‘FOOTWORK’:  

THE INFLUENCE OF THE  
FOOT’S ARCH AND FOOT STRIKE TECHNIQUE  
ON THE MECHANICS AND ENERGETICS OF RUNNING.  

SARAH MICHELLE STEARNE  B.SC. (HONS)  

THE UNIVERSITY OF WESTERN AUSTRALIA  

THIS THESIS IS PRESENTED FOR THE DEGREE OF DOCTOR OF PHILOSOPHY  

SCHOOL OF SPORT SCIENCE, EXERCISE AND HEALTH  
THE UNIVERSITY OF WESTERN AUSTRALIA  

APRIL 2015
DECLARATION FOR THESES CONTAINING PUBLISHED WORK AND/OR WORK PREPARED FOR PUBLICATION

This thesis contains published work and/or work prepared for publication, some of which has been co-authored. The bibliographical details of the work are presented for each paper. The work involved in designing the studies described in this thesis was performed primarily by Sarah Stearne (candidate). The thesis outline and experimental design was planned and developed by the candidate, in consultation with Dr Jonas Rubenson, Dr Jacqueline Alderson (the candidate’s supervisors), Dr Cyril Donnelly and Mr Ian North. Dr Cyril Donnelly provided advice on the experimental design and manuscript preparation for Study One (included as Chapter Three of this thesis). Mr Ian North provided medical expertise and contributed to the design of Studies Two and Three (included in this thesis as Chapters Four and Five, respectively).

All participant recruitment, implementation of data collection, data processing and data analysis was carried out by the candidate. The candidate drafted the original thesis chapters as well as papers arising from this thesis that have been published or prepared for future publication. Dr Jonas Rubenson and Dr Jacqueline Alderson provided guidance on data collection, data analysis and all drafts associated with the thesis until the examinable version was finalised.

Student Signature …………………………………………………………….
(Sarah Stearne)

Coordinating Supervisors Signature ………………….
(Dr. Jonas Rubenson)
Perhaps not surprisingly, foot biomechanics has received significant attention in human locomotor studies as it affects not only foot function but also whole body biomechanics and energetics. Two key aspects of the foot that play an important role in dictating the mechanics and energetics of running are: 1) foot posture (rearfoot strike [RFS] and forefoot strike [FFS] techniques), and 2) its longitudinal arch mechanics. Despite considerable research in these areas, a detailed knowledge of their effect on joint and lower limb mechanics and metabolic energetics, and the interaction between arch mechanics and foot strike technique does not exist.

Disparities in lower limb biomechanics have been identified between runners using RFS and FFS techniques, yet it remains unclear which, if any, technique offers a mechanical advantage with respect to improved performance or injury reduction. Additionally, whether the joint-level characteristics and any associated benefits can be adopted by changing to an imposed (non-preferred) technique is not clear. One factor that may differentiate foot strike techniques is longitudinal arch mechanics. Compression/recoil of the foot’s arch is hypothesised to lower the metabolic cost of running due to recycling of elastic energy in the arch’s tendons and ligaments. However, no study has specifically assessed this during locomotion. Greater arch compression has been reported in FFS compared with RFS runners, although whether this facilitates an advantage is unknown, nor is it clear, how the arch spring influences other lower limb mechanics such as joint work. The aim of this thesis was to explore these questions, and in doing so establish an integrated understanding of the effect of foot strike technique and the spring function of the arch on running mechanics and energetics.

Study One provides a comprehensive analysis of the lower limb joint kinetics differentiating habitual RFS and FFS runners, as well as the mechanical consequences of switching techniques. Inverse dynamic analysis revealed that neither habitual technique offers a clear mechanical advantage, although variation in negative work distribution may have implications for injury, with RFS and FFS runners at potentially greater risk of knee and ankle injuries, respectively. Total limb positive average power (sum of the ankle, knee and hip rate of mechanical work production) suggests switching to an imposed FFS may be detrimental to performance (increases average power requirement), while switching to an imposed RFS may provide a useful rehabilitation strategy for reducing ankle loading without incurring an overall mechanical penalty.
Abstract

Study Two extended the findings of the first study by employing a novel insole approach to investigate, for the first time during locomotion, the effect of arch compression and recoil on metabolic cost, and the influence of foot strike technique. Each participant was fitted with rigid custom foot insoles which significantly reduced arch compression and ensuing elastic energy storage. During level running, the insoles resulted in a significant increase in gross and net metabolic cost (6.0% and 7.4%, respectively) and a similar estimated (8.7%) loss of mechanical work supplied via elastic arch recoil, whilst overall limb mechanical work remained unchanged. Surprisingly, the energetic effect of reducing arch compression was the same in RFS and FFS runners, despite FFS runners exhibiting greater arch compression during shoe only running. Although the energetic cost of running was affected by restricting arch compression, walking and incline running were not. These results provide among the most direct evidence supporting previous interpretations that the arch spring functions to reduce the energy cost of running and further highlights that this is independent of foot strike technique but not locomotor gait or task (incline).

Study Three aimed to discover how the arch spring affects other lower limb mechanical functions in both RFS and FFS runners. Employing the same novel insole approach, inverse dynamic analysis uncovered differences in joint-level mechanical work when arch compression was restricted during running. Positive mechanical work production at the knee was increased to compensate for the reduction in arch elastic energy storage and return. Negative work at the knee was also increased and both positive and negative work at the ankle decreased. These alterations likely contribute to the increase in metabolic cost observed in Study Two by requiring a larger proportion of the total positive and negative mechanical work to be performed by muscle fibres rather than the usual reliance on the longitudinal arch and Achilles tendon elastic energy storage and return. Alteration of muscle efficiency may also have increased metabolic cost. Foot strike did not influence the compensatory mechanisms employed to overcome the reduced arch compression/recoil despite the clear differences in joint kinetics between techniques (Study One).

Taken together, these findings indicate that neither habitual RFS nor FFS techniques offer a clear joint-level mechanical advantage during running, nor do they effect the arch’s elastic contribution to reducing metabolic cost or the arch’s influence on lower limb mechanics. However, joint-level mechanical differences between running techniques may alter injury predisposition. This thesis quantifies the importance of arch compression to the metabolic cost
of level running and its influence on moderating the mechanical cost of proximal joint musculature. These findings have important implications for understanding fundamental structure-function relationships of the human foot and its evolution, and are of practical importance for optimising orthotic prescription, footwear design and treatments for individuals with impaired arch function.
Anyone who has done a PhD or even just been around someone who has will understand that it can be all consuming. Without your friends and family to bring you to the surface every now and again, it would envelope you. I was lucky enough to have an amazing network of support and I would like to sincerely thank you all, I couldn’t have done it without each and every one of you.

Firstly, I would like to thank my supervisors, Jonas and Jacque.

**Jonas.** Thank you sincerely for everything. You have taught me so much during my PhD journey. Your incredible knowledge, enthusiasm, expertise, passion and patience are inspirational. I hope one day I can live up to the example you have set. Without you none of this would have been possible, so thank you. Please pass on my apologies to Cindy and Leo for all the hours reviewing my thesis has kept you away from them.

**Jacque.** It was your lectures in undergrad that ignited my love of biomechanics. You have provided invaluable advice and insight into my research, and your extensive international network of associates gave me the opportunity to meet some of the world’s most renowned researchers. For everything you have done for me throughout my entire time at SSEH, thank you.

**Nev.** My partner in crime, you have been with me every step of the way and I honestly don’t think I could have done it without you (and not just because of your incredible MatLab skills). You made me laugh with stories of your crazy weekend escapades, awed me with your magic tricks, bought me chocolate when I looked stressed and were there to listen when I needed to vent. You also educated me with your wealth of knowledge about everything biomechanics, MatLab, computers, movies, Game of Thrones….the list goes on. I will miss not seeing you in the office next door but wish you the best of luck wherever your life takes you.

**Ben and Kirsty.** Teaming up with the two of you while you completed your honours degrees was the best thing that could have happened to my PhD. Some people complain that having an honours student around makes their life harder, but without the two of you my PhD testing would not have happened. Ben, with your amazing network of elite running friends, I think we collected some of the best biomechanics data around. I look forward to continuing to work with you at Precision Biomechanics. Kirsty, your biomechanical knowledge, experience from WAIS and enthusiasm were so helpful. Good luck with your PhD and I hope you can team up with an honours student who is as helpful to you as you were to me.

To the past and present members of **Office 1.55** and the rest of the Sports Science **Postgrads.** When I began my PhD, I remember being warned that it was a lonely journey, but thanks to all of you it most certainly was not, in fact, it was quite the opposite so thank you. But perhaps if I had been a bit lonelier my PhD wouldn’t have taken me so long! Your willingness to offer assistance and advice whenever I needed it made my PhD accomplishable. We have shared
lacks of laughs, stories, chocolate, beer, good times and bad, you made my PhD journey so much more than just research. I wish you all the best with your future endeavours and insist that we all keep in touch.

**Ian North, Jon Donnelly, Tony Robey, Charles Oxnard, Rob Day** and everyone else who has contributed to my PhD research. I sincerely appreciate all the time you have volunteered to helping me accomplish my PhD, whether it was providing intellectual advice, inventing magnetic marker sets or spending your weekends measuring my participant’s feet for orthotics.

**My Participants.** Without you my PhD would not have been possible. I made you run covered in tape and markers, in minimal clothing, with rock hard orthotics in your shoes, in a sometimes ridiculously hot and sometimes ridiculously cold lab, whilst breathing into a mouth piece and being jabbed by needles, I cannot thank you enough.

And finally an enormous thank you to my friends and family.

**My Family.** Mum, Dom, Dad, Kev, Linda, Granny and Granddad. I know you used to question what I was doing at Uni when I came home with tales of skipping routines and tennis classes, and you probably still don’t really know what I do all day at Uni. But you encouraged me regardless and 8.5 years later I have finally finished. Your constant support is what got me through, so an enormous thank you to each of you. Mum, the regular coffee breaks, pilates classes and family board game nights were a welcome relief from my computer. Dom, you have been a participant for me more times than you would have liked, but didn’t complain once. Thank you both so much, I can’t put into words how much your support means to me.

**My Friends.** Some of you are doing a PhD, some of you have already finished and the rest of you have heard enough to understand exactly what it is like so know what to say and when to say it. When I wanted to complain, you were there to listen, when I said “no PhD today” you entertained me with tales of your much more exciting lives. Without your encouragement and the constant, but welcome, distractions you provided I might have gone insane these last 4.5 years.

**Brennen.** Without your love and support none of this would have been possible and for that I cannot thank you enough. I know that half of what I’ve said over the last few years has been biomechanics gibberish, but nevertheless you listened attentively and offered support and guidance wherever possible. You were a shoulder for me to cry on (more often than I would like to admit) and celebrated every little win with me. I struggle to put into words how incredibly grateful I am, but I hope you know.
The following is a list of publications and conference presentations arising from my PhD to which I have contributed during the course of my candidature.

**Peer Reviewed Publication**


**Publication Submitted to Journal (for review)**

**Stearne SM, McDonald KA, Alderson JA, North I, Oxnard CE, Rubenson J.** Sole-searching: The foot’s arch and the energetics of human locomotion. *Proceedings of the Royal Society B: Biological Sciences*, submitted September 2014. (Chapter Four of this thesis).

