation work is required. It is acknowledged that runners do also run at speeds outside of the relatively narrow range used in these studies, and on a variety of different surfaces. Further work is therefore required to better understand the relationships between tibial acceleration, running speed and surface characteristics, and how these also effect lower extremity stiffness and effective mass. The follow-up timeframe for the gait retraining intervention was relatively short, and the incidence of subsequent injuries were not recorded or analysed. To understand the effect of this haptic feedback intervention on injuries in experienced runners, this study needs to be replicated with a larger cohort, over a longer period of time.

Outside initial pilot testing with runners, no research was conducted to further enhance or optimise the design of the feedback system. One specific limiting feature of the system, which is inherent in many forms of haptic feedback, was that runners could not determine where they were at any given time with reference to the threshold. For example, those runners who exceeded the threshold by only a small degree received the same feedback as those runners who substantially higher. Additionally, exploration of variances to the temporal and spatial parameters of vibration to further enhance the feedback experience (e.g. providing differing feedback related to the relative proximity to the target (scaled to error).

Finally, in sports medicine and biomechanics, practitioners work with individual athletes, and therefore an intervention approach optimised for individual learning styles warrants further exploration. For example, there is substantial evidence to support the 8 session, faded feedback protocol design can be effective, there is no evidence that this was the optimal design for each individual. Superior results may have been achieved with greater or lesser number of sessions, of a shorter or longer duration. Additionally, this research has supported the identification of those that positively responded to the intervention, there is still further work required to improve the identification of non-responders earlier, so that time and effort of clinicians and runners is not needlessly expended, and alternative approaches can be sought.

## **8.2 Conclusions**

The research presented in this thesis demonstrates that a real-time haptic feedback gait retraining system can be used to reduce tibial acceleration in runners. Gait retraining with a haptic feedback system appears to be as effective as more established modalities, such as visual feedback. Haptic feedback offers coaches and sports science practitioners a feedback modality that is ideally suited for use in normal training environments due to being practical and portable, and leaving vision and hearing available for other demands. A large degree of inter-individual variability in biomechanics was observed where runners were afforded the opportunity to self-select their technique to achieve a reduction in tibial acceleration. This reinforces the need for an individually specific, data-driven approach, to running gait retraining. Further work is still required to optimise a haptic feedback interface and schedule, as well as the implementation of feedback in runners' typical training. It is also essential that runners are monitored over time to track the influence of gait modifications on injury incidence.

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## 10 Appendices

## Appendix 1. The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners. Published in Sports Biomechanics.

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### The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners\*

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**ABSTRACT**

Tibial acceleration is a surrogate measure for impact loading and lower limb fatigue injury in runners. Triaxial accelerometers may offer reliable and practical measurement of resultant peak tibial acceleration (PTA). With such potential in mind, this study examined variability and measurement reliability of tibial acceleration in 14 runners at baseline at one week, and eight of the runners again at six months. Triaxial accelerometers were attached to the distal tibiae of runners before they ran on a treadmill for two minutes each, at speeds of 2.7, 3.0, 3.3 and 3.7 m/s in standardised shoes. Resultant PTAs were calculated for each speed and session. Reliability outcomes were presented as percentage change and effect sizes. Variability outcomes included intraclass correlation coefficients and the typical error of the measurement. Smallest worthwhile change and performance/noise ratio were also calculated. While runners demonstrated marginally lower reliability and higher variability over six months, compared to one week, in all cases the measures of reliability and variability were of 'good' to 'moderate' reliability, and 'small' to 'moderate' variability using magnitude-based inferences. We can be confident that resultant PTA can be used with runners to assess and monitor their impacts throughout a six-month intervention.

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### Introduction

Running is a popular form of exercise known to have a positive influence on fitness and health; however, it is also associated with a high incidence of lower extremity overuse injuries (Pratt, 1988; van Gent et al., 2007). When the foot strikes the ground during running, a reaction force is created as the velocity of the foot rapidly decelerates to zero, while the body continues to move forwards. The shock or acceleration due to this impact is transmitted throughout the entire body. Tibial acceleration (TA), measured as peaks in the acceleration time signals, from accelerometers attached to the skin overlying the tibia, are used to represent the acceleration of the bone in runners (Lafortune, Henning, & Valiant, 1995).

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## Appendix 2. The influence of running velocity on resultant tibial acceleration in runners. Published in Sports Biomechanics.

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ARTICLE

### The influence of running velocity on resultant tibial acceleration in runners

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**ABSTRACT**  
Tibial acceleration is a surrogate measure for impact loading and might be useful for identifying lower limb fatigue injury in runners. The resultant tibial acceleration calculated from all three axes of a triaxial accelerometer provides a single metric that is independent of the sensor orientation. The purpose of this study was to investigate the relationship between resultant tibial acceleration and running velocity, and to establish a normative database of tibial acceleration profiles. Triaxial accelerometers were attached to the distal tibiae of 85 runners before they ran on a treadmill for 2 min each, at speeds of 2.7, 3.0, 3.3, and 3.7 m/s. Differences in resultant tibial acceleration were calculated using a one-way ANOVA, and the relationship between tibial acceleration and velocity was determined using a Pearson correlation coefficient and a multiple linear regression analysis. Tibial acceleration increased with higher velocities, with an average increase of 3.8 g (38%) between the slowest and fastest speeds. A moderate correlation was demonstrated between tibial acceleration and running velocity, and 19% of tibial acceleration was explained by velocity. While velocity influences tibial acceleration, individual variances to this relationship exist, highlighting the need for a personalised approach to understanding the response of each runner.

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### Introduction

Running is a popular activity with easy accessibility and relatively low cost; however, endurance running is associated with a high incidence of overuse injuries (Van Mechelen, Hlobil, & Kemper, 1992). The transient impact shock that results when the foot impacts the ground surface is thought to play a role in the pathophysiology of many common running injuries (Beck, 1998; Lake, 2000; Meardon & Derrick, 2014). The axial component of peak tibial acceleration (PTA-A) magnitude has been reported as one of a number of biomechanical variables to discriminate female runners with a history of tibial fatigue fractures from uninjured controls (PTA-A =  $6.5 \pm 3.4$  vs.  $5.5 \pm 2.5$  g, respectively) (Pohl, Mullineaux, Milner, Hamill, & Davis, 2008). Additionally, a PTA-A of 8 g (running velocity of 3.7 m/s) was used as upper threshold to identify

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## Appendix 3. The measurement of tibial acceleration in runners. A review of the factors that can affect tibial acceleration during running and evidence based guidelines for its use. Published in Gait & Posture

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### The measurement of tibial acceleration in runners—A review of the factors that can affect tibial acceleration during running and evidence-based guidelines for its use

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**ABSTRACT**

**Background:** Impact loading in runners, assessed by the measurement of tibial acceleration, has attracted substantial research attention. Due to potential injury links, particularly tibial fatigue fractures, tibial acceleration is also used as a clinical monitoring metric. There are contributing factors and potential limitations that must be considered before widespread implementation.

**Aim:** The objective of this review is to update current knowledge of the measurement of tibial acceleration in runners and to provide recommendations for those intending on using this measurement device in research or clinical practice.

**Methods:** Literature relating to the measurement of tibial acceleration in steady-state running was searched. A narrative approach synthesised the information from papers written in English. A range of literature was identified documenting the selection and placement of accelerometers, the analysis of data, and the effects of intrinsic and extrinsic factors.

**Results and discussion:** Tibial acceleration is a proxy measurement for the impact forces experienced at the tibia commonly used by clinicians and researchers. There is an assumption that this measure is related to bone stress and strain, however this is yet to be proven. Multi-axis devices should be secured firmly to the tibia to limit movement relative to the underlying bone and enable quantification of all components of acceleration. Additional frequency analyses could be useful to provide a more thorough characterisation of the signal.

**Conclusions:** Tibial accelerations are clearly affected by running technique, running velocity, lower extremity stiffness, as well as surface and footwear compliance. The interrelationships between muscle pre-activation and fatigue, stiffness, effective mass and tibial acceleration still require further investigation, as well as how changes in these variables impact on injury risk.

### 1. Introduction

Running is a popular activity, but the high participation rate is accompanied by a high incidence of injuries [1]. The majority of running-related injuries occur in the lower limbs, are chronic in nature, and are related to cumulative loading [2]. The repetitive impacts associated with running is thought to play an important role in the pathophysiology of many common running injuries, especially bony fatigue fractures (commonly termed stress fractures) [3–5]. In runners, between 35% and 49% of all fatigue fractures occur in the tibia [6–9].