**Publication Ready for Journal Submission**

**Stearne SM, Alderson JA, McDonald KA, North I, Rubenson J.** How do we run without a spring in our step? Alteration in joint mechanics following arch compression restriction. *PLOS ONE*, intended submission September 2014. (Chapter Five of this thesis).
CONFERENCE PRESENTATIONS AND ABSTRACTS


# Table of Contents

**Abstract** ................................................................................................................................. 3

**Acknowledgements** ................................................................................................................ 6

**Publications and Conference Presentations** ........................................................................ 8

**Table of Contents** .................................................................................................................... 10

**List of Tables** ........................................................................................................................ 12

**List of Figures** ......................................................................................................................... 13

**List of Equations** .................................................................................................................... 16

**List of Abbreviations** .............................................................................................................. 17

**Chapter One** ............................................................................................................................ 19

  **Introduction**
    1.2 Statement of the Problem .................................................................................................... 21
    1.3 Thesis Outline .................................................................................................................... 23
    1.4 Limitations and Delimitations .......................................................................................... 26
    1.5 References ....................................................................................................................... 28

**Chapter Two** ........................................................................................................................... 31

  **Review of the Literature**
    2.1 Introduction ...................................................................................................................... 31
    2.2 Anatomy of the Foot ......................................................................................................... 31
    2.3 Evolution of the Foot ........................................................................................................ 33
    2.4 The Elastic Mechanisms of the Arch .............................................................................. 34
    2.5 Altering the Foot’s Arch Function and Elastic Mechanisms ............................................ 36
    2.6 Effect of Foot Strike Technique on Running Mechanics and Energetics ....................... 40
    2.7 Summary ......................................................................................................................... 48
    2.8 References ..................................................................................................................... 49

**Chapter Three** .......................................................................................................................... 57

  **Joint Kinetics in Rearfoot versus Forefoot Running: Implications of Switching Technique**
    3.1 Abstract ............................................................................................................................. 59
    3.2 Introduction ....................................................................................................................... 60
    3.3 Methods ............................................................................................................................. 62
    3.4 Results ............................................................................................................................... 65
    3.5 Discussion ......................................................................................................................... 74
    3.6 Conclusion ......................................................................................................................... 76
    3.7 References ....................................................................................................................... 77
# Table of Contents

## Chapter Four
Sole Searching: The Foot’s Arch and the Energetics of Human Locomotion.

- 4.1 Abstract ........................................................................................................... 83
- 4.2 Introduction ........................................................................................................ 84
- 4.3 Methods ............................................................................................................. 85
- 4.4 Results ............................................................................................................... 93
- 4.5 Discussion .......................................................................................................... 98
- 4.6 Conclusion ......................................................................................................... 104
- 4.7 References ....................................................................................................... 105

## Chapter Five
How Do We Run Without A Spring In Our Step? Alteration In Joint Mechanics Following Arch Compression Restriction

- 5.1 Abstract ............................................................................................................. 111
- 5.2 Introduction ....................................................................................................... 112
- 5.3 Methods .......................................................................................................... 113
- 5.4 Results ............................................................................................................. 116
- 5.5 Discussion ........................................................................................................ 121
- 5.6 Conclusion ....................................................................................................... 125
- 5.7 References ...................................................................................................... 126

## Chapter Six
Synthesis of Findings and Conclusion

- 6.1 Executive Summary .......................................................................................... 129
- 6.2 Synthesis of Findings ..................................................................................... 131
- 6.3 Limitations and Delimitations ......................................................................... 140
- 6.4 Practical Application ....................................................................................... 144
- 6.5 References ...................................................................................................... 145

## Appendices

- Appendix A – Ethics ............................................................................................ 150
- Appendix B – Participant Forms .......................................................................... 154
- Appendix C – Additional Data ........................................................................... 168
- Appendix D – Publications & Media ..................................................................... 175
Table 2.1  Summary of joint kinetic variable comparisons in the literature between habitual and imposed, rearfoot strike and forefoot strike techniques. .................................................................46

Table 3.1  Spatial-temporal gait parameters in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions...............................................................66

Table 3.2  Instantaneous peak moments (N m kg\(^{-1}\)) and positive (produced) and negative (absorbed) peak instantaneous power (W kg\(^{-1}\)) at the ankle, knee and hip joints during stance in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions. ..........................................................................................................................68

Table 3.3  Mechanical work (J kg\(^{-1}\)) and average power (W kg\(^{-1}\)) at the ankle, knee and hip joints and the total lower limb (sum of ankle, knee and hip across stance and swing) in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions. ................................................................................................................................................... 71

Table 3.4  Average moment rate (rate of loading) (N m kg\(^{-1}\) s\(^{-1}\)) at the ankle, knee and hip joints during stance in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions. ............................................................................................................................................................................72

Table 5.1  Lower limb positive and negative mechanical work during shoe only and insole running conditions in rearfoot strike (RFS), forefoot strike (FFS) and both foot strike groups combined (Group). ............................................................................................................................ 118

Table A.1  Questionnaire results. .......................................................................................................................... 170

Table A.2  All data from Study 2........................................................................................................................171

Table A.3  Temporal parameters during walking, level running and incline running across shoe only, half arch insole (HAI) and full arch insole (FAI) in rearfoot strike (RFS) and forefoot strike (FFS) groups. ........................................................................................................................................172

Table A.4  Percentage each joint contributes to positive and negative lower limb mechanical stance work during shoe only and insole running conditions in rearfoot strike (RFS), forefoot strike (FFS) and both foot strike groups combined (Group)...174
LIST OF FIGURES

Figure 2.1 Foot bones and arches, right foot lateral view. *Image adapted from studyblue.com* (56) ... 32

Figure 2.2 Model of difference forces (top) acting on the arch of the foot in forefoot strike (A) and rearfoot strike (B) runners and the time during stance when compression begins to occur (bottom). *Image adapted from Perl et al.* (89) ................................................................. 39

Figure 2.3 Example vertical, anterior-posterior (AP) and medial-lateral (ML) ground reaction forces (GRF) in body weights (BW) for rearfoot strike (black line) and forefoot strike (grey line) techniques during the stance phase of running. Rearfoot strike runners typically experience greater peak vertical impact transients while forefoot strike runners experience greater vertical active peaks. Forefoot strike runners also typically experience greater AP and ML peak forces. ............................................................... 42

Figure 3.1 Group mean data for (A) ankle, knee and hip joint moments and (B) instantaneous powers across the stride (both stance and swing) in habitual forefoot strike (FFS) runners (solid black line), habitual rearfoot strike (RFS) runners (solid grey line), imposed FFS (black broken line) and imposed RFS (grey broken line). Vertical dashed lines represent toe-off in habitual FFS running (black) and habitual RFS runners (grey). ................................................................. 67

Figure 3.2 Distribution of positive and negative work and average power between the ankle, knee and hip joints during the stance phase in habitual rearfoot strike running (RFS), imposed forefoot strike running (FFS), habitual FFS and imposed RFS. (*) indicates significant difference to habitual RFS, (#) indicates significant difference to habitual FFS p< 0.0167. (A) Habitual RFS positive work/average power in stance. (B) Habitual RFS negative work/average power in stance. (C) Imposed FFS positive work/ average power in stance. (D) Imposed FFS negative work/ average power in stance. (E) Habitual FFS positive work/ average power in stance. (F) Habitual FFS negative work/ average power in stance. (G) Imposed RFS positive work/ average power in stance. (H) Imposed RFS negative work/ average power in stance. ............................ 73

Figure 4.1 Right foot medial view illustrating foot marker positions, the sole axis and navicular displacement measure. Windows were cut in the shoe upper which allowed markers to remain visible and the positions unchanged between conditions. ....................................................... 89
Figure 4.2  A) Arch elastic strain energy – ankle compressive load relationship adapted from Ker et al. (27) used to estimate arch strain energy from the participant’s maximum ankle joint compressive load. B) Subject specific load-displacement curve used to predict stored arch elastic energy during different conditions from measured arch compression.

Figure 4.3 Average arch compression (mm) ± standard error when running in the shoe only (light grey line), half arch insole (dark grey line) and full arch insole (black line) throughout the stance phase in level running. Zero indicates arch height at initial contact in the shoe only condition. Positive values indicate a slackened state of the arch elastic structures compared to initial contact in the shoe only condition and negative values indicate stretch compared to the shoe only initial contact.

Figure 4.4 Maximum arch compression (mm) (mean + standard error) relative to arch height at shoe only level running foot contact. * indicates significantly different (p < 0.05) to the shoe only trial in the same condition, † indicates significant difference between the half arch insole (HAI) and full arch insole (FAI) (level running only).

Figure 4.5 Percent change in the rate of oxygen consumption (VO\textsubscript{2}) (mean + standard error) from the shoe only to insole trial across walking, level running and incline running conditions. FAI = full arch insole, HAI = half arch insole. * indicates significant (p < 0.05) difference between raw VO\textsubscript{2} values between the shoe only and insole trial within the same condition.

Figure 4.6  A) Estimated elastic energy (J kg\textsuperscript{-1}) stored in the arch of the foot and B) total limb mechanical work (J kg\textsuperscript{-1}) during shoe only and insole conditions (mean + standard error). * indicates significantly different (p < 0.05) to shoe only trial within the same condition (walk, level run, incline run).

Figure 4.7 Average insole load-displacement curve from six insoles indicating minimal energy return from the insoles.

Figure 5.1 Ankle and knee net joint moments (N m kg\textsuperscript{-1}) during the stance phase in shoe only (solid line) and insole (dashed line) conditions. Data represents the average of rearfoot and forefoot strike groups. * indicates significant difference in peak joint moment between shoe only and insole p < 0.05.

Figure 5.2 Change in positive and B) negative ankle, knee and hip mechanical work (J kg\textsuperscript{-1}) during stance between shoe only and insole conditions. C) Change in total positive stance, swing and stride mechanical work (J kg\textsuperscript{-1}) between shoe only and insole conditions. Positive
values indicate an increase in work when the insole was worn compared to the shoe only condition and negative values indicate a decrease. Values are average ± standard error of rearfoot and forefoot strike groups combined. * indicates significant difference between shoe only and insole (p < 0.05). ..........................................................................................................................120

**Figure 5.3** Ankle (black), knee (grey) and hip (white) joint percentage contribution to total positive (A & B) and negative (C & D) mechanical work during the stance phase in shoe only (A & C) and insole (B & D) conditions. Data represents the average from rearfoot and forefoot strike runners. * indicates significantly different to shoe only (p < 0.05). .................................121

**Figure A.1** Average ankle frontal, knee transverse and hip transverse moments (N kg$^{-1}$) for habitual forefoot strike (FFS) (black solid), habitual rearfoot (RFS) (grey solid), imposed FFS (black dashed) and imposed RFS (grey dashed) conditions for one complete stride (from foot contact to foot contact). Vertical lines represent toe off in RFS (grey) and FFS (black). Positive values indicate ankle inversion and knee and hip internal rotation. Negative values indicate ankle eversion and knee and hip external rotation. No significant differences in any peak moment data exist (p > 0.05). ..........................................................................................................................168

**Figure A.2** Rearfoot inversion/eversion angle across the stance phase of running (from foot contact to toe off) during the shoe only (grey) and Full Arch Insole (FAI) (black) conditions of Chapters Four and Five. Data represents average of forefoot strike and rearfoot strike groups ± one standard deviation. No statistically significant (p < 0.05) difference was observed between peak eversion angles ..........................................................................................................................173
LIST OF EQUATIONS

Equation 3.1 \[ W_{j}^{pos} = \int_{t_i}^{t_f} P_j \, dt \text{ FOR } P_j > 0 \] ...............................................................................64

Equation 3.2 \[ W_{j}^{neg} = \int_{t_i}^{t_f} P_j \, dt \text{ FOR } P_j < 0 \] ...............................................................................64

Equation 4.1 \[ P = \vec{F} \cdot \vec{v}_{COM} = F_x v_{x,COM} + F_y v_{y,COM} + F_z v_{z,COM} \] ...........................................92

Equation 4.2 \[ W_{pos} = \int_{t_i}^{t_f} P \, dt \text{ for } P > 0 \] .............................................................................92

Equation 4.3 \[ W_{neg} = \int_{t_i}^{t_f} P \, dt \text{ for } P < 0 \] .............................................................................92

Equation 5.1 \[ W_{j}^{pos} = \int_{t_i}^{t_f} P_j \, dt \text{ FOR } P_j > 0 \] .............................................................................115

Equation 5.2 \[ W_{j}^{neg} = \int_{t_i}^{t_f} P_j \, dt \text{ FOR } P_j < 0 \] .............................................................................115
# LIST OF ABBREVIATIONS

<table>
<thead>
<tr>
<th>ABBREVIATION</th>
<th>DEFINITION</th>
</tr>
</thead>
<tbody>
<tr>
<td>UWA</td>
<td>University of Western Australia</td>
</tr>
<tr>
<td>RFS</td>
<td>Rearfoot strike</td>
</tr>
<tr>
<td>MFS</td>
<td>Midfoot strike</td>
</tr>
<tr>
<td>FFS</td>
<td>Forefoot strike</td>
</tr>
<tr>
<td>Hab</td>
<td>Habitual</td>
</tr>
<tr>
<td>Imp</td>
<td>Imposed</td>
</tr>
<tr>
<td>HAI</td>
<td>Half arch insole</td>
</tr>
<tr>
<td>FAI</td>
<td>Full arch insole</td>
</tr>
<tr>
<td>COM</td>
<td>Centre of mass</td>
</tr>
<tr>
<td>MTP</td>
<td>metatarsophalangeal</td>
</tr>
<tr>
<td>MTU</td>
<td>Muscle tendon unit</td>
</tr>
<tr>
<td>Pos</td>
<td>Positive</td>
</tr>
<tr>
<td>Neg</td>
<td>Negative</td>
</tr>
<tr>
<td>EMG</td>
<td>Electromyography</td>
</tr>
<tr>
<td>ANOVA</td>
<td>Analysis of variance</td>
</tr>
<tr>
<td>MANOVA</td>
<td>Multiple analysis of variance</td>
</tr>
<tr>
<td>n</td>
<td>Number</td>
</tr>
<tr>
<td>s</td>
<td>Seconds</td>
</tr>
<tr>
<td>min</td>
<td>Minute</td>
</tr>
<tr>
<td>yrs</td>
<td>Years</td>
</tr>
<tr>
<td>mm</td>
<td>Millimetre</td>
</tr>
<tr>
<td>cm</td>
<td>Centimetre</td>
</tr>
<tr>
<td>m</td>
<td>Metre</td>
</tr>
<tr>
<td>m s(^{-1})</td>
<td>Metres per second</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Description</td>
</tr>
<tr>
<td>--------------</td>
<td>-------------</td>
</tr>
<tr>
<td>kg</td>
<td>Kilogram</td>
</tr>
<tr>
<td>BW</td>
<td>Body weights</td>
</tr>
<tr>
<td>SD</td>
<td>Standard deviation</td>
</tr>
<tr>
<td>3D</td>
<td>Three dimensional</td>
</tr>
<tr>
<td>Hz</td>
<td>Hertz</td>
</tr>
<tr>
<td>F</td>
<td>Force</td>
</tr>
<tr>
<td>N</td>
<td>Newtons</td>
</tr>
<tr>
<td>kN</td>
<td>Kilo newtons</td>
</tr>
<tr>
<td>Nm</td>
<td>Newton metres</td>
</tr>
<tr>
<td>GRF</td>
<td>Ground reaction force</td>
</tr>
<tr>
<td>vGRF</td>
<td>Vertical ground reaction force</td>
</tr>
<tr>
<td>AP</td>
<td>Anterior-posterior</td>
</tr>
<tr>
<td>ML</td>
<td>Medial-lateral</td>
</tr>
<tr>
<td>J</td>
<td>Joules</td>
</tr>
<tr>
<td>J kg⁻¹</td>
<td>Joules per kilogram</td>
</tr>
<tr>
<td>(\dot{V}O_2)</td>
<td>Rate of oxygen consumption</td>
</tr>
<tr>
<td>W</td>
<td>Work</td>
</tr>
<tr>
<td>P</td>
<td>Power</td>
</tr>
<tr>
<td>j</td>
<td>Joint</td>
</tr>
<tr>
<td>(t_i)</td>
<td>Time initial</td>
</tr>
<tr>
<td>(t_f)</td>
<td>Time final</td>
</tr>
<tr>
<td>N m kg⁻¹</td>
<td>Newton metres per kilogram</td>
</tr>
<tr>
<td>N m kg⁻¹ s⁻¹</td>
<td>Newton metres per kilogram per second</td>
</tr>
</tbody>
</table>
CHAPTER ONE

INTRODUCTION

1.1 INTRODUCTION

This research will focus on two key aspects of the foot that play an important role in running; the posture of the foot at ground contact (foot strike technique) and the arched anatomy of the foot, as well as the interaction between these factors. The knowledge gained from this research not only has the potential to assist athletes, coaches, clinicians, and footwear designers improve locomotor performance and reduce the running-related injury risk, but provide greater insight into the structure-function relationship of the foot and its evolution.

A runner may adopt any of the following foot strike techniques to make initial ground contact; a rearfoot strike technique where the heel makes initial contact, a midfoot strike when the ball and heel contact simultaneously, or a forefoot strike where initial ground contact is made with the ball of the foot\(^5, 17\). The foot strike technique that runners naturally choose to adopt is known as their habitual technique. It has been proposed by some in the running and scientific communities, that a forefoot strike technique offers performance and injury benefits over a rearfoot strike technique\(^1, 6, 7, 17, 19, 23\). The assumed improvement in performance is largely derived from the greater percentage of elite distance runners who adopt a forefoot strike compared with recreational runners\(^{11, 12}\), however whether this is a function of speed or a superior foot strike technique is unknown. The claim of reduced injury rate has also not been thoroughly substantiated. A recent study by Daoud et al.\(^6\) reported that forefoot strike runners developed fewer injuries than rearfoot strikers, though two larger sample size investigations concluded that injury incidence does not differ between techniques, only injury site\(^{14, 28}\). Although the scientific evidence remains inconclusive, the view that a forefoot strike technique improves performance and reduces injuries has resulted in some coaches recommending their habitually rearfoot striking athletes switch to an imposed forefoot strike\(^1, 23\). However, the ability of a runner to effectively change their foot strike to mimic a habitual technique, and hence adopt any of the purported benefits, is largely unknown. More importantly, it remains unclear which, if any technique offers a performance and/or injury advantage.

Inverse dynamic analysis details the internal forces acting at each joint to produce movement and provides insight into the loading related injury risk. Significant differences in lower limb
moments, powers and mechanical work have been quantified by a number of research groups investigating rear- and forefoot strike kinetics (9, 15, 16, 20, 29, 31). Yet there remain extensive gaps in our knowledge, for example, positive and negative instantaneous power and mechanical work at all lower limb joints are yet to be investigated. Despite the differences in stride frequency observed between foot strike techniques (8, 26), the influence of this on kinetic variables has not been considered. Additionally, the majority of studies have ignored the swing phase in their analyses, thereby preventing differences in total mechanical limb work from being distinguished. Further, few studies have compared habitual rearfoot strike and habitual forefoot strike runners with most recruiting one foot strike group and asking them to perform both techniques. Consequently, these findings may not necessarily reflect variation due to foot strike, but simply the difference between habitual versus imposed techniques. A comprehensive investigation is needed to facilitate our understanding of how lower limb mechanics of habitual rearfoot and habitual forefoot strike techniques differ, such that we can ascertain the merits/disadvantages of each technique and the impact of switching to an imposed foot strike.

In addition to altering lower limb mechanics, it has been proposed that foot strike technique can influence the elastic energy storing function of the arch of the foot (17, 21). During the progression from arboreal life to obligate bipedalism, a number of changes occurred to the structure of the foot. Perhaps one of the most important adaptations was the development of plantar arches (4, 10). Evolutionary biologists hypothesise that the arched foot benefits bipedal locomotion by assisting with impact absorption, stability on varying surfaces, by providing a rigid lever for propulsion during walking, and by acting as a spring to reduce the energy cost of running (2, 3, 10, 13). Simulation testing of the arch spring theory has been performed in cadaveric models and it was confirmed that the arch’s tendons and ligaments are capable of storing and returning significant amounts of elastic energy (13). Extrapolating these results to replicate the forces of running, Ker et al. (13) concluded that the 17 joules of elastic energy returned by the arch of the foot (70 kg male running at 4.5 ms\(^{-1}\) ) equated to \(\sim17\%\) of the total mechanical work and was enough to significantly reduce the metabolic cost of running. This theory is widely accepted, however to date, it has not been tested during locomotion. It therefore remains unclear whether elastic energy stored/returned via compression/recoil of the foot’s arch during locomotion translates to a measurable metabolic advantage, or if there is a difference in the effectiveness of the energy sparing function of the arch between foot strike techniques.
By landing on the ball of the foot during a forefoot strike, the arch immediately experiences a compressive force. Compression is not experienced by a rearfoot strike runner until later in stance when both the heel and ball of the foot are in contact with the ground. For this reason, it has been proposed that a forefoot technique elicits greater elastic energy storage in the arch than a rearfoot strike. Supporting this hypothesis, Perl and colleagues (21) found that the arch of the foot undergoes greater compression during forefoot compared with rearfoot strike running. Although, whether this increased compression facilitates a beneficial metabolic saving is unknown and warrants further investigation.

The influence of arch mechanics on other lower limb joint biomechanics during locomotion is also not well understood. Elastic energy is returned from the arch spring during the propulsive phase of running, potentially reducing the demand on other lower limb structures to produce positive work. However, in which structures mechanical work production is reduced is unknown. By extension, it is also unclear where additional work is produced when elastic energy return from the arch is restricted (e.g. by orthotic insoles). Given the biomechanical differences already identified between rear- and forefoot strike runners, the influence of elastic energy return from the arch of the foot on lower limb mechanics may vary and subsequently different compensatory mechanisms may be employed when compression of the arch is restricted or in conditions of altered arch morphology or mechanics. Such information is not only of practical importance for optimising orthotic prescription, footwear design and treatments for individuals with impaired arch function (18, 24, 25, 27), but is also important for understanding fundamental structure-function relationships of the human foot and its evolution.

1.2 STATEMENT OF THE PROBLEM

Foot biomechanics plays an essential role in running as it not only affects foot function but also dictates overall joint and limb mechanics, which has important implications for performance and injury. Two key aspects of the foot that influence running are the posture at ground contact (foot strike technique), and its unique arched shape. Despite considerable research in these areas, detailed knowledge of their effect on joint and lower limb mechanics and metabolic energetics, and the interaction between foot strike technique and arch mechanics, does not exist.
Recent debate has surrounded rearfoot and forefoot strike running techniques regarding their potential to improve performance and/or reduce injuries. Previous studies have identified mechanical disparities between techniques \((9, 15, 16, 20, 29)\), although there is yet to be a comprehensive analysis of lower limb kinetics in habitual runners. Despite this, many athletes are changing their foot strike technique in search of improvement. However, until a thorough understanding of both techniques is obtained, it is not known which, if any, foot strike technique offers an advantage or whether a runner can adopt an imposed foot strike in order to profit from these possible benefits. Further research in this area is warranted in order to better advise athletes and to avoid potential injury and/or performance decrements associated with technique changes.