While many factors influence bony remodeling and ultimately the manifestation of a fatigue fracture [10], biomechanics dictate the level

of mechanical loading on bone during running [11,12]. When the foot strikes the ground, its velocity decelerates to zero and large ground reaction forces (GRF) are generated [13]. This momentum change produces compressive loading of the lower limbs, and results in an impact shock transmitted through the musculoskeletal system, with local segment peak accelerations occurring at successively later times [14,15]. To minimise damage to proximal structures the shock is attenuated, which is accomplished through an interaction of passive and active mechanisms [16–20]. A failure of the lower extremity muscles to adequately absorb the energy of impact may lead to an over-reliance on passive mechanisms for attenuation [20].

Direct in-vivo measurement of bone strain would be ideal for

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## Appendix 4. The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners. Published in Physical Therapy in Sport

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Original Research

### The effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration with runners



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**Objectives:** To examine the effectiveness of real-time haptic feedback gait retraining for reducing resultant tibial acceleration (TA-R) with runners, the retention of changes over four weeks, and the transfer of learning to overground running.

**Design:** Case control.

**Setting:** Biomechanical laboratory treadmill, and track-based overground, running.

**Participants:** 18 experienced uninjured high tibial acceleration runners.

**Main outcome measures:** TA-R measured while treadmill and overground running assessed at pre-, post- and 4-weeks post-intervention.

**Results:** Across the group, a 50% reduction in TA-R was measured post-intervention (ES: 0.9,  $z = -18.2$ ,  $p < .001$ ), and 41% reduction at 4-weeks (ES: 0.8,  $z = -12.9$ ,  $p < .001$ ) with treadmill running. A 28% reduction (ES: 0.7,  $z = -13.2$ ,  $p < .001$ ), and a 17% reduction in TA-R were measured at these same time points when runners ran overground (ES: 0.7,  $z = -11.2$ ,  $p < .001$ ). All but two runners responded positively to the intervention at the post-intervention assessment. Eleven runners were categorised as positive responders to the intervention at the 4-week post-intervention.

**Conclusions:** Haptic feedback based on TA-R appears to be as effective, but less invasive and expensive, compared to other more established modalities, such as visual feedback. This new approach to movement retraining has the potential to revolutionise the way runners engage in gait retraining.

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#### 1. Introduction

There are clear health and fitness benefits that come from regular participation in running (Hoffman & Hoffman, 2008; Lee et al., 2014), however running is also associated with a higher rate of overuse injury than other aerobic training (Bertola, Sartori, Corrêa, Zotz, & Gomes, 2014; Hauret et al., 2015; van Gent et al., 2007). Bony fatigue fractures (frequently termed stress fractures) are a common running injury (Francis, Whatman, Sheerin, Hume, & Johnson, 2018), with the tibia being the most common site (Burr, 1997; Grimston et al., 1994). The causal mechanism of fatigue fractures is related to an imbalance between mechanical loading and an accumulation of damage to the tissue and the remodeling of the bone (Warden, Davis, & Fredericson, 2014). Various mechanical

loading variables have been correlated to the chronicity and recurrence of tibial fatigue fractures, including surrogates such as tibial acceleration (TA) and instantaneous vertical force loading rates (Milner, Davis, & Hamill, 2006; Milner, Ferber, Pollard, Hamill, & Davis, 2006; Napier, MacLean, Maurer, Taunton, & Hunt, 2018; Warden et al., 2014). Previous studies have demonstrated that with appropriate feedback and guidance, running biomechanics can be successfully altered to reduce biomechanical injury risk factors in relatively short periods of time (Clansey, Hanlon, Wallace, Nevill, & Lake, 2014; Creaby & Smith, 2016; Crowell & Davis, 2011; Noehren, Scholz, & Davis, 2011; Willy et al., 2016). There is also some evidence that these changes can translate to injury reductions, at least in novice runners (Chan et al., 2018). Within the context of changing injury-related biomechanical risk factors, it is not sufficient for runners to demonstrate a change in their running performance; they should be able to consistently reproduce an altered technique in the long-term, such that they have relearned this motor skill (Davis & Fittrell, 2016).

Several gait retraining interventions have used biofeedback

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## **Appendix 5. The reliability and variability of three-dimensional tibial acceleration during running. Presented at the International Society of Biomechanics in Sports Conference**

This appendix comprises the following presentation at the *International Society of Biomechanics in Sports Conference 2016* which is in support of Chapter 3.

### **Reference:**

Sheerin, K., Besier, T., Reid, D., & Hume, P. (2016). The reliability and variability of three-dimensional tibial acceleration during running. *International Society of Biomechanics in Sports Conference*. Tsukuba, Japan.

### **Author contribution:**

Sheerin, K., 85%; Besier, T., 5%; Reid, D., 5%; Hume, P., 5%.

To achieve an accurate and reliable assessment of running gait, facilities with highly equipped motion capture systems and force plates are required. These systems are complex, expensive, require considerable space, and are time consuming for data collection and analyses. The laboratory conditions, and therefore results, are often difficult to apply to real-world sporting environments. Improvements in sensor technology and data analysis techniques have enabled the development of real-time mobile devices capable of measuring meaningful running biomechanics information. Wearable sensors, such as accelerometers, are light-weight, low cost and user friendly, and therefore have the potential to make quantified gait assessment more readily available (Higginson, 2009). Wearable accelerometer sensors can also be used both in the laboratory and during field-based assessments (Sinclair, Hobbs, & Protheroe, 2013). A number of studies have examined the effectiveness of accelerometers compared to in-ground force plates, force instrumented treadmills, digital video and motion analysis systems, in determining running gait variables (Auvinet, Gloria, Renault, & Barrey, 2002; Dufek, Mercer, & Griffin, 2009; Laughton, Davis, & Hamill, 2003; Sinclair et al., 2013). While these studies have provided support for the use of accelerometers, the varied placements of devices, the small selection of running speeds, and the potential change in running kinematics from force plate targeting (Challis, 2001), has left questions. Specifically, if accelerometers are to be used to influence decisions made by coaches and clinicians, an appreciation of the running movement variability at different speeds is one element that is required. In addition, between-session reliability needs to be determined before the devices can be used longitudinally to evaluate the



effects of interventions designed to reduce the risk of injury or improve performance. The purpose of this study was to determine the between-session reliability and variability of 3D tibial acceleration at four running speeds.

**METHODS:** Testing was conducted in a laboratory over two identical sessions, separated by one week. Procedures were approved by institutional human research ethics and participants provided written informed consent. Fourteen male runners, aged  $33.6 \pm 11.6$  years, who were injury free at the time of data collection, volunteered for this study. Runners had been regularly participating in running for  $8.7 \pm 8.1$  years, and ran on average  $30.3 \pm 25.5$  Km in  $4.4 \pm 5.2$  training sessions per week.

**Equipment:** Acceleration data were collected from tri-axial wireless accelerometers (IMeasureU Limited, Auckland), which were attached at the intersection of the middle and distal thirds of the antero-medial aspect of participants' right and left tibia. The y-axis of the device was aligned with the long axis of the tibia using the same technique employed by Sinclair et al., (2013). Data were logged to the onboard memory of the accelerometers at 1000 Hz for the duration on the running trials, and then downloaded after each session for processing. Runners wore standardised neutral running shoes (Asics Kudrow, Kobe, Japan).

**Procedures:** Height, weight and tibia length were measured according to standard ISAK protocols. Running trials were completed on an instrumented treadmill (Bertec, Columbus, OH). Following a 5-minute warm-up and familiarisation at a self-selected pace, runners ran for two minutes at 2.7, 3.0, 3.3 and 3.7 m/s.

**Data processing:** Processing was carried out using a custom Matlab script (Mathworks, MA, USA). At each speed, data were visualised to ensure stabilization in gait patterns after changes in treadmill speed, and a subsequent 50 s trial defined. A fourth order, dual pass 60 Hz Butterworth low-pass filter with a cut-off frequency of 16 Hz was applied and the resultant acceleration was calculated as  $r = (x^2 + y^2 + z^2)^{0.5}$ . Peak tibial accelerations (PTAs) were determined from axial and resultant acceleration data for statistical analysis.

**Data analysis:** Processing was carried out using a custom Matlab script (Mathworks, MA, USA). At each speed, data were visualised to ensure stabilization in gait patterns after changes in treadmill speed, and a subsequent 50 s trial defined. A fourth order, dual pass 60 Hz Butterworth low-pass filter with a cut-off frequency of 16 Hz was applied and the resultant acceleration was calculated as  $r = (x^2 + y^2 + z^2)^{0.5}$ . Peak tibial accelerations (PTAs) were determined from axial and resultant acceleration data for statistical analysis.