Foot strike technique also affects mechanics of the arch of the foot, with greater arch compression recorded in forefoot compared with rearfoot strike runners \((21)\). It is well accepted that the arch of the foot acts as a spring mechanism during running, returning significant passive elastic energy \((13)\). The elastic energy return is estimated to be of a large enough magnitude during running to significantly reduce metabolic cost \((13)\). However, this theory has not been tested during locomotion itself, but rather assumed from \textit{in situ} experiments. It is also unknown whether the greater arch compression observed in fore- vs rearfoot strike runners \((21)\) leads to a level of elastic energy storage that facilitates a metabolic advantage.

It also remains unclear what influence the arch spring has on other lower limb mechanics such as joint work. If the contribution of the arch is as meaningful as hypothesised \((13)\), restricting this source of elastic energy is likely to significantly affect lower limb mechanics as the muscles act to maintain the required work output for locomotion. Given the differences identified in lower limb kinetics between rear- and forefoot strike runners, it is likely that they respond differently when the spring function of the arch is restricted or compromised. Knowledge of the compensatory mechanisms employed, may reduce the risk of overloading lower limb structures and musculature in individuals with impaired arch function.

1.2.1 \textbf{RESEARCH AIMS}

The general aim of this thesis is to establish an integrated understanding of the effect of foot strike technique and the spring function of the arch on running mechanics and energetics. More specifically, this research aims to: 1) gain a comprehensive understanding of the joint-level mechanics that differentiate habitual rearfoot and habitual forefoot strike runners and the consequences of switching to an imposed technique, 2) ascertain whether elastic energy
storage in the arch of the foot reduces the metabolic cost of locomotion and the impact of foot strike, and 3) investigate how the arch spring affects other lower limb joint-level mechanics in both rear- and forefoot strike runners.

1.2.2 Significance of the Research

This thesis serves to broaden our understanding of the foot’s role in running, specifically the influence of foot strike technique and longitudinal arch compression on lower limb mechanics and energetics. This thesis will provide a comprehensive kinetic analysis of rearfoot and forefoot strike techniques and investigate some of the consequences of switching to an imposed foot strike, providing evidence-based research to assist coaches and athletes to make informed decisions regarding the most appropriate technique for optimising performance and avoiding injury. Knowledge of how arch compression influences the metabolic cost of locomotion and lower limb mechanics in both rear- and forefoot strike runners has performance and injury implications, and is of practical importance for optimising orthotic prescription, footwear design and treatments for individuals with impaired arch function. Finally, the results from this work can further our understanding of basic structure-function relationships of the foot and the evolution of the human foot and bipedal mode of locomotion.

1.3 Thesis Outline

This thesis is presented as a series of papers addressing the role of the foot in running. The first study investigates the influence of foot strike technique on lower limb kinetics. Studies Two and Three examine the effect of the foot’s arch mechanics on metabolic and mechanical energetics, respectively. More specifically, Study One explores how the lower limb kinetics of habitual rearfoot strike runners differ from habitual forefoot strike runners. This study also assesses the ability of runners to switch to an imposed foot strike technique. Study Two implements a novel insole approach in an investigation of how compression/recoil of the arch of the foot influences the metabolic cost of locomotion. How the role of the arch and the metabolic cost differs between rearfoot and forefoot strike runners is also examined. Finally, Study Three explores the influence of arch compression/recoil on ankle, knee and hip mechanical work and whether this differs in rearfoot and forefoot strike runners.
1.3.1 CHAPTER TWO – REVIEW OF THE LITERATURE

The literature review builds the context of the thesis through a comprehensive overview of previously published research. It highlights the ‘gaps’ in the current knowledge and provides a theoretical rationale for the research undertaken.

1.3.2 CHAPTER THREE - STUDY ONE

JOINT KINETICS IN REARFOOT VERSUS FOREFOOT RUNNING: IMPLICATIONS OF SWITCHING TECHNIQUE

The aims of this study are to:

- Comprehensively understand the lower limb kinetic differences between habitual rearfoot strike and habitual forefoot strike running techniques.
- Explore the effect of switching to an imposed (non-preferred) foot strike technique on lower limb kinetics.

Research questions are:

- How do the peak joint moments and peak instantaneous joint powers (both positive and negative) at the ankle, knee, and hip differ between habitual rearfoot and habitual forefoot strike runners?
- Does the average moment rate, and the mechanical work and average power generation/absorption (rate of work) differ between habitual rearfoot and habitual forefoot strike runners, both at the individual joints and in the combined lower limb?
- What effect does switching to an imposed foot strike technique have on lower limb mechanics?

1.3.3 CHAPTER FOUR - STUDY TWO

SOLE SEARCHING: THE FOOT’S ARCH AND THE ENERGETICS OF HUMAN LOCOMOTION

The aims of this study are to:

- Determine if restricting compression of the longitudinal arch of the foot (and elastic energy storage in the tendons and ligaments of the arch) influences the metabolic cost of level running.
- Investigate if restricting arch compression near maximally and by approximately 50% alters the impact on the metabolic cost of running.
• Explore the impact of restricting arch compression on the metabolic cost of other forms of locomotion, including walking and incline running.
• Understand the influence of rearfoot and forefoot strike techniques on arch mechanics and the contribution of the arch spring to the metabolic cost of running.

It is hypothesised that:

• When arch compression is restricted during level running there will be an increase in the metabolic cost.
• Metabolic cost will increase by a similar amount when arch compression is restricted maximally and by ~50%, due to the exponential nature of the tendon and ligament compression-elastic energy storage relationship \(^ {13}\).
• The metabolic cost of walking and incline running will be less affected by arch compression restriction compared with level running, due to the lower loads in walking and the greater percentage of positive mechanical work in incline running that is supplied via muscle work as opposed to work supplied from stored elastic energy.
• There will be a greater increase in metabolic cost when arch compression is restricted in forefoot strike compared with rearfoot strike runners.

1.3.4 CHAPTER FIVE - STUDY THREE

HOW DO WE RUN WITHOUT A SPRING IN OUR STEP? ALTERATION IN JOINT MECHANICS FOLLOWING ARCH COMPRESSION RESTRICTION

The aims of this study are to:

• Determine how the mechanics of the lower limb joints adjust to compensate for a reduction in arch compression and thus a reduction in the mechanical work provided by storage and release of arch elastic energy.
• Investigate the compensatory mechanisms employed by rearfoot and forefoot strike runners when arch function is compromised.

It is hypothesised that:

• When arch compression is restricted, both rear- and forefoot strike runners will increase their total joint work production in order to overcome the loss of arch elastic work.
• The ankle joint will contribute the majority of the additional positive mechanical work.
- The increase in positive work at the ankle will be greater in rearfoot compared with forefoot strike runners and instead, forefoot runners will also utilise the hip joint as a source of additional propulsion.

1.3.5 CHAPTER SIX - SYNTHESIS OF FINDINGS AND CONCLUSION

The final chapter aims to provide an overall synthesis of results presented throughout the thesis, integrating the major findings from each study and providing an overall summary of the research scope and highlighting the broader conclusions of the research.

1.3.6 THESIS AS A SERIES OF PAPERS

The University of Western Australia supports the submission of PhD theses that comprise a series of papers prepared for publication. This structure has been adopted by the candidate in the submission of this thesis. As such, while the theoretical linking between the studies (i.e. papers) must be clear for the examiner, each study must be stand-alone in content. Consequently, theses adopting a series of papers approach sometimes result in repetition of methodology from study to study. Please note that where possible reference to previous papers (i.e. previous studies) has been undertaken, however at times the examiner may find some repeated methodology redundant in the course of reading.

1.4 LIMITATIONS AND DELIMITATIONS

1.4.1 LIMITATIONS

The following limitations should be acknowledged when interpreting the results of this research:

Study One Limitations
- The differences in kinetics observed when switching foot strike techniques are acute effects and may diminish with further familiarisation using the imposed technique.
- Foot type (e.g. low arch vs high arch) was not measured or accounted for, which may have introduced kinetic variation into the foot strike groups (30).

Study Two and Three Limitations
- The running speed imposed is slower than many studies of running biomechanics and energetics. However, this speed was carefully selected after pilot testing for a number
of reasons; vertical loads were low enough to prevent excessive insole compression, discomfort caused by the insoles was minimised and it allowed the same speed to be used during level and incline conditions without encroaching on anaerobic metabolism.

- A surface mounted navicular marker was used to estimate arch compression; while this is a commonly used method in the literature it is likely that it does not fully represent internal movement of the arch.

### 1.4.2 DELIMITATIONS

The following delimitations were imposed, limiting the generality of the findings:

#### General Delimitations

- All trials were completed on a treadmill in order to: control speed, allow steady running, impose an incline, and to collect metabolic cost data. This may limit the data’s applicability to over ground running.
- Only healthy male runners aged 18 – 35 years were included and the findings may not be transferable to females, runners of different ages or clinical populations.
- Due to the provision of standardised footwear, only participants with US 10 – 12 size feet were recruited. Therefore, findings may differ in populations with larger or smaller feet.

#### Study One Delimitations

- Running speed was delimited to 4.5 ms\(^{-1}\) which represented a comfortable training speed for participants.

#### Study Two and Three Delimitations

- Treadmill speed was delimited to 2.7 ms\(^{-1}\) during the level and incline running trials and 1.1 ms\(^{-1}\) during walking, which may limit the applicability of the findings to different speeds.
- Only participants with ‘normal’ feet (as determined by an experienced clinician using the Foot Posture Index \(^{(22)}\)) were recruited. This may limit the applicability of the findings to individuals with varied foot form and function.
1.5 REFERENCES


CHAPTER TWO

REVIEW OF THE LITERATURE

2.1 INTRODUCTION

Distance running is one of the fastest growing sports in the world, with participation in the New York City marathon, one of the most revered amateur distance running contests, having doubled in size from 1980 to 2000 (14,012 to 29,930 participants) (81). Further increases have continued, so much so that a record breaking 50,304 runners took part in 2014 (81). Coupled with increased rates of participation is a growing focus, both within the scientific and the recreational and elite running communities, to understand and improve running performance (e.g. the London 2012 Olympic marathon was won by less than 30 seconds (69)) as well as to decrease the high frequency of running related injuries (19.4 - 79.3% (37, 106)). In the pursuit for improved performance and reduced injury risk, foot biomechanics has received significant attention as it plays a primary role influencing not only the foot itself but both joint mechanics and overall locomotor biomechanics and energetics. In particular, two key aspects of the foot that play an important role in dictating the mechanics and energetics of running are 1) its longitudinal arch mechanics, and 2) foot posture at ground contact (foot strike technique) (60, 67).

This literature review commences with a description of the foot’s anatomy and an overview of how its structure has changed with the evolution of bipedal locomotion. One of the key evolutionary adaptions, the longitudinal arch, will then be discussed in the context of running mechanics and energetics. Finally, research regarding the influence of foot strike technique on running mechanics and energetics is reviewed.

2.2 ANATOMY OF THE FOOT

The foot is a complex structure containing 107 ligaments, 33 joints, and 19 muscles and tendons (46). Despite contributing to less than 3% of total body weight, the human feet contain 25% of the body’s bones (26, 46). The bones of the foot are divided into three separate regions; 1) the tarsus (proximal and distal tarsal groups), 2) metatarsus, 3) phalanges (46). The proximal tarsal group consists of: 1) the calcaneus which is the largest and strongest bone of the foot, 2) the talus, the most superior bone of the foot that articulates with the tibia to form the ankle joint, and 3) the navicular bone which is positioned medially and creates the apex of the arch.
The distal group of tarsals include, from medial to lateral, the: medial cuneiform, intermediate cuneiform, lateral cuneiform and the cuboid. Attaching to this group are five metatarsal bones followed by fourteen phalangeal bones which together compose five toes. The first toe, or hallux, contains two phalanges while the other four toes contain three phalanges each (proximal, intermediate and distal) \cite{46, 95} (Figure 2.1).

![Figure 2.1: Foot bones, regions and arches. Right foot lateral view. Image adapted from studyblue.com \cite{56}](image)

The majority of the tarsus bones do not make contact with the ground, instead the bones of the foot are arranged in a series of arches; two longitudinal and one transverse \cite{46, 95} (Figure 2.1). The medial longitudinal arch is the most prominent and is formed by the calcaneus, talus, navicular, cuneiforms, and metatarsals I to III. The lateral longitudinal arch is formed by the calcaneus, cuboid, and metatarsals IV and V. Finally, the transverse arch includes the cuboid, cuneiforms, and the proximal heads of the metatarsals. The integrity of these arches is maintained by a number of tendons and ligaments, the primary being the plantar fascia (or plantar aponeurosis), the plantar calcaneonavicular ligament (or the spring ligament), the long plantar ligament, the short plantar ligament and the tendons of the tibialis posterior, tibialis anterior, flexor hallucis longus, flexor digitorum longus, peroneus brevis and peroneus longus \cite{46, 95}. The navicular bone creates the apex of medial longitudinal arch and is often used as a reference point by podiatrists and researchers to assess arch function \cite{49, 74, 89}.

When under load, the tarsus bones displace downwards and the two ends of the longitudinal arch, the calcaneous and the metatarsal-phalangeal joints move further apart. This action is referred to as arch compression and causes the tendons and ligaments of the arch to stretch. The presence of arches set human feet aside from other mammals and primates, highlighting their fundamental importance in bipedal locomotion \cite{51}. 

\cite{46, 95}
2.3 Evolution of the Foot

Evidence suggests that during the progression from arboreal life to obligate bipedalism \(^{(51)}\) the structure of the human foot changed significantly. The first evidence of bipedal locomotion was seen in *Sahelanthropus* approximately 6 - 7 million years ago \(^{(20,114)}\) and there is abundant fossil evidence that *Australopithecines* habitually walked bipedally by at least 4.4 million years ago \(^{(109)}\). The foot of the *Australopithecines* has many features in common with the modern day human foot that it does not share with its Chimpanzee ancestors suggesting that they may be beneficial for bipedal locomotion. These features include, but are not limited to; a large flat heel bone, decreased tarsal length, a non-opposable hallux in line with the other digits, and longitudinal arches \(^{(51)}\). Evolutionary biologists hypothesise that these adaptions, particularly the development of arches, assist with impact absorption and increase the economy of both bipedal walking \(^{(16, 51)}\) and running \(^{(4)}\). Despite the fact that *Australopithecus* habitually walked, fossil evidence suggests they only had partial longitudinal arches, still possessing a plantar process on the navicular and a partially divergent hallux \(^{(51, 55)}\). While this partially arched foot likely provided *Australopithecus* with an advantage in bipedal walking, it lacked the spring like properties to assist in running \(^{(89)}\). Approximately 1.5 million year-old footprints found in northern Kenya suggest that the modern foot, with fully formed arches capable of assisting both walking and running, likely did not evolve until the genus *Homo* \(^{(13)}\).

With the foot being the sole interface with the ground during bipedal locomotion, its ability to withstand and generate ground reaction forces most likely contributed to its change in form. When under load the arches of the foot compress, assisting in the absorption and damping of ground reaction forces \(^{(5, 60)}\). Given that these forces are high during running, approximately 1.5-3 times body weight, effective absorption and damping is important. Additional to impact absorption, the arched foot is believed to assist walking and running, by acting as a rigid lever and spring, respectively \(^{(16, 42, 60)}\).

During walking, the foot/ankle complex produces the majority of propulsive work \(^{(85, 90)}\) and the arched nature of the foot is thought to play an important role by increasing mid-foot rigidity to enhance plantarflexion power about the ankle at toe off \(^{(16, 29)}\). Comparison of foot movement between human and non-human apes have revealed substantial differences in foot motion, with the arched nature of the human foot allowing it to act as a fulcrum during push off \(^{(16, 29, 32)}\). This paradigm is supported by dynamic simulation investigations which have identified that increasing stiffness of the human midfoot can reduce the metabolic cost of walking \(^{(36, 99)}\). Nevertheless, a recent study by Bates et al. \(^{(11)}\) identified significant overlap in mid-foot plantar...
pressure measurements between humans and non-human apes, with some human individuals recording the highest mid-foot pressures (insinuate large mid-foot flexibility). This new evidence does not refute the lever theory because average human mid-foot pressures were low. However, the results do suggest the influence of mid-foot rigidity on the evolution of the foot’s shape may not be as significant as previously thought. The authors speculate that the increased mid-foot compliance identified in certain individuals may indicate a functionally tuneable system that provides an alternative benefit by aiding stability on variable surfaces.

2.4 The Elastic Mechanisms of the Arch

The arched nature of the foot is believed to offer additional benefits during running compared with walking. The spring-mass gait mechanics employed during running, combined with large ground reaction forces allow the arch of the foot to act as an elastic spring purportedly reducing the metabolic cost of running. When the arch of the foot is loaded during running it compresses, causing the supporting tendons and ligaments to stretch. Their high collagen content results in the majority of this strain energy being stored and subsequently returned as positive mechanical work. Alexander et al. established that tendons can stretch approximately 8% without breaking and return approximately 93% of the energy they store, with the remainder lost as heat. The storage and return of energy in tendons and ligaments is highly effective when compared with muscles, storing 3200 J kg\(^{-1}\) versus 3 J kg\(^{-1}\). Instead of storing elastic energy, muscles generally act to absorb energy (during the deceleration phase of gait), which is largely dissipated as heat, then produce new mechanical work at each step in the acceleration phase of the locomotive cycle. While elastic energy returned from certain tendons and ligaments (such as the Achilles tendon) has been shown experimentally to reduce the metabolic cost of running, whether the magnitude of energy returned from the arch of the foot is large enough to elicit a metabolic saving has received less attention.

A landmark investigation by Ker at al. was the first to establish that compression of the arch of the foot has the potential to reduce the metabolic cost of running. They studied the elastic energy stored in the tendons and ligaments of cadaver feet, and when the results were extrapolated to replicate the forces experienced by a 70 kg man running at 4.5 ms\(^{-1}\), they determined that the arch of the foot was capable of storing approximately 17 joules of energy during stance. This equates to approximately 17% of the mechanical work of the stance phase of running, leading Ker and colleagues to posit that the foot was capable of storing and
returning a large enough magnitude of elastic energy to significantly contribute to the metabolic cost of running. This important finding has led to the hypothesis that the arches are vital for economical running and that they were a key development in the evolution of the foot’s modern form \(^{(19)}\). However, while the energy saving function of the arch has been accepted in fields ranging from running mechanics \(^{(89)}\), human evolution \(^{(19)}\) and footwear design \(^{(76)}\), the contention that the foot acts as an energy saving spring has, to the best of the author’s knowledge, never been tested directly during running itself.

### 2.4.1 Arch Mechanics

While the link between elastic energy storage in the arch and metabolic cost has not been investigated during locomotion, a number of studies have detailed the mechanical function of the arch during walking and running. The most common method for assessing longitudinal arch movement during dynamic tasks is via a skin mounted marker on the navicular tuberosity (depicting the apex of the arch) \(^{(12, 18, 49, 60, 74, 89)}\). When the static, neutral foot is subjected to significant compressive loading, the distance between the calcaneus and metatarsal heads increases, and the navicular tuberosity displaces downwards \(^{(43)}\). The integrity of the arch of the foot is maintained by a number of tendons and ligaments providing differing levels of support \(^{(54, 60)}\), suggesting they may have varying energy storage roles. The plantar fascia provides the highest contribution to arch stability followed by the plantar ligaments and spring ligament \(^{(54)}\). However, a finite element model developed by Giddings et al. \(^{(44)}\) estimated that plantar ligaments experience forces 4.8 times body weight during level running at \(3.71 \text{ ms}^{-1}\), while the plantar fascia only experiences forces 3.7 times body weight. This research suggests, while the plantar fascia is the most important structure for maintaining the integrity of the arch, the plantar ligaments may be the most important for storing elastic energy. It is also believed that both extrinsic \(^{(8, 97, 102)}\) and intrinsic musculature \(^{(38)}\) contributes to arch integrity.

Kayano et al. \(^{(58)}\) were one of the first groups to assess the arch dynamically during walking and documented a pattern of bi-phasic compression and recoil throughout the contact phase of a walking gait cycle. Highlighting the influence of ground reaction force in arch compression, a significant correlation has been reported between vertical ground reaction force (vGRF) and navicular displacement, with peak displacement occurring within 5% of peak vGRF. \(^{(49)}\). Vicenzino et al. \(^{(107)}\) demonstrated that at midstance in running the arch reaches a greater vertical displacement compared with walking, most likely owing to the greater vGRF and compressive loads. Given the greater navicular displacement in running it is anticipated that
greater elastic energy storage occurs. Ker et al. (60) demonstrated that the tendons and ligaments of the foot exhibit exponential load-displacement and stress-strain relationships. The greater loads and navicular displacement observed in running therefore logically lead to a substantial increase in elastic energy storage. However, testing the direct relationship between elastic energy storage in the arch and metabolic cost is difficult due to the complex multifaceted nature of locomotion. Even if the magnitude of elastic energy stored in the arch was measured via invasive methods (63) or estimated via modelling (60), determining the significance of this contribution to the metabolic cost of locomotion is difficult.

2.5 ALTERING THE FOOT’S ARCH FUNCTION AND ELASTIC MECHANISMS

2.5.1 ARCH SUPPORT

Perhaps one method to effectively investigate the energy saving role of the arch is to remove the arches spring function, for example with orthotic insoles, and monitoring the subsequent effect on metabolic cost. Orthotics are commonly prescribed to treat foot and lower limb injuries, particularly those arising from abnormal foot biomechanics (92). Most orthotic devices possess arch support that either conforms to the shape of the medial longitudinal arch or aims to control deformation (34). While orthotics can be effective at reducing pain associated with patella-femoral pain syndrome (79), plantar fasciitis (6) and other lower limb ailments (30), it is possible they have a detrimental effect on elastic energy storage in the arch of the foot (63). Orthotics can be fashioned in a variety of ways, but if intended, have been shown to effectively reduce tension in the plantar fascia (35, 63) and likely the other tendons and ligaments in the arch. A number of studies have investigated the effect of orthotic insoles on metabolic cost with the general consensus being that orthotic use leads to a significant increase in metabolic cost (14, 21, 53). This greater metabolic cost is generally attributed to the weight of the orthotic given that 100 grams of additional mass per foot increases oxygen cost by ~1% (41). However, it remains possible the increased metabolic cost may in part be due to reduced elastic return from the arch. Previous studies have employed orthotics of different designs and material properties and did not report the level of restriction imposed on arch compression. This makes drawing conclusions regarding the impact of removing the arch as a source of elastic energy on the metabolic cost of running problematic. Future research should document the elastic properties of the orthotics, control for weight and directly measure arch compression.
An alternative way to assess the importance of the arch in running is to examine the effect of removing its spring function on lower limb mechanics. Sagittal plane mechanics produce the majority of forward propulsion in running. Therefore, if the elastic energy returned from the arch does significantly contribute to running propulsion, its restriction would likely result in a compensatory increase in positive work at other lower limb joints. Again, orthotic intervention studies may provide some insight although these have generally been aimed at optimising transverse and frontal plane motion (92), with the limited number of orthotic studies including sagittal plane motion in their analyses reporting mixed findings. Peak ankle dorsiflexion angle has been shown to decrease (10, 31, 70) or remain unchanged (33) with orthotic use. Similarly, ankle range of motion has been shown to both increase (65, 66) and decrease (31). At the knee, the majority of studies report no difference in sagittal plane kinematics (31, 33, 66, 70, 71), although two reported an increase in peak knee flexion angle (10, 66), and one a reduced range of motion (65). These mixed results are likely due to differences in orthotic construction and participant injury status, and make it difficult to draw conclusions regarding compensatory mechanisms employed when the arch’s function is compromised. Nevertheless, perhaps a better way to understand the change in muscle stress with restriction of the arch is to examine lower limb sagittal plane kinetics.