Data analysis: Group mean and standard deviations were calculated for the peak axial and resultant PTAs across the four running speeds from sessions one and two. Using the Hopkins

(2015) analysis of reliability, data were log transformed to reduce bias arising from non-uniformity of error. Reliability and variability outcomes were presented as percentage changes. Measurement variability outcomes included intraclass correlation coefficients (ICC) and the typical error of the measurement expressed as a coefficient of variation percentage (CV%). An ICC <0.70 is indicative of 'poor' agreement and high measurement variability,  $0.7 \leq \text{ICC} \leq 0.80$  represents a questionable outcome, and ICC >0.8 represents an excellent outcome (Atkinson & Nevill, 1998; Cormack, Newton, McGuigan, & Doyle, 2008; Hopkins, 2015). A CV of <10% is considered small variation. Between session reliability measures included percentage differences in the means (MDiff%) and Cohen's effect sizes (ES). Effect sizes were interpreted as trivial (0.0-0.1), small (0.11-0.3), moderate (0.31-0.5), large (0.51-0.7), very large (0.71-0.9), or extremely large (0.91-1.0) (Hopkins, Marshall, Batterham, & Hanin, 2009).

An interpretation of the average measurement variability and reliability was based on the methods of Bradshaw, Hume, Carlton and Aisbett (2010), where average measurement variability was interpreted as 'small' when the ICC was >0.70 and the CV was <10%, 'moderate' when ICC was <0.70 or CV was >10%, and 'large' when ICC <0.70 and CV >10%. Average reliability was interpreted as 'good' when the difference in the mean was less than 5% and the ES was trivial to small. Average reliability was interpreted as 'moderate' when the aforementioned criteria for 'good' were breached for either the MDiff% or the ES (MDiff >5% or ES = moderate to large). Average reliability was categorised as 'poor' when both the MDiff% and the ES criteria were breached (MDiff >5% and ES = moderate to large).

**RESULTS AND DISCUSSION:** An average of  $61 \pm 1.5$  steps were analysed for the 14 runners for each running speed during each testing session. Descriptive, reliability and variability statistics for axial PTA are presented in Table 1, and for resultant PTA in Table 2. The mean axial PTA values ranged from 5.5 g to 8.1 g, which are consistent with other research, where runners have run at comparable speeds (Abt et al., 2011; Giandolini, Horvais, Farges, Samozino, & Morin, 2013; Hamill, Derrick, & Holt, 1995). The average mean differences in axial PTA between sessions ranged from 0.0 to 0.3 (4.5 to 5.7%). An ES of less than 0.3 is indicative of a minimal change in a variable of interest from one session to the next. No measures of axial PTA exceeded an ES of 0.17.

**Table 1: Axial peak tibial acceleration - between-session variability and reliability**

Speed (m/s)	Left				Right			
	2.7	3.0	3.3	3.7	2.7	3.0	3.3	3.7
Session 1 Mean $\pm$ SD (g)	5.5 $\pm$ 1.8	6.4 $\pm$ 1.9	7.2 $\pm$ 2.1	8.1 $\pm$ 2.5	5.8 $\pm$ 1.7	6.4 $\pm$ 1.7	7.1 $\pm$ 2.0	8.1 $\pm$ 2.2
Session 2 Mean $\pm$ SD (g)	5.6 $\pm$ 2.0	6.7 $\pm$ 1.8	7.0 $\pm$ 2.1	8.0 $\pm$ 2.7	5.8 $\pm$ 1.6	6.2 $\pm$ 1.7	7.1 $\pm$ 2.1	7.8 $\pm$ 2.5
Mdiff (g)	0.1	0.3	-0.2	0.0	0.2	0.1	0.0	-0.3
CV% (90%CL)	16.3 (12.2-25.1)	11.9 (8.9-18.1)	9.8 (7.4-14.9)	12.3 (9.2-18.8)	17.9 (13.4-27.7)	7.5 (5.7-11.3)	10.6 (8.0-16.2)	9.2 (7.0-14.0)
ICC (90%CL)	0.83 (0.61-0.93)	0.88 (0.72-0.95)	0.92 (0.79-0.97)	0.91 (0.78-0.96)	0.73 (0.43-0.89)	0.95 (0.87-0.98)	0.91 (0.78-0.96)	0.93 (0.84-0.97)
Variability rating	Moderate	Moderate	Small	Moderate	Moderate	Small	Moderate	Small
Mdiff% (90%CL)	0.0 (-9.6-10.6)	5.7 (-2.0-13.9)	-2.2 (-8.1-4.2)	-1.1 (-8.5-6.9)	3.7 (-7.1-15.8)	-3.3 (-7.9-1.4)	-0.5 (-7.0-6.4)	-4.5 (-9.9-1.3)
Effect Size	0.05	0.17	0.07	0.01	0.12	0.11	0.11	0.11
Reliability rating	Good	Moderate	Good	Good	Good	Good	Good	Good
Use	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes

While axial PTA (often measured using a uni-axial accelerometer) is the most common variable measured and reported, with the emergence of more accessible tri-axial sensors, the reporting of resultant PTA values is emerging. The advantage of using the resultant PTA compared to the axial PTA is that the resultant PTA is not influenced by the orientation of the device on the limb. As expected, the mean resultant PTAs were all higher in magnitude than axial PTAs, ranging from 7.6 to 12.1 g, due to the resultant value incorporating the additional two axes (x and z). While the absolute mean difference in PTA between sessions one and two were comparable between the axial and resultant measures (0.0 to 0.3 g), because of the smaller absolute axial PTA magnitudes, the Mdiff% in PTA was higher for the axial direction. This, combined with marginally larger effect sizes at some running speeds, indicated that the reliability of the resultant measures was slightly better.

**Table 2: Resultant peak tibial acceleration - between-session variability and reliability**

Speed (m/s)	Left				Right			
	2.7	3.0	3.3	3.7	2.7	3.0	3.3	3.7
Session 1 Mean $\pm$ SD (g)	7.6 $\pm$ 3.0	9.2 $\pm$ 3.2	10.5 $\pm$ 3.6	11.8 $\pm$ 3.7	7.9 $\pm$ 2.9	9.1 $\pm$ 3.0	10.3 $\pm$ 3.2	12.0 $\pm$ 3.6
Session 2 Mean $\pm$ SD (g)	8.0 $\pm$ 2.7	9.4 $\pm$ 2.9	10.6 $\pm$ 3.3	12.1 $\pm$ 3.8	7.9 $\pm$ 2.4	8.7 $\pm$ 2.5	10.5 $\pm$ 3.0	11.9 $\pm$ 3.4
Mdiff (g)	0.3	0.2	0.1	0.3	0.0	-0.3	0.1	-0.1
CV% (90%CL)	12.7 (9.6-19.5)	6.9 (5.2-10.4)	6.3 (4.8-9.5)	8.7 (6.5-13.1)	10.1 (7.6-15.4)	5.9 (4.5-8.9)	13.9 (10.4-21.3)	6.6 (5.0-10.0)
ICC (90%CL)	0.90 (0.75-0.96)	0.97 (0.92-0.99)	0.97 (0.93-0.88)	0.95 (0.87-0.98)	0.93 (0.82-0.97)	0.97 (0.92-0.99)	0.84 (0.64-0.93)	0.96 (0.91-0.99)
Variability rating	Moderate	Small	Small	Small	Moderate	Small	Moderate	Small
Mdiff% (90%CL)	5.1 (-3.0-13.9)	2.4 (-2.1-7.1)	1.7 (-2.4-5.9)	2.0 (-3.5-7.8)	1.9 (-4.5-8.7)	-2.5 (-6.2-1.3)	1.5 (-7.0-10.7)	-0.9 (-5.1-3.4)
Effect Size	0.12	0.05	0.03	0.08	0.01	0.11	0.05	0.04
Reliability rating	Moderate	Good	Good	Good	Good	Good	Good	Good
Use	Yes	Yes	Yes	Yes	Yes	Yes	Yes	Yes

A CV of 10% or less is considered small in pure test-repeats (Bennell, Crossley, Wrigley, & Nitschke, 1999). For the axial direction, only three from eight of our CV% were greater than 10% (9.2 to 17.9%). The coefficients of variation combined with intra-class correlation coefficients

between 0.73 and 0.95, indicate that axial PTA measures had moderate or small levels of variability at all running speeds. However, when compared with the equivalent resultant PTA variables, in all cases the CVs and ICCs were higher. While it is not possible to determine the cause of this increase in measurement variability in the axial direction, one possible reason is the need to align the y-axis of the accelerometer as closely as possible with the long axis of the tibia. While every effort was made to ensure this was achieved, it is a difficult task that does not need to be considered when using resultant PTAs.

**CONCLUSION:** Tibial acceleration is a useful measure in injury prevention studies in runners. It is therefore important to use a measure of acceleration that shows the least variability across sessions. In all cases the quantitative measures of measurement reliability and variability were of a magnitude indicating 'good' to 'moderate' reliability and 'small' to 'moderate' measurement variability. No bias in the differences in variability or reliability between left and right sides, or between the different running speeds were detected. The results were however superior for resultant over axial measurements. We can be confident that measures of peak resultant tibial acceleration can be used with runners to assess and monitor their impacts throughout an intervention.