If the arch spring does provide a significant contribution to propulsive energy, its restriction due to orthotics or arch support may result in an increase in moments, power and mechanical work at other lower limb joints (if velocity is held constant). However, very few studies have investigated sagittal plane kinetics at the ankle and knee with the use of orthotics during running and none have examined the hip joint. Mundermann and colleagues reported a change in both ankle and knee moments with the use of orthotic insoles (77), however this result was not the primary focus of the study and hence the direction of change was not stated. MacLean et al. found a significant increase in knee extension moments in a healthy population with the use of orthotics (70), but no difference in participants with knee injuries (71). Nigg et al. (84) also found a 2% increase in knee extension moments in a healthy population, but this difference failed to reach statistical significance.

Alongside the larger knee extension moments with the use of orthotics in a healthy population, Mundermann et al. (78) also reported an increase in the electromyography (EMG) amplitude of all four quadriceps muscles. In contrast, Kelly et al. (59) reported a decrease in activation in the vastus medialis, but did not measure the other quadriceps musculature. In an injured population, the absence of change in knee extension moments with orthotic use (71)
was supported by the EMG results of Nawoczenski and Ludewig \cite{80}, who found no change in vastus medialis or vastus lateralis. At the ankle mixed EMG results have been reported. While Mundermann et al. \cite{78} identified an increase in gastrocnemius medialis EMG amplitude in the propulsion phase of stance, Nawoczenski and Ludewig \cite{80} found no change, while Kelly et al. \cite{59} reported a decrease. With such contrasting results, further research is required to better decipher the role of orthotic use and its potential effect on arch mechanics and subsequent adjustments to lower limb mechanics. For example, an investigation into the relationship between joint power and mechanical work following orthotic intervention is yet to be conducted, despite the potential for providing some of the most useful information regarding muscle demand. In addition to improving our understanding of the arch’s role in running, information surrounding changes in lower limb kinetics, when the arch of the foot cannot effectively store elastic energy, may have useful implications for individuals with impaired arch function, such as flatfoot deformity \cite{105}, rigid pes cavus \cite{39}, midfoot break \cite{72}, following a plantar fasciotomy \cite{97}, and pre- and postpartum \cite{96}.

### 2.5.2 THE ARCH OF THE FOOT AND FOOT STRIKE TECHNIQUE

Another factor proposed to alter the elastic energy storage in the arch is the foot strike technique employed during running \cite{67,89}. Foot contact with the ground can be classified as either a rearfoot, midfoot or forefoot strike. A rearfoot strike is defined when the heel makes initial contact, a midfoot strike when the heel and ball of the foot land simultaneously, and a forefoot strike when the ball of the foot strikes the ground first \cite{23,67}. Due to this variation in foot contact location with the ground, it has been suggested that forefoot strike runners are capable of storing greater elastic energy in the arch of the foot than rearfoot strike runners. By landing on the ball of the foot in a forefoot strike technique the arch immediately experiences a bending force. This compressive force is not experienced by a rearfoot strike runner until later in stance when both the heel and ball of the foot make contact with the ground (Figure 2.2).

To date, only one study has investigated the effect of foot strike technique on the arch of the foot and found that during barefoot running the navicular undergoes greater vertical displacement, and the arch experiences increased strain, when a forefoot strike technique is used compared with a rearfoot strike \cite{89}. However, the participants of this study were habitual forefoot strike runners and therefore were likely more comfortable using the forefoot strike than the imposed rearfoot strike technique, particularly when running barefoot: it is common
for runners to switch from a rearfoot to a forefoot strike technique when running barefoot on a stiff surface to avoid discomfort (2, 15, 47, 67, 100). It therefore remains possible that the discrepancy in arch compression between foot strike techniques recorded by Perl et al. (89) was the consequence of technique unfamiliarity and discomfort. Nevertheless, this study alludes to possible differences in arch function between rearfoot and forefoot strike techniques hypothesised to influence the energetics of running (60). Interestingly, the study by Perl et al. (89) found no difference in metabolic cost between rear and forefoot strike techniques when participants ran shod, despite the differences in arch compression recorded when participants ran barefoot. Further investigation is required to first establish if elastic energy storage in the arch of the foot impacts the metabolic cost and lower limb mechanics of running, and secondly if forefoot or rearfoot strike techniques elicit differences in the arch elastic energy contribution.

Figure 2.2 Model of different forces (top) acting on the arch of the foot during a forefoot strike (A) and a rearfoot strike (B) technique. Arch compression commences later in stance when using a rear- compared to a forefoot strike (bottom). Image adapted from Perl et al. (89)
2.6 **Effect of Foot Strike Technique on Running Mechanics and Energetics**

2.6.1 **Barefoot Running**

In addition to potentially increasing the elastic energy storage in the arch of the foot, using a forefoot in preference to a rearfoot strike technique, is suggested to offer other performance and injury prevention advantages. A number of investigations (albeit not all), have identified that barefoot runners choose to adopt a forefoot strike \(^{15, 47, 67, 73, 100}\). Some researchers suggest this may indicate that the human foot is optimally evolved for this foot strike technique \(^{67}\). Lieberman et al. \(^{67}\) studied the foot strike patterns of habitually barefoot versus habitually shod runners from the Rift Valley Province in Kenya, and found habitually barefoot runners predominantly employ a forefoot strike (66%), followed by a midfoot (22%) and rearfoot strike technique (12%), respectively. Conversely, habitually shod Kenyan runners preferred a rearfoot strike technique (97%) followed by a midfoot strike (3%), with none adopting a forefoot technique when wearing shoes \(^{67}\).

Some in the running, scientific and allied health communities hold the view that because the human foot evolved whilst barefoot, it is ‘specialised’ for barefoot/forefoot running \(^{1, 82}\). These views have helped popularise barefoot or minimally shod running, with advocates suggesting that modern day running shoes, which have only been mass marketed and distributed since the early 1970’s, do not allow the foot to function naturally, possibly increasing the risk of injury and having a detrimental impact on performance \(^{1}\). However, the barefoot running performance benefits identified in the literature, can largely been attributed to the resultant reduction in foot mass \(^{21, 28, 40, 103}\). Furthermore, a number of studies have reported running in lightweight (< 150 gram) shoes is more economical than barefoot running, a finding possibly explained by energy return from the elastic shoe midsole and/or the ‘cost of cushioning’ theory, where muscles expend additional energy to cushion impact on hard surfaces \(^{40, 41, 103}\). There also remains limited peer reviewed evidence of reduced injury rates in barefoot compared with shod running \(^{101}\), although this form of running may indeed have benefits for certain individuals \(^{101}\). It may also be that the running technique *per se* (e.g. foot strike) may be more important than whether an individual runs barefoot or shod \(^{99}\), but as has been previously stated, “what is on ones feet may affect how one runs” (Lieberman et al. p. 64 \(^{68}\)).
2.6.2 Foot Strike Technique and Injury

Foot strike technique has been proposed to influence a runner’s risk of injury, with a forefoot strike believed to pose a lower risk \(^{25, 27, 93, 112}\). In support, a recent study by Daoud et al. \(^{25}\) found habitual forefoot strike runners experienced fewer than half the number of injuries of habitual rearfoot strike runners. Additionally, switching to a forefoot from a rearfoot strike technique has been shown to reduce symptoms of exertional compartment syndrome \(^{27}\). One explanation proposed for the high injury rates observed in rearfoot strike runners is the vGRF profile \(^{25, 67}\). The vGRF of rearfoot strike runners typically displays two peaks; an initial impact transient followed by an active peak \(^{23, 64, 67}\), whereas forefoot strikers typically display only an active peak \(^{64, 67}\) (Figure 2.3). The presence of the initial impact peak in rearfoot strikers, and more specifically the higher initial vertical loading rate, has been suggested to increase the risk of injury \(^{25, 67}\). Lieberman et al. \(^{67}\) found that the average vertical loading rate of the impact transient (or when not present, from foot contact to \(~6.2\%\) of stance phase), was seven times higher in barefoot rearfoot striker runners (habitually shod) when compared with barefoot forefoot striker runners (habitually barefoot). Nevertheless, while some studies have loosely associated high vGRF loading rates with injury, many have failed to identify a clear relationship between the two \(^{111}\). In fact, one prospective study found that subjects with high loading rates received fewer running-related injuries than those with low loading rates \(^{83}\). It should be noted, however, that all existing studies investigating loading rates and injury have been on habitual rearfoot strike runners and the link between loading rate and injury in forefoot strike runners is yet to be researched.

The double impact peak present in rearfoot strike runners is not the only variation in force profiles differentiating foot strike techniques. Interestingly, forefoot strike runners typically experience larger magnitude vGRF active peaks \(^{17, 62, 65, 110}\), larger peak medial-lateral \(^{17}\) and larger peak anterior-posterior ground reactions forces \(^{17, 62, 65, 94}\) than their rearfoot strike counterparts (Figure 2.3), most likely due to shorter ground contact times \(^{48, 87}\). Rearfoot strikers have also been reported to experience lower resultant tibial accelerations than forefoot strikers \(^{45, 66}\). High tibial accelerations have been linked to tibia stress injuries \(^{75}\). Rearfoot strike runners may therefore have a lower injury risk disposition for stress related tibial injuries than forefoot runners \(^{45}\), although there is currently no epidemiological data to confirm this. Rather than influencing injury frequency, the divergent ground reactions force profiles of rear- and forefoot runners may partially explain differences identified in injury locations. Separate studies by Kleindienst \(^{61}\) and Walther \(^{108}\), both found that the overall injury incidence did not differ between rear- and forefoot strike runners, however, Walther
reported that forefoot strikers were more prone to Achilles tendon injuries, while rearfoot strikers experienced higher rates of tibialis posterior, knee and hip injuries. Daoud et al. (25) also reported disparities in injury location between foot strike groups with forefoot runners experiencing greater rates of Achilles tendinopathy and traumatic joint sprains, while rearfoot runners were more prone to hip pain, sacrum stress fractures, repetitive joint sprains and plantar fascia injuries.

Figure 2.3 Example vertical, anterior-posterior (AP) and medial-lateral (ML) ground reaction forces (GRF) in body weights (BW) for rearfoot strike (black line) and forefoot strike (grey line) techniques during the stance phase of running. Rearfoot strike runners typically experience greater peak vertical impact transients while forefoot strike runners experience greater vertical active peaks. Forefoot strike runners also typically experience greater AP and ML peak forces.

2.6.3 FOOT STRIKE TECHNIQUE AND LOWER LIMB KINEMATICS

Significant differences in lower limb joint kinematics have been identified between rearfoot and forefoot strike runners, which likely contribute to the divergent ground reactions force profiles and injury locations. Rearfoot strike runners make initial foot contact with the ankle in dorsiflexion, then plantarflex until the foot is flat on the ground, after which they dorsiflex until mid-stance (98, 111). Forefoot strikers land with their ankle in a plantarflexed position and immediately dorsiflex until mid-stance (98, 111). The peak ankle dorsiflexion angle achieved is similar, but forefoot strikers experience greater dorsiflexion excursion (65, 66, 87) and velocity (66)
compare with rearfoot strike runners. In the frontal plane, both techniques contact the ground with the foot in an inverted position which is immediately followed by ankle eversion \(^9, 62\). Rearfoot runners generally achieve a greater peak ankle eversion angle (at approximately 35-45\% of stance \(^9\)) \(^62, 66, 91\), while forefoot strikers experience greater eversion excursion and velocity due to a greater angle at touchdown \(^62, 65, 66, 91\). Studies report mixed kinematic results at the knee. At initial contact, knee flexion is typically greater in forefoot compared with rearfoot runners \(^98\), although this is not always statistically significant \(^64, 111\). Peak knee flexion \(^64\) and flexion excursion \(^65, 66\) are reported by some studies to be greater in rearfoot runners; however other investigations identify no differences between foot strike techniques \(^87, 89, 98\).

No disparities have been identified in peak knee adduction angles \(^64\). Hip flexion angle at initial contact is reported to be the same in rear- and forefoot strike runners in studies by Kulmala et al. \(^64\) and Williams et al. \(^111\). However, Shih et al. found a greater initial hip flexion angle in rearfoot strikers accompanied by larger hip sagittal plane excursion \(^98\). Kulmala et al. \(^64\) reported greater peak hip adduction in rear- compared with forefoot strike runners.

Knowledge of lower limb joint kinematics gives valuable insight into how foot strike techniques differ, however they only provide information regarding the movement, not the cause. An additional approach to improve our understanding of the potential injury risk posed by rearfoot and forefoot strike techniques is to examine joint kinetics via inverse dynamic analysis.

### 2.6.4 Habitual Rearfoot Strike versus Habitual Forefoot Strike Joint Kinetics

Inverse dynamic analysis incorporates the transfer of ground reactions forces to the lower limb and the internal forces acting at each joint to produce movement, and can provide insight into loading related injury risk. Significant differences in lower limb moments, powers and mechanical work have been quantified by a number of research groups investigating rearfoot and forefoot strike kinetics (Table 2.1) \(^50, 62, 64, 87, 110, 111\). Only three studies \(^50, 64, 87\) employed habitual runners in both foot strike groups (habitual rearfoot versus habitual forefoot strike). Habitual rearfoot runners are reported to place more demand on the knee joint than habitual forefoot runners, experiencing larger peak abduction moments \(^64\), peak negative instantaneous power \(^87\), mechanical work absorption \(^50\) and reduced knee stiffness \(^50\). Habitual forefoot runners are instead reported to encounter greater plantar-flexion moments \(^64\), reduced ankle stiffness \(^50\) and to absorb more mechanical work at the ankle during the first half of stance \(^50\). Only joint moments have been investigated at the hip with no differences identified between foot strike groups \(^64\). Positive instantaneous power and
mechanical work production have received very little attention despite their importance in understanding muscular fatigue and energy use. Paquette et al. investigated peak positive power at the ankle and knee joints between foot strike groups but reported no differences. However, as this was not a primary aim of their study the statistical analyses used may have prevented differences from being identified.

2.6.5 Habitual versus Imposed Foot Strike Technique Joint Kinetics

Other research examining the effect of foot strike technique on ankle, knee and hip joint kinetics include work by Kleindienst et al. and Williams et al. These studies recruited habitual rearfoot strike runners who each performed two conditions; a habitual rearfoot strike and an imposed forefoot strike. Thus, the findings may not necessarily reflect variation in foot strike techniques, but rather the differences between habitual versus imposed patterns. Williams et al., focusing on negative joint power during stance, found that habitual rearfoot strikers experienced greater peak negative power at both the knee and hip joints, yet displayed greater ankle joint peak negative power when adopting an imposed forefoot strike. Kleindienst et al. reported comparable results in peak power absorption, namely greater absorption at the knee and ankle in rearfoot and forefoot strike respectively, but found no differences at the hip joint. Kleindienst et al. also found ankle plantar-flexion moments were larger when adopting an imposed forefoot strike compared with those measured during a habitual rearfoot strike. At the knee joint, peak extension moments were significantly greater when using a habitual rearfoot strike, yet peak internal and external rotation moments were greater with an imposed forefoot strike. Knee abduction moments were greater during landing with an imposed forefoot strike but no difference was identified in peak abduction moments.

An earlier study by Williams et al. compared ankle and knee joint moments, negative instantaneous power and mechanical work absorption between habitual forefoot strike runners and habitual rearfoot runners performing a forefoot strike (imposed forefoot strike). This partial cross-over design found that while the imposed forefoot strike group were able to replicate the kinematics of the habitual forefoot strike runners, they were not able to replicate the vGRF, peak plantar-flexion moments or peak negative power at the ankle. This highlights the necessity of recruiting habitual rearfoot and habitual forefoot strike runners if one is to adequately elucidate differences in the two foot strike patterns. These results also emphasise that runners should be cautious when changing to an imposed foot strike technique.
as initially they may not be able to replicate the lower limb mechanics of the habitual technique, potentially placing them at greater risk of injury. Further investigation is needed in this area in order to understand the merits/detriments of changing foot strike technique.

### 2.6.6 Altering Arch Function: Influence of Foot Strike Technique

As previously discussed, it has been hypothesised that forefoot strike running results in greater elastic energy storage in the arch of the foot than rearfoot strike running \(^{(67, 89)}\). The elastic function of the arch has been proposed to significantly contribute to the metabolic cost of running \(^{(60)}\). If this is the case, removing or hindering the arch’s function is likely to result in significant compensatory mechanisms in the lower limb to maintain the required work output for locomotion. Given the previously identified lower limb kinetic differences between rearfoot and forefoot strike techniques (Table 2.1), it is possible that hindering the arch spring may result in a myriad of compensatory mechanisms. Laughton-Stackhouse et al. \(^{(65)}\) are the only group to have investigated the effect of modifying arch function in runners with different foot strike techniques. They found no interaction effect of custom orthotic intervention and foot strike technique on joint kinematics or kinetics. Although, very few variables were analysed, those previously identified to differ between rear- and forefoot strike runners, such as sagittal plane negative peak power and mechanical work absorption (Table 2.1), were not included. All runners were also habitual rearfoot strikers who performed both foot strike techniques, which as highlighted by Williams et al. \(^{(110)}\) may not adequately represent the differences between habitual techniques. Additionally, the magnitude of arch compression restriction was not recorded and subsequently the link, if any, between the function of the arch spring and changes in lower limb kinetics in rear- and forefoot strike techniques, remains unclear. Further research is clearly warranted as it will provide important information for; rear- and forefoot strike runners who are considering orthotic intervention or the purchase of supportive footwear; for individuals with permanently impaired arch function such as flatfoot deformity \(^{(105)}\), rigid pes cavus \(^{(39)}\), midfoot break \(^{(72)}\), or following plantar fasciotomy \(^{(97)}\); and for women during and after pregnancy who experience changes in arch morphology \(^{(96)}\).
Table 2.1 Summary of joint kinetic variable comparisons in the literature between habitual and imposed, rearfoot strike and forefoot strike techniques.

<table>
<thead>
<tr>
<th>Group</th>
<th>Peak Flexion-Extension Moments</th>
<th>Net Peak Power</th>
<th>Mechanical Work</th>
<th>Sing Phase Moments, Power, Work</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Ankle</td>
<td>Knee</td>
<td>Hip</td>
<td>Ankle</td>
</tr>
<tr>
<td></td>
<td>Neg</td>
<td>Pos</td>
<td>Neg</td>
<td>Pos</td>
</tr>
<tr>
<td>Hab RFS vs. Hab FFS</td>
<td>FFS</td>
<td>No diff</td>
<td>RFS</td>
<td>No diff</td>
</tr>
<tr>
<td>Hab RFS vs. Imp FFS</td>
<td>FFS</td>
<td>RFS</td>
<td>FFS</td>
<td>RFS</td>
</tr>
<tr>
<td>Hab FFS vs. Imp RFS</td>
<td>FFS</td>
<td>No diff</td>
<td>FFS</td>
<td>No diff</td>
</tr>
<tr>
<td>Hab RFS vs. Imp RFS</td>
<td>FFS</td>
<td>No diff</td>
<td>FFS</td>
<td>No diff</td>
</tr>
</tbody>
</table>

Hab = habitual, Imp = imposed, RFS = rearfoot strike, FFS = forefoot strike, Neg = negative, Pos = positive, Total = sum of ankle, knee and hip stance phase work, diff = difference. Blank = no statistical data exists in the literature.
2.6.7 Foot Strike Technique and Performance

Foot strike technique has not only been proposed to alter injury risk, but it has also been suggested that a forefoot technique offers a performance advantage over a rearfoot strike\(^{93, 112}\). Interestingly, compared with recreational runners, a larger percentage of elite distance runners prefer a forefoot strike or midfoot strike technique\(^{52, 53, 57}\). However, a recent study by Gruber et al.\(^{48}\) identified no difference in metabolic cost between habitual rearfoot strike and habitual forefoot strike runners, suggesting that the greater number of elite forefoot strike runners is perhaps an effect of speed rather than a superior foot strike technique. Other studies have investigated the metabolic cost of rear- versus forefoot strike techniques\(^{7, 24, 89}\), however these only recruited one cohort who performed both foot strike techniques (habitual and imposed) and then performed within-subject comparisons. While these studies cannot conclusively answer the question of which technique elicits a lower metabolic cost, they do provide information concerning the merits of switching to an imposed foot strike technique. Interestingly, both Gruber et al.\(^{48}\) and Perl et al.\(^{89}\) found no difference in metabolic cost when habitual forefoot strike runners switched to an imposed rearfoot strike. However, when habitual rearfoot strikers switched to an imposed forefoot strike, Gruber et al.\(^{48}\) found runners significantly increased their rate of oxygen consumption. In a similar research design, Ardigo et al.\(^{7}\) found no difference in metabolic cost between the two conditions but external mechanical work was significantly higher when the forefoot strike was imposed on natural rearfoot strikers. Cunningham et al.\(^{24}\) also found no difference in the rate of oxygen consumption during running; however, they did not specify the preferred foot strike technique of their sample. Only one study has been identified as investigating the metabolic cost of midfoot striking and found habitual rearfoot runners ran with a significantly lower metabolic cost than habitual midfoot strikers, despite recording similar VO\(_2\) max and anaerobic threshold results\(^{86}\).

At present no clear metabolic advantage appears to exist between habitual rearfoot and habitual forefoot strike runners\(^{48}\), although rearfoot runners may have a metabolic advantage over habitual midfoot strikers\(^{86}\). Interestingly, it may be possible for forefoot runners to switch to an imposed rearfoot strike without altering their metabolic cost\(^{48, 89}\), although switching to an imposed forefoot strike may incur a metabolic penalty for rearfoot runners\(^{48}\).
2.7 **SUMMARY**

The foot plays an essential role in dictating overall joint and limb mechanics in running. Its unique arched shape and ability to contact the ground in a variety of ways (rearfoot and forefoot strike) are two keys aspects that can affect both lower limb joint mechanics and metabolic energetics. During running, compression of the foot’s arch causes the associated elastic structures to stretch and store energy. Return of this elastic energy during the propulsion phase has been hypothesised to significantly reduce the metabolic cost of running \(^{60}\). However, no study has directly assessed this during locomotion. If the contribution of the arch is as meaningful as hypothesised \(^{60}\), restricting this source of elastic energy is likely to significantly affect lower limb mechanics, as the muscles may be required to compensate in order to maintain the required work output for forward propulsion. Knowledge of the compensatory mechanisms employed may reduce the risk of overloading lower limb structures in individuals with impaired arch function.