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## **Appendix 6. The influence of running velocity on resultant tibial acceleration in uninjured runners. Presented at the Sports Medicine Australia Conference.**

This appendix comprises the following presentation at the *Sports Medicine Australia Conference 2017* which is in support of Chapter 4.

### **Reference:**

Sheerin, K., Reid, D., & Besier, T. (2017). The influence of running velocity on resultant tibial acceleration in uninjured runners. *Sports Medicine Australia Conference*. Langkawi, Malaysia.

Sheerin, K., Reid, D., & Besier, T. (2017). The influence of running velocity on resultant tibial acceleration in uninjured runners. *Journal of Science & Medicine in Sport*, 20(Supplement 3), 85–86. <http://doi.org/10.1016/j.jsams.2017.09.366>

### **Author contribution:**

Sheerin, K., 90%; Besier, T., 5%; Reid, D., 5%

**Background:** Impact loading in runners, assessed by the measurement of tibial acceleration (TA), has attracted substantial research attention. In order to accurately quantify the magnitude of TA, it is important to measure and report all three axes (axial, medio-lateral and anterior-posterior). Calculation of the resultant vector (RTA) provides a single metric that takes into account all components. Peak RTA between 5 and 13 g at speeds between 2.3 and 3.7 m/s have been reported. However, the inter-relationships between RTA and running velocity are still not well understood. The aim of this study was to determine the distribution of RTA in runners and investigate the impact of increasing running velocity.

**Methods:** 85 uninjured runners (age: 39.6 ±9.0 years), who ran on average 45.0 ±23.2 km, 3.9 ±1.5 times weekly, were recruited. Acceleration data were collected at 1000 Hz from tri-axial wireless accelerometers attached bilaterally to the distal aspect of participants' tibia. Runners wore standardised neutral running shoes while they ran on an instrumented treadmill for two minutes at 2.7, 3.0, 3.3 and 3.7 m/s. A fourth order, dual pass 60 Hz Butterworth low-pass filter was applied and the RTA was calculated as  $r = (x^2 + y^2 + z^2)^{0.5}$ . Local peaks in the acceleration related to foot impact were identified. The differences between RTA over the different velocities were determined using repeated measurements of one-way analysis of variance (ANOVA). Pearson correlation coefficients were calculated to establish the strength of the relationships between RTA and

running velocity. Correlation magnitude was interpreted as 0.3 (moderate), 0.5 (large), 0.7 (very large) and 0.9 (extremely large). The level of significance was set at  $p<0.05$ .

**Results:** Mean RTA increased across speeds ( $p<0.05$ ), with an overall increase of 3.7 g (38%) between the slowest and fastest speeds. Resultant TA values were  $9.8 \pm 2.7$  g at 2.7 m/s,  $11.0 \pm 3.0$  g at 3.0 m/s,  $12.1 \pm 3.1$  g at 3.3 m/s and  $13.5 \pm 3.1$  g at 3.7 m/s. A significant moderate correlation was demonstrated between RTA and running velocity ( $r=0.42$ ;  $p<0.05$ ).

**Discussion:** Peak RTA measured from this cohort were in line with those published in the literature. Resultant TA is clearly influenced by running velocity, with peak magnitudes increasing with increasing running velocity, and a moderately strong correlation between these variables. This study provides a basis for further study into impact loading in runners.

## **Appendix 7. Haptic gait retraining is effective at reducing tibial impacts during running. Presented at Sports Medicine New Zealand Conference**

This appendix comprises the following presentation at the *Sports Medicine New Zealand Conference 2019* which is in support of Chapter 6.

### **Reference:**

Sheerin, K., Reid, D., Taylor, D. & Besier, T. (2019). Haptic gait retraining is effective at reducing tibial impacts during running. Sports Medicine New Zealand Conference. Dunedin, NZ.

### **Author contribution:**

Sheerin, K., 92.5%; Reid, D., 2.5%; Taylor, D., 2.5%; Besier, T., 2.5%

**Background:** The development of running related injuries is multifactorial, and altered biomechanics are considered one contributor. Therefore, it is important to address these as part of injury prevention or rehabilitation strategies. There is growing evidence to support gait retraining to facilitate changes in running technique, which can result in a reduction in both injury risk [1] and injuries [2]. Haptic feedback has been used with some success in walking gait, but use with runners is limited to a single exploratory study [3].

**Aim:** To assess the effectiveness of real-time haptic feedback for runners with high tibial acceleration.

**Methods:** Eighteen runners with high baseline tibial acceleration (TA) participated in a gait retraining programme where they received real-time haptic feedback to reduce their TA over 8-sessions. Runners received a vibration for every step that exceeded a pre-defined TA threshold via a custom wireless feedback system incorporating an accelerometer (SparkFun, CO, USA), Labview software and a haptic feedback watch. Pre- and post-intervention TA data were collected using a tri-axial accelerometer (IMeasureU, Auckland, NZ) mounted to the distal medial tibia, and a sample of 25 consecutive steps were analysed. Group and individual statistics including medians (Md) and interquartile range (IQR) were calculated. The difference between pre- and post-intervention TA were analysed using a Wilcoxon Signed-Rank Test. Probability of Superiority Effect Sizes (ES) were calculated as:  $ES = ZScore/\sqrt{n}$ . ES ranged from 0 to 1, where 0.5 indicated a 50% probability that there was no difference, and 1.0 indicated absolute certainty there is a difference, between pre- and post-intervention TA.



**Results:** At the group level, a Wilcoxon Signed-Rank Test revealed a reduction in TA following the intervention ( $z=-18.24$ ,  $p<.001$ ), with an ES indicating almost certainty ( $ES=0.9$ ). The median score in TA decreased by 47% ( $Md=15.0$ ,  $12.6-16.6g$  to  $Md=7.9$ ,  $6.9-9.2g$ ). One runner demonstrated no response to the intervention, with a 2.4% increase in TA post-intervention ( $ES=0.5$ ).

**Discussion and Conclusion:** Group-based data indicated that an 8-session haptic feedback intervention was effective to reduce TA in runners who have previously demonstrated high values. This provides impetus for the development of wearable gait retraining systems that could be used in a normal training environment. However, individual analysis revealed one runner did not respond to the intervention. Further research is being performed to better understand the biomechanical changes following the intervention, including the non-responder.

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## Appendix 8. Agreement For Use of Shared Data: Chapter 3



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24<sup>th</sup> September 2015

**RE: Use of shared data for the PhD thesis of Kelly Sheerin.**

Kelly is enrolled in a PhD at AUT University. As part of this PhD we would like to include some data from a study that was collected collaboratively with Mr. Alex Wilson from June to December 2015, in the SPRINZ Labs at AUT. The project aim is to analyze the tibial acceleration, joint kinematics and kinetics of runners.

This written agreement is to outline the sharing of recruitment of subjects, data collection using the same protocol (walk at three speeds and running at five speeds) and data processing.

Alex will be acknowledged in my thesis papers that relate to the use of the Matlab and OpenSim software platforms that have been used for the data processing. Kelly will be acknowledged in Alex's thesis that relate to the use of gait data collected in the SPRINZ laboratory.

Sincerely

Ass. Prof. Duncan Reid  
Primary Supervisor  
Kelly Sheerin

Ass. Prof. Thor Besier  
Secondary Supervisor  
Alex Wilson & Kelly  
Sheerin

Kelly Sheerin  
PhD Candidate

Alex Wilson  
Masters student

## Appendix 9. Ethics approval: Chapters 3 and 4



9 May 2013

Patria Hume  
Faculty of Health and Environmental Sciences

Dear Patria

Re Ethics Application: **13/51 Reliability and comparability of different methods and measures of performance and lower limb biomechanical loading during running.**

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

Your ethics application has been approved for three years until 8 May 2016.

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

- A brief annual progress report using form EA2, which is available online through <http://www.aut.ac.nz/researchethics>. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 8 May 2016;
- A brief report on the status of the project using form EA3, which is available online through <http://www.aut.ac.nz/researchethics>. This report is to be submitted either when the approval expires on 8 May 2016 or on completion of the project.

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

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

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

All the very best with your research,

A handwritten signature in black ink, appearing to read 'Madeline Banda'.

Madeline Banda  
Acting Executive Secretary  
Auckland University of Technology Ethics Committee

Cc: Kelly Sheerin [ksheerin@aut.ac.nz](mailto:ksheerin@aut.ac.nz)

## Appendix 10. Information and Consent Forms: Chapter 3

18 September 2018

page 1 of 4

# Participant Information Sheet



### Date Information Sheet Produced:

21st March 2013

### Project Title

***Reliability and comparability of different methods and measures of performance and lower limb biomechanical loading during running.***

### An Invitation

My name is Kelly Sheerin and I'm a PhD student at AUT University. Together with my supervisors Prof. Patria Hume and Dr Thor Besier, I would like to invite you to assist in a project that aims to compare different methods of measuring performance and lower limb stress during running.