The angle at which the foot contacts the ground during running (rearfoot versus forefoot strike technique) results in significant disparities in lower limb kinematics and kinetics, which may provide insight into the merits/disadvantages of each technique, although there remain many gaps in the current knowledge. Positive and negative instantaneous power and mechanical work at all lower limb joints are yet to be investigated between rearfoot and forefoot strike techniques and despite the differences in stride frequency \(^{48, 98}\), the impact of this on kinetic variables has not been extensively considered. Additionally, the majority of studies have ignored the swing phase in their analysis, thus preventing differences in total mechanical limb work from being distinguished. Due to claims of improved performance and reduced injuries, athletes are being encouraged by the lay media to switch from their habitual (preferred) rearfoot to a forefoot strike technique \(^{93, 112}\), despite little understanding of the mechanical consequences. A comprehensive investigation is needed to examine how lower limb mechanics of habitual rearfoot and habitual forefoot strike techniques differ and to assess the impact of switching to an imposed foot strike technique.

Foot strike technique also affects the arch of the foot, with greater compression recorded when using a forefoot compared with rearfoot strike technique \(^{89}\). However, is it not known whether this results in a level of elastic energy storage that facilitates a metabolic advantage. Given the variation in lower limb kinetics between rearfoot and forefoot strike runners, it is likely that they respond differently when the spring function of the arch is restricted. Despite
the applicability of this information for individuals with impaired arch function, the variation in compensatory mechanisms is currently not known.

In summary, it is known that the foot plays an integral role in dictating running mechanics, although there still remain many unanswered questions as to the exact role of the medial longitudinal arch and the adopted foot strike technique on lower limb joint mechanics and energetics, and importantly, the interplay between these two aspects of foot biomechanics. A comprehensive understanding of how lower limb kinetics differ between habitual and imposed foot strike techniques may reduce a runner’s susceptibility to injury, and clarifying the relationship between the arch of the foot, foot strike technique and the energy cost of running may uncover both fundamental structure-function relationships, as well as important insights into performance enhancement and injury reduction.

2.8 REFERENCES


45. Glauberman MD, Cavanagh PR. Rearfoot strikers have smaller resultant tibial accelerations at foot contact than non-rearfoot strikers. *J Foot Ankle Res.* 2014;7(Suppl 1):A93.


94. Rooney BD. Joint contact loading in forefoot and rearfoot strike patterns during running [Graduate Theses and Dissertations]. Digital Repository @ Iowa State University: Iowa State University; 2011. 39 p.


CHAPTER THREE

JOINT KINETICS IN REARFOOT VERSUS FOREFOOT RUNNING:
IMPlications of Switching Technique.

This manuscript was accepted for publication into Medicine & Science in Sports & Exercise, in December 2013, available online February 5th 2014 and published in print July 16th 2014.


The published manuscript and related media attention is provided in the Appendix D of this thesis.

The PhD candidate, Sarah M. Stearne, accounted for 80% of the intellectual property associated with the final manuscript. Collectively, the remaining authors contributed 20%.
LINKING STATEMENT

The following chapter (Study One) investigates the different foot strike techniques employed during distance running, specifically rearfoot and forefoot strike techniques. The study aims to quantify the differences in lower limb joint kinetics between habitual rearfoot strike and habitual forefoot strike runners in order to gain a better understanding of whether either technique offers a performance and/or injury advantage. Additionally, this study will investigate whether the joint-level characteristics, and any associated benefits, can be adopted by a runner performing an imposed foot strike technique.
3.1 **ABSTRACT**

**Purpose:** To better understand the mechanical factors differentiating forefoot strike (FFS) and rearfoot strike (RFS) running, as well as the mechanical consequences of switching techniques, we assessed lower limb joint kinetics in habitual and imposed techniques in both groups. **Methods:** All participants performed both RFS and FFS techniques on an instrumented treadmill at 4.5 ms\(^{-1}\) while force and kinematic data were collected. **Results:** Total (sum of ankle, knee and hip) lower limb work and average power did not differ between habitual RFS and FFS runners. However, moments, negative work and negative instantaneous and average power during stance were greater at the knee in RFS and at the ankle in FFS techniques. When habitual RFS runners switched to an imposed FFS they were able to replicate the sagittal plane mechanics of a habitual FFS, however the ankle internal rotation moment was increased by 33%, while knee abduction moments were not reduced, remaining 48.5% higher than a habitual FFS. Additionally, total positive and negative lower limb average power was increased by 17% and 9%, respectively. When habitual FFS runners switched to an imposed RFS they were able to match the mechanics of habitual RFS runners with the exception of knee abduction moments, which remained 38% lower than a habitual RFS and, surprisingly, a reduction of total lower limb positive average power of 10.5%. **Conclusions:** There appears to be no clear overall mechanical advantage of a habitual FFS or RFS. Switching techniques may have different injury implications given the altered distribution in loading between joints but should be weighed against the overall effects on limb mechanics; adopting an imposed RFS may prove the most beneficial given the absence of any clear mechanical performance decrements.
3.2 INTRODUCTION

Runners can be classified as either rearfoot, midfoot or forefoot strikers depending on the part of the foot that makes initial contact with the ground. A rearfoot strike (RFS) can be defined as when the heel of the foot/shoe makes initial contact with the ground. A midfoot strike (MFS) can be defined as when the heel and ball of the foot land simultaneously, and a forefoot strike (FFS) when the ball of the foot makes initial contact with the ground followed by the heel \[^6, 21\].

While a RFS is more predominant among long distance runners versus a MFS and FFS, the percentage of a MFS and FFS has been found to be greater among elite distance runners \[^14, 16\], (although this may be related, in part, to increased running speed). The higher prevalence of a FFS in elite runners has contributed to a recent sub-culture among recreational and competitive runners to adopt a FFS pattern, despite the majority naturally preferring a RFS \[^1, 11\]. Of the hypothesized performance advantages for adopting a FFS running technique purported by certain members of the running community, among the most common is a reduction in the prevalence of lower limb injuries \[^23\]. While a study by Walther \[^33\] concluded that the overall injury prevalence is similar between foot strike techniques, with differences only observed in the nature and/or location of the injuries, a recent retrospective study by Daoud et al. \[^7\] has shown that injury rates may indeed be lower among FFS when compared with RFS runners. Given the high incidence of running-related injuries (19.4 - 79.3% of runners incur a lower limb injury annually \[^9, 31\]) it is becoming common practice for coaches to recommend an athlete change their foot strike technique \[^1\] in attempts to mitigate injury risk.

However, despite the alleged advantage of FFS running for lowering injuries, how mechanical alterations are linked to differences/similarities in running mechanics and injury risk between foot strike techniques remains poorly understood. As a result, there is little empirical evidence supporting the rationale for switching between a habitual RFS to an imposed FFS technique (or vice versa). In context of sports injury prevention, misinterpretation of the factors that can contribute to injury and/or improve performance can have unintentional negative effects such as the implementation of an unsuccessful training intervention, or worse, unintentionally increasing an athletes’ risk of injury \[^2, 8\].

Recent studies indicate that joint kinetics may play a key functional role in differentiating between RFS and FFS running \[^35\], and may help to reveal how different foot strike techniques could offer potential functional advantages/disadvantages. For example, greater peak plantar-
flexion moments, negative instantaneous joint power (power absorption) and negative work have been reported at the ankle in habitual FFS running, while greater peak knee flexion torque, negative power and negative work are observed at the knee in habitual RFS running \(^{(13, 17, 19, 25, 27, 35)}\). Most of these alterations can be replicated by switching from a RFS to a FFS, notwithstanding the high ankle moments and peak negative ankle power in stance in habitual FFS runners \(^{(35)}\). The different distribution of joint moments, negative power and work observed between FFS and RFS runners may subsequently alter muscle and tendon function, leading to changes in the prevalence, location and risk of musculoskeletal injury.

Although these previous studies investigating joint kinetics have provided crucial knowledge concerning foot strike biomechanics \(^{(13, 17, 19, 25, 27, 35)}\), none have provided a comprehensive analysis of the lower limb joint kinetics between RFS and FFS running that includes an analysis of both habitual and imposed techniques in both groups. To start, the large majority of studies investigating foot strike technique have focused on the ankle and knee. No study to the authors knowledge have reported the effect of habitual and imposed foot strike techniques on joint moments, powers and work, at all of the major lower limb joints or their combined effect on the overall lower limb mechanics. This consideration is important given that any difference in ankle or knee joint mechanics may lead to functional adaptations at the hip joint that could also affect injury susceptibility and/or overall lower limb mechanics and subsequently running performance. Secondly, few papers report positive power and to the authors knowledge, none report positive work in the context of foot strike technique. This is surprising since positive power and work affect muscular fatigue and energy use \(^{(26, 30)}\) and their inclusion can prove essential for understanding the complex interaction between foot strike technique, gait mechanics and energetics, and injury risk. Thirdly, and importantly, no study has examined rates of joint loading (average moment rate) or work production/absorption (average joint power) despite the known association between loading rates and injury \(^{(22, 32)}\) and the higher stride frequency in FFS running \(^{(12, 28)}\). Finally, with the exception of Hamill et al. \(^{(13)}\), studies addressing the mechanics of altering foot strike technique have focused nearly exclusively on habitual RFS runners, and the effect on the above parameters when habitual FFS runners alter their technique to a RFS remains largely unknown. Consequently, our understanding of the potential merits of switching between either technique remains incomplete.

The purpose of this study was therefore to expand upon the recent work of Hamill et al. \(^{(13)}\) to further determine what mechanical differences exist between RFS and FFS techniques in competitive runners at each of the major lower limb joints, and their effect on the overall
lower limb mechanics. More specifically, we addressed two related questions. 1) How do the peak joint moments and peak instantaneous joint powers (both positive and negative) at the ankle, knee and hip differ between foot strike techniques? 2) How does the average moment rate and the average power generation/absorption (and work) differ between foot strike techniques, both at the individual joints and in the combined lower limb? In each of these questions we compared both habitual RFS and FFS runners and, importantly, also addressed the effect of switching between habitual and imposed foot strike techniques in both groups.

3.3 METHODS

3.3.1 PARTICIPANTS

Sixteen trained competitive male distance runners were recruited for the study, eight habitual RFS and eight habitual FFS runners. Those who landed with the heel and ball of their foot simultaneously (MFS runners) were not included due to their variable kinematic and kinetic profiles. Foot strike technique was confirmed using high-speed video recording at 100 Hz positioned in the sagittal plane (Basler A602fc-2, Ahrensburg Germany). Participants ran $89 \pm 35$ km per week (mean $\pm$ S.D.) with a history of $7.5 \pm 4.7$ years running experience (mean $\pm$ S.D.). All runners were experienced in using their imposed running technique and had practiced using this technique in training runs. There were no significant differences in the age, height or weight between RFS and FFS groups (age: RFS 21.9 $\pm$ 3.3 years, FFS 23.6 $\pm$ 4.1 years; height: RFS 187.1 $\pm$ 5.6 cm, FFS 184.6 $\pm$ 6.7 cm; weight: RFS 76.4 $\pm$ 3.4 kg, FFS 73.1 $\pm$ 8.9 kg; mean $\pm$ S.D.). The participants had not experienced any lower limb injuries in the six months prior to testing and presented without any pre-existing gait abnormalities. Participants provided written, informed consent prior to partaking in the study. All procedures were approved by The University of Western Australia (UWA) Human Research Ethics Committee prior to the commencement of testing (Approval ID: RA/4/1/4541, Appendix A).

3.3.2 JOINT KINEMATICS AND KINETICS

Participants completed a five-minute warm up on a force-plate instrumented treadmill (Bertec Corporation, Columbus OH, USA) at 3.33 ms$^{-1}$ in