It is entirely your choice as to whether you participate in the project or not. If at any time you decide you no longer want to participate you are free to withdraw from the study without consequences. Your consent to participate in this research will be indicated by your signing and dating the consent form.

Signing the consent form indicates that you have read and understood this information sheet, freely, given your consent to participate, and that there has been no coercion or inducement to participate by the researchers from AUT.

### What is the purpose of this research?

There are currently several different methods of measuring performance and lower limb stress during running. This research aims to compare each of these in terms of how user friendly and reliable they are.

From this research, tests for high-risk athletes will be developed to assist in injury prevention. The results of this research are intended for publication and will contribute to part of my PhD thesis.

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

You have been identified as a current active runner, between the age range of 18-40, who runs at least three times per week, and can complete 10km in under 50 minutes.

### What will happen in this research?

Once you have decided to participate in the study you will be asked to come to the AUT-Millennium Running and Cycling Mechanics Clinic on two separate occasions for about an hour each time.

Firstly, we will ask you about your injury history, before measuring your height and weight. You will then have small reflective markers and match box like measurement devices attached on your right leg and shoe. You will be provided with two different pairs of shoes that you will use for the duration of the testing.

You will firstly warm-up by running on the treadmill at a self-selected pace for 5 mins. You will complete three 8 min blocks of running at four different speeds between 9.7 and 15 Km/h. Information will be collected as you run from our force detecting treadmill, a 3D motion analysis system, and a standard video camera focussed on your lower legs.

Between the first and second running blocks you will change between the two different pair of shoes. Between the second and third running blocks you will complete a calf fatigue protocol. The calf fatigue protocol will be conducted on a machine that is designed to continually assess your force output. Your maximum calf strength will first be assessed, before you complete continuous ankle plantar-dorsiflexion (heel-raise action) repetitions until your force output drops to 50% of their maximum effort. Upon completion of the fatigue protocol, you will immediately perform the third running trial block.

#### **What are the discomforts and risks?**

There should be no significant risk to you during the study. You may experience localised muscle fatigue and soreness toward the end of the test session, or on the following day. It is expected that this soreness will lessen and disappear over the next few days.

#### **How will these discomforts and risks be alleviated?**

If you are experiencing discomfort at any stage you are encouraged to inform the researcher with you at the time in order that they can best address the problem. If you have any questions regarding and risk or comfort that you anticipate, please feel free to address these concerns to the researcher so that you feel comfortable at all times throughout the process.

#### **What are the benefits?**

You will be given a written 1-2 page report with a summary of your results, from which we will be able to provide you with recommendations to improve your performance or reduce your risk of future injury.

The summary results of this research will be submitted for publication and will make up a portion of my PhD thesis.

#### **What compensation is available for injury or negligence?**

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

#### **How will my privacy be protected?**

The only people who will have access to any of the data collected will be myself and my supervisors, Prof. Patria Hume and Dr Thor Besier. If any of your data is published you will be anonymous. Your name will be coded so that all of your data will be stored under the code name. Your privacy and anonymity will be of primary concern when handling the data.

Our 3D motion analysis system does not record visual information, as would a normal camera (they record only the position and movement and of reflective markers). The standard digital video camera will be focused on your lower legs only, and therefore no identifiable features will be recorded during this study.

The data collected will be 3D video and therefore, your face and any other identifying characteristics will not appear.

All data will be stored on password-protected computers or in locked files. Following completion of data analysis your de-identified data will be stored in the SPRINZ ethics storage room indefinitely and made available for future research.

#### **What are the costs of participating in this research?**

There are no costs associated with participating in this research, and each of the two testing sessions will take about an hour.

#### **What opportunity do I have to consider this invitation?**

We would appreciate it if you could let us know within one week whether you would be available to take part in the study or not. After consideration you may withdraw your participation at any time.

**How do I agree to participate in this research?**

If you agree to participate please complete and return the attached consent form.

**Will I receive feedback on the results of this research?**

Yes a written report will be sent to you detailing the results in graphical form as well as specific recommendations for yourself based on your test results.

**Is this research funded?**

Yes, this research study has received financial support from AUT's Faculty of Health and Environmental Studies. The funders have no direct involvement in the design, management, or outcome of the research.

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

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor (contact details below).

Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTEK, Dr Rosemary Godbold, [rosemary.godbold@aut.ac.nz](mailto:rosemary.godbold@aut.ac.nz), 921 9999 ext 6902.

**Whom do I contact for further information about this research?**

***Researcher Contact Details:***

Kelly Sheerin  
Sport Performance Research Institute New Zealand,  
AUT University,  
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***Project Supervisor Contact Details:***

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# Consent Form



**Project title:** *Reliability and comparability of different methods and measures of performance and lower limb biomechanical loading during running.*

**Project Supervisor:** *Professor Patria Hume*

**Researcher:** *Kelly Sheerin*

- ☐ I have read and understood the information provided about this research project in the Information Sheet dated 21st March 2013.
- ☐ I have had an opportunity to ask questions and to have them answered.
- ☐ I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way.
- ☐ I am not suffering from heart disease, high blood pressure, any respiratory condition (mild asthma excluded), any illness or any lower limb injuries (including during the last 3 months) that could impair my physical performance.
- ☐ I agree to take part in this research.
- ☐ I agree to allow my de-identified data to be kept indefinitely and made available for future research.
- ☐ I wish to receive a copy of the report from the research (please tick one): Yes ☐ No ☐

Participant's signature: .....

Participant's name: .....

Participant's Contact Details (if appropriate):

.....  
.....  
.....

Date:

**Approved by the Auckland University of Technology Ethics Committee on 08.04.13 AUTEK Reference number 13/51**

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

## Appendix 11. Ethics approval: Chapters 4, 6 and 7



### AUTEC Secretariat

Auckland University of Technology  
D-89, WA505F Level 5 WA Building City Campus  
T: +64 9 921 9999 ext. 8316  
E: [ethics@aut.ac.nz](mailto:ethics@aut.ac.nz)  
[www.aut.ac.nz/researchethics](http://www.aut.ac.nz/researchethics)

**AUT**

1 September 2015

Duncan Reid  
Faculty of Health and Environmental Sciences

Dear Duncan

Re Ethics Application: **15/181 Running lower leg impact loading and visual feedback to help improve risk in runners.**

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

Your ethics application has been approved for three years until 31 August 2018.

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

- A brief annual progress report using form EA2, which is available online through <http://www.aut.ac.nz/researchethics>. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 31 August 2018;
- A brief report on the status of the project using form EA3, which is available online through <http://www.aut.ac.nz/researchethics>. This report is to be submitted either when the approval expires on 31 August 2018 or on completion of the project.

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

AUTEC grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to obtain this.

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

All the very best with your research,

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

Cc: Kelly Sheerin



## Appendix 12. Information and Consent Forms: Chapter 4

18 September 2018

page 1 of 3

# Participant Information Sheet



### Date Information Sheet Produced:

25<sup>th</sup> May 2015

### Project Title

**Running lower leg impact loading and visual feedback to help improve technique – Screening.**

This study forms part of my PhD and will have the technical chapter title: Real-time visual gait retraining to reduce tibial acceleration and tibial stress injury risk in runners – Screening study.

### An Invitation

My name is Kelly Sheerin and I'm a PhD student at AUT University. Together with my supervisors I would like to invite you to assist in a project that aims to establish improved screening methods for runners at risk of developing lower limb overuse injuries (i.e. tibial stress injuries).

It is entirely your choice as to whether you participate in the project or not. If at any time you decide you no longer want to participate you are free to withdraw from the study without consequences. Your consent to participate in this research will be indicated by your signing and dating the consent form.

Signing the consent form indicates that you have read and understood this information sheet, freely, given your consent to participate, and that there has been no coercion or inducement to participate by the researchers from AUT.

### What is the purpose of this research?

Tibial acceleration or shock has been identified as contributing to tibial stress injuries. However there is little information on what threshold value for tibial shock, at various running speeds, puts runners at risk of developing tibial stress injuries.

From this research we intend to establish a population range for tibial acceleration at a range of running speeds, and tibial acceleration thresholds that would potentially put runners at risk of developing tibial stress injuries.

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

You have responded to advertisements about the study and have identified yourself as a current active runner, who runs on at least three occasions per week, and for a minimum of 20 km per week.

Please note: Females who are pregnant, and runners who are currently injured (or have been in the last 3 months) in a way that has the potential to modify their running technique will be excluded from this research.

### What will happen in this research?

Once you have decided to participate in the study you will be asked to come to the AUT Millennium Running and Cycling Clinic once for approximately half an hour.

Firstly, we will ask you about your running injury history, then measure your height and weight. You will be provided with a pair of running shoes that you will use for the duration of the testing.



Attachment site for accelerometer

We will attach a matchbox like acceleration measurement device to your lower leg with tape. We will attach small reflective markers to your pelvis and legs with tape so that the movements of your legs can be tracked by our three dimensional motion analysis camera system.

You will then warm-up by running on the treadmill at a self-selected pace for 5 minutes. Finally you will complete 2-minute blocks of running at four different speeds between 9.7 and 15 Km/h, as well as at your 'preferred' training speed.

The acceleration data from your lower legs, and the movement of your legs from the camera information, as well as the impact forces measured by the force plate in the treadmill will enable us to assess your running technique.

#### **What are the discomforts and risks?**

There should be no significant risk to you during the study. Given the speeds that you will run at, and for the short duration, it is not expected that you will experience any muscle soreness or fatigue that would effect any of your training.

#### **How will these discomforts and risks be alleviated?**

If you are experiencing discomfort at any stage you are encouraged to inform the researcher with you at the time in order that they can best address the problem. If you have any questions regarding and risk or comfort that you anticipate, please feel free to address these concerns to the researcher so that you feel comfortable at all times throughout the process.

#### **What are the benefits?**

You will be given a written 1-2 page report with a summary of your results, from which we will be able to provide you with recommendations to improve your performance or reduce your risk of future injury.

The summary results of this research will be submitted for publication and will make up a portion of my PhD thesis.

#### **What compensation is available for injury or negligence?**

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

#### **How will my privacy be protected?**

The only people who will have access to any of the data collected will be myself and my supervisors Associate Prof. Duncan Reid, Associate Prof. Thor Besier and Prof. Denise Taylor. If any of your data are published you will be anonymous. Your name will be coded so that all of your data will be stored under the code name. Your privacy and anonymity will be of primary concern when handling the data.

Our three-dimensional motion analysis system does not record visual information, as would a normal camera (they record only the position and movement of reflective markers). The data collected will be 3D images and therefore, your face and any other identifying characteristics will not appear.

All data will be stored on password-protected computers or in locked files. Following completion of data analysis your de-identified data will be stored in the SPRINZ ethics storage room indefinitely and made available for future research.

#### **What are the costs of participating in this research?**

There are no monetary costs associated with participating in this research, with the only cost being your time. The running-based screening session will be approximately half an hour in duration.

**What opportunity do I have to consider this invitation?**

We would appreciate it if you could let us know within one week whether you would be available to take part in the study or not. After consideration you may withdraw your participation at any time.

**How do I agree to participate in this research?**

If you agree to participate please complete and return the attached consent form.

**Will I receive feedback on the results of this research?**

Yes a written report will be sent to you detailing the results in graphical form as well as specific recommendations for yourself based on your test results.

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

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor (contact details below).

Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTECH, Kate O'Connor, [koconnor@aut.ac.nz](mailto:koconnor@aut.ac.nz), 921 9999 ext 6038.

**Whom do I contact for further information about this research?*****Researcher Contact Details:***

Kelly Sheerin  
AUT Millennium Running and Cycling Clinic  
Sport Performance Research Institute New Zealand  
Auckland University of Technology,  
Ph 921 9999 ext. 7354,  
[ksheerin@aut.ac.nz](mailto:ksheerin@aut.ac.nz)

***Project Supervisor Contact Details:***

Associate Professor Duncan Reid  
Sport Performance Research Institute New Zealand  
Auckland University of Technology  
Ph 921 9999 ext. 7806  
[duncan.reid@aut.ac.nz](mailto:duncan.reid@aut.ac.nz)

***Approved by the Auckland University of Technology Ethics Committee on 15.16.15 AUTECH  
Reference number 15/181***

## Appendix 13. Ethics Amendment: Chapters 6 and 7



### Auckland University of Technology Ethics Committee (AUTEC)

Auckland University of Technology  
D-88, Private Bag 92006, Auckland 1142, NZ  
T: +64 9 921 9999 ext. 8316  
E: [ethics@aut.ac.nz](mailto:ethics@aut.ac.nz)  
[www.aut.ac.nz/researchethics](http://www.aut.ac.nz/researchethics)

11 September 2018

Duncan Reid  
Faculty of Health and Environmental Sciences  
Dear Duncan

Re: Ethics Application: **15/181 Running lower leg impact loading and visual feedback to help improve risk in runners.**

Thank you for your request for approval of amendments to your ethics application.

The amendment to research aim 3 (response to real-time vibration feedback) and the research methodology (modification of feedback modality and modification of retraining schedule) is approved.

An extension for one year is approved.

I remind you of the **Standard Conditions of Approval**.

1. A progress report is due annually on the anniversary of the approval date, using form EA2, which is available online through <http://www.aut.ac.nz/research/researchethics>.
2. A final report is due at the expiration of the approval period, or, upon completion of project, using form EA3, which is available online through <http://www.aut.ac.nz/research/researchethics>.
3. Any amendments to the project must be approved by AUTEC prior to being implemented. Amendments can be requested using the EA2 form: <http://www.aut.ac.nz/research/researchethics>.
4. Any serious or unexpected adverse events must be reported to AUTEC Secretariat as a matter of priority.
5. Any unforeseen events that might affect continued ethical acceptability of the project should also be reported to the AUTEC Secretariat as a matter of priority.

Please quote the application number and title on all future correspondence related to this project.

AUTEC grants ethical approval only. If you require management approval for access for your research from another institution or organisation then you are responsible for obtaining it. If the research is undertaken outside New Zealand, you need to meet all locality legal and ethical obligations and requirements.

For any enquiries please contact [ethics@aut.ac.nz](mailto:ethics@aut.ac.nz)

Yours sincerely,

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

Cc: [ksheerin@aut.ac.nz](mailto:ksheerin@aut.ac.nz)

## Appendix 14. Information and Consent Forms: Chapters 6 and 7



### Participant Information Sheet

**Project title:** *Running lower leg impact loading and vibration feedback to help improve technique*

**Project Supervisor:** *Professor Duncan Reid*

**Researcher:** *Kelly Sheerin*

**Date Information Sheet Produced:** Sept 6th 2018

**Project Title:** Running lower leg impact loading and vibration feedback to help improve technique – Intervention.

*This study forms part of my PhD and will have the technical chapter title: Real-time gait retraining to reduce tibial acceleration and tibial stress injury risk in runners – Intervention study.*

#### An Invitation

My name is Kelly Sheerin and I'm a PhD student at AUT University. Together with my supervisors I would like to invite you to assist in a project that aims to assess the effectiveness of a feedback system designed to alter your running technique.

It is entirely your choice as to whether you participate in the project or not. If at any time you decide you no longer want to participate you are free to withdraw from the study without consequences. Your consent to participate in this research will be indicated by your signing and dating the consent form.

Signing the consent form indicates that you have read and understood this information sheet, freely, given your consent to participate, and that there has been no coercion or inducement to participate by the researchers from AUT.

#### What is the purpose of this research?

Tibial acceleration or shock has been identified as contributing to tibial stress injuries. Tibial acceleration changes with running technique. While there are different feedback methods aimed at modifying running technique, most of them are subjective in nature and involve little scientific backing. Very little research has been conducted into the effectiveness of feedback to change running technique. The retention of changes in technique over a long period of time, or how runners change in response to feedback have not yet been investigated.

We will assess the effectiveness of real-time vibration feedback to reduce tibial acceleration, and to change your running technique, as your current technique has been identified as potentially putting you at risk of sustaining tibial stress injuries.

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

You have been identified, via our tibial acceleration screening study, as a current active runner who either has a recent history of tibial stress injury, or who has been classified based on your tibial acceleration, as potentially having increased risk of developing tibial stress injuries in the future.

Please note: Females who are pregnant, and runners who are currently injured (or have been in the last 3 months) in a way that has the potential to modify their running technique will be excluded from this research.

**What will happen in this research?**

Once you have decided to participate in the study you will be asked to come to the AUT Millennium Running and Cycling Clinic on a number of occasions. In total you would be attending:

- 1 x 60 minute baseline assessment and retraining session (30 minutes each)
- 7 x 30 minute sessions over 2 weeks (running technique feedback training)
- 1 x 60 minute reassessment, 3 months after retraining sessions

During the first two weeks you will undergo running feedback training. For each visit, you will be provided with a pair of running shoes that you will use for the duration of the testing.



*Attachment site for accelerometer*

We will attach a matchbox like acceleration measurement device to your lower leg with tape. We will attach small reflective markers to your pelvis and legs with tape so that the movements of your legs can be tracked by our three dimensional motion analysis camera system.

You will then warm-up by running on the treadmill at your self-selected pace for five minutes.

While then running at your preferred training pace you will receive real-time vibration feedback of your leg impacts from a small bracelet attached to your lower leg which will contain several small vibrating motors (similar to those found in mobile phones). The vibration provided by the bracelet will be controlled by a computer, and will be activated when your leg impacts are deemed too high. You will be asked to run while keeping, or trying to keep, the bracelet from vibrating.

You will be provided verbally with some suggestions on how to achieve this goal, and you will also be encouraged to make your own technique modifications.

While you are running, we will take intermittent measures of your running technique. The acceleration data from your lower legs, and the movement of your legs from the camera information, as well as the impact forces measured by the force plate in the treadmill will enable us to assess your running technique.

Three months after the two-week feedback period you will return to the clinic to undergo a reassessment of your technique.

Over the re-assessment period you will keep a training diary to record your weekly training load (number of sessions per week and duration and intensity of sessions). We would also like you to provide information on any discomfort, pain or injuries that develop over the study period, as well as how easy / comfortable you have found the adoption of your new running technique.

**What are the discomforts and risks?**

While undertaking this study there is the potential for localised muscle or joint discomfort as your body adapts to your changed technique.

**How will these discomforts and risks be alleviated?**

It is expected that this soreness will lessen and disappear over the course of a few days, but if it does not, you are encouraged to contact the researchers to seek advice. If you have any questions regarding and risk or comfort that you anticipate, please feel free to address these concerns to the researcher so that you feel comfortable at all times throughout the process.

**What are the benefits?**

You have been identified as a runner that has either running-related injury in the past, or is potentially at risk of sustaining one in the future. The intervention in place as part of this study is designed to help reduce your risk of future injury.

The summary results of this research will be submitted for publication and will make up a portion of my PhD thesis.

**What compensation is available for injury or negligence?**

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

**How will my privacy be protected?**

The only people who will have access to any of the data collected will be myself and my supervisors Prof. Duncan Reid, Associate Prof. Thor Besier and Prof. Denise Taylor. If any of your data are published you will be anonymous. Your name will be coded so that all of your data will be stored under the code name. Your privacy and anonymity will be of primary concern when handling the data.

Our three-dimensional motion analysis system does not record visual information, as would a normal camera (they record only the position and movement of reflective markers). The data collected will be 3D images and therefore, your face and any other identifying characteristics will not appear.

All data will be stored on password-protected computers or in locked files. Following completion of data analysis your de-identified data will be stored in the SPRINZ ethics storage room indefinitely and made available for future research.

**What are the costs of participating in this research?**

There are no monetary costs associated with participating in this research, the only cost being your time. The initial baseline assessment and retraining session will take approximately 60 minutes, while all seven subsequent retraining sessions will take approximately 30 minutes each. The final reassessment, 3-months after the intervention session will take approximately 60 minutes.

**What opportunity do I have to consider this invitation?**

We would appreciate it if you could let us know within one week whether you would be available to take part in the study or not.

**Opportunity to withdraw data from the study**

- It is reiterated that your participation in this research is completely voluntary. You may withdraw from the study at any time without there being adverse consequences of any kind.
- If you register for the study and complete any number of feedback sessions, we would still like to use your data. However, if you do not want us to use this information please email us and we will delete it, otherwise your data will still be used for the research even if you do not complete all components of the study.

**How do I agree to participate in this research?**

If you agree to participate please complete and return the attached consent form.

**Will I receive feedback on the results of this research?**

Yes a written report will be sent to you detailing the results in graphical form as well as specific recommendations for yourself based on your test results.

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

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor (contact details below).

Concerns regarding the conduct of the research should be notified to the Executive Secretary, AUTEC, Kate O'Connor, [kocconnor@aut.ac.nz](mailto:kocconnor@aut.ac.nz), 921 9999 ext 6038.

**Whom do I contact for further information about this research?****Researcher Contact Details:**

Kelly Sheerin  
Sport Performance Research Institute New Zealand,  
Auckland University of Technology,  
Ph 921 9999 ext. 7354  
[ksheerin@aut.ac.nz](mailto:ksheerin@aut.ac.nz)

**Project Supervisor Contact Details:**

Professor Duncan Reid  
Sport Performance Research Institute New Zealand  
Auckland University of Technology  
Ph 921 9999 ext. 7806  
[duncan.reid@aut.ac.nz](mailto:duncan.reid@aut.ac.nz)

**Approved by the Auckland University of Technology Ethics Committee on 15.16.15 AUTEC Reference number 15/181**  
**Approved by the Auckland University of Technology Ethics Committee on 15.16.15 AUTEC Reference number 15/181**



## Consent Form

*Project title:* **Running lower leg impact loading and vibration feedback to help improve technique**

*Project Supervisor:* **Professor Duncan Reid**

*Researcher:* **Kelly Sheerin**

- ☐ I have read and understood the information provided about this research project in the Information Sheet dated Sept 6th 2018.
- ☐ I have had an opportunity to ask questions and to have them answered.
- ☐ I understand that I may withdraw myself, or any information that I have provided, from this project at any time prior to completion of data collection, without being disadvantaged in any way.
- ☐ I am not pregnant, suffering from heart disease, high blood pressure, any respiratory condition (mild asthma excluded), any illness or any current lower limb injuries that could impair my physical performance.
- ☐ I agree to take part in this research.
- ☐ I agree to allow my de-identified data to be kept indefinitely in the SPRINZ database and be made available for future research.
- ☐ I wish to receive a copy of the report from the research (please tick one): Yes ☐ No ☐

Participant's signature: .....

Participant's name: .....

Participant's Contact Details (if appropriate):

.....  
.....  
.....  
.....

Date:

**Approved by the Auckland University of Technology Ethics Committee on 15.16.15 AUTEK Reference number 15/181**

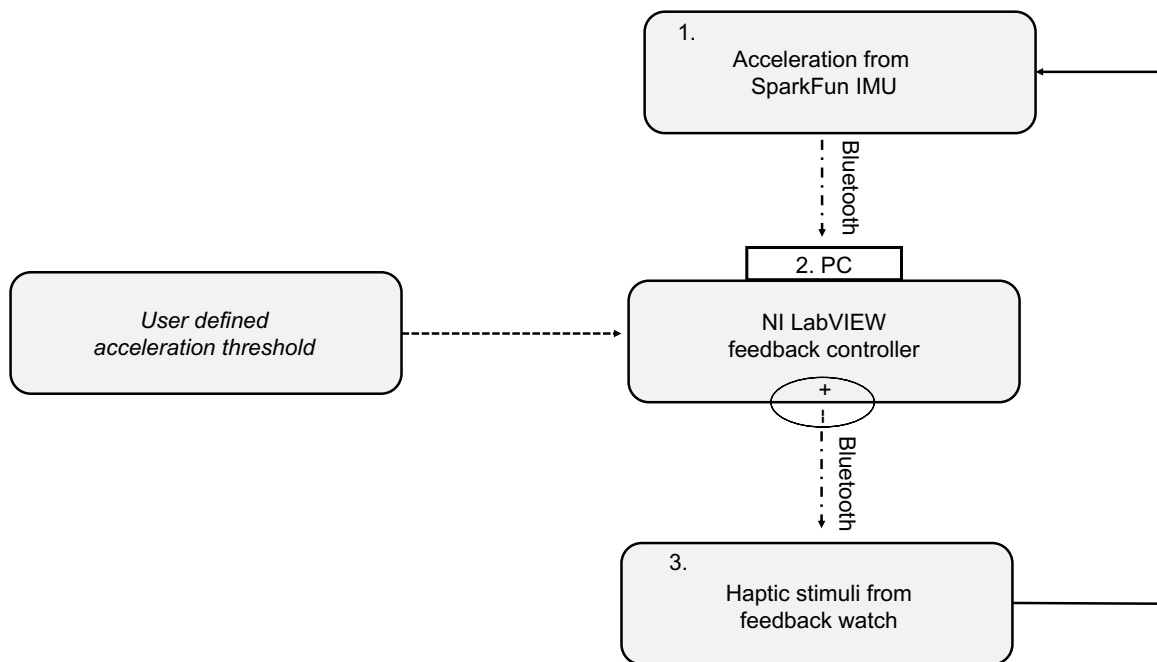
*Note: The Participant should retain a copy of this form.*



## Appendix 15. Real-time haptic feedback system and LabVIEW programme details

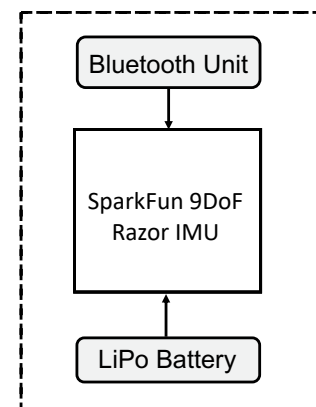
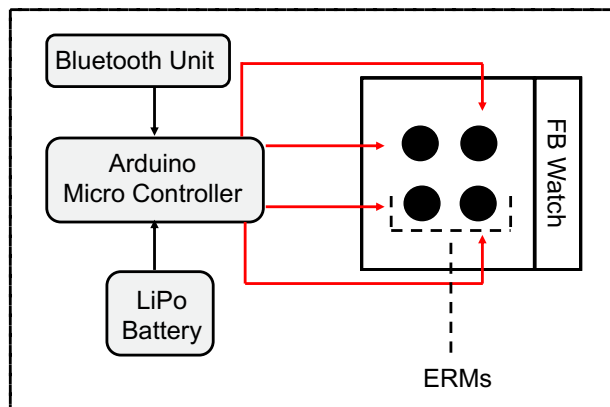
The real-time haptic feedback gait retraining system has three functional components:

1. Sensing user tibial acceleration
2. Processing of raw acceleration data and initiation of feedback
3. Provision of haptic feedback guiding users to alter their running gait patterns.



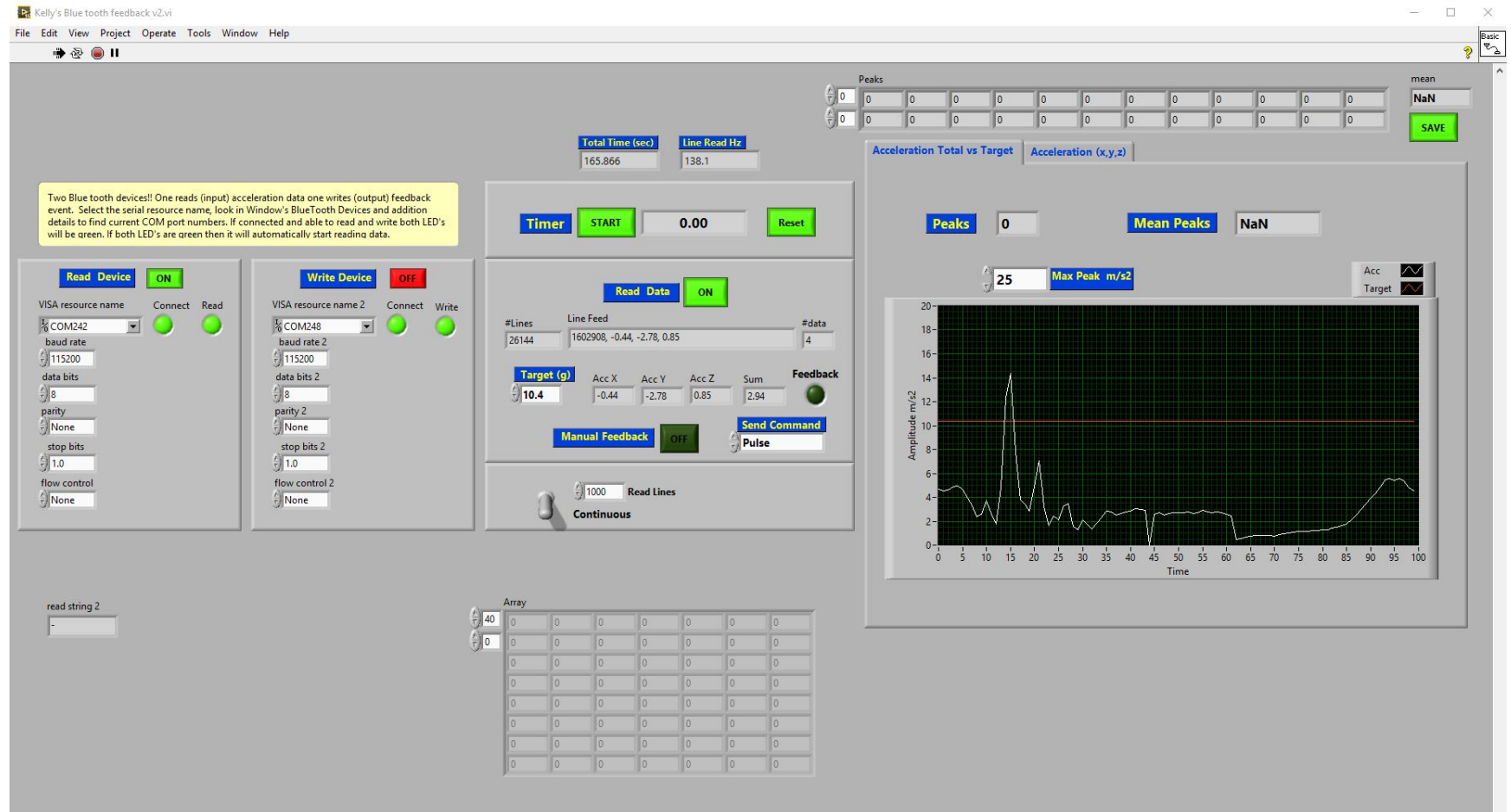
**Block diagram showing the three aspects of the haptic feedback gait retraining system.**

Acceleration measured from a SparkFun IMU sensor is transmitted via Bluetooth to a computer running a custom LabVIEW programme which calculates the resultant acceleration. Where peaks in the resultant acceleration (corresponding to foot-ground impact) exceeded a user defined threshold, an output signal was transmitted via Bluetooth to the custom-made haptic feedback watch. An integrated Arduino MICRO development board served as a controller driving four eccentric rotating mass motors (ERMs) to illicit a vibration. Within the LabVIEW programme was a function to switch off the feedback bracelet, so that users could measure from the accelerometer without providing feedback to runners. The battery and electronics for both the accelerometer and feedback watch were housed in neoprene sleeves.



**Block diagram of the components of the haptic feedback and IMU sensor**

Within the LabVIEW programme was a function to switch off the feedback bracelet, so that users could measure from the accelerometer without providing feedback to runners.



**Screen-shot of real-time feedback LabVIEW software**

## Appendix 16. Chapter 5: Modified Downs and Black score of methodological quality

[illegible]

17. In trials and cohort studies, do the analyses adjust for different lengths of follow-up of patients, or in case-control studies, is the time period between the intervention and outcome the same for cases and controls? (yes =1, no = 0, unable to determine = 0)	1	1	1	1	1	1	1	1	1	1	1	1	0	1	1	1
18. Were the statistical tests used to assess the main outcomes appropriate? (yes =1, no = 0, unable to determine = 0)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	0
19. Was compliance with the intervention/s reliable? (yes =1, no = 0, unable to determine = 0)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1
20. Were the main outcome measures used accurate (valid and reliable)? (yes =1, no = 0, unable to determine = 0)	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1
<b>Internal Validity - Bias Score (x/7)</b>	<b>5</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>4</b>	<b>5</b>	<b>3</b>	<b>4</b>	<b>4</b>	<b>3</b>
<b>INTERNAL VALIDITY – CONFOUNDING (SELECTION BIAS)</b>																
21. Were the patients in different intervention groups (trials and cohort studies) or were the cases and controls (case-control studies) recruited from the same population? (yes =1, no = 0, unable to determine = 0)	1	1	1	1	1	1	1	1	1	1	1	0	1	0	0	1
22. Were study subjects in different intervention groups (trials and cohort studies) or were the cases and controls (case-control studies) recruited over the same period of time? (yes =1, no = 0, unable to determine = 0)	1	1	1	1	1	1	0	0	0	0	0	0	0	0	0	0
23. Were study subjects randomised to intervention groups? (yes =1, no = 0, unable to determine = 0)	1	1	0	0	0	0	0	0	0	0	1	1	0	0	0	0
24. Was the randomised intervention assignment concealed from both patients and health care staff until recruitment was complete and irrevocable? (yes =1, no = 0, unable to determine = 0)	0	1	0	0	0	0	0	0	0	0	0	0	0	0	0	0
25. Was there adequate adjustment for confounding in the analyses from which the main findings were drawn? (yes =1, no = 0, unable to determine = 0)	1	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
26. Were losses of patients to follow-up taken into account? (yes =1, no = 0, unable to determine = 0)	1	0	1	1	1	1	1	1	1	1	0	1	1	1	1	1
<b>Internal Validity – Confounding Score (x/6)</b>	<b>5</b>	<b>4</b>	<b>3</b>	<b>3</b>	<b>3</b>	<b>3</b>	<b>2</b>	<b>2</b>	<b>2</b>	<b>2</b>	<b>2</b>	<b>2</b>	<b>2</b>	<b>1</b>	<b>1</b>	<b>2</b>
<b>POWER (x/1)</b>	<b>1</b>	<b>1</b>	<b>0</b>	<b>1</b>	<b>0</b>	<b>1</b>	<b>0</b>	<b>0</b>	<b>0</b>	<b>0</b>	<b>0</b>	<b>0</b>	<b>0</b>	<b>0</b>	<b>0</b>	<b>0</b>
<b>OVERALL SCORE</b>	<b>21</b>	<b>19</b>	<b>17</b>	<b>17</b>	<b>16</b>	<b>16</b>	<b>15</b>	<b>15</b>	<b>15</b>	<b>15</b>	<b>15</b>	<b>15</b>	<b>14</b>	<b>13</b>	<b>12</b>	<b>12</b>

**Thanks for your interest in my work.**

**Happy running!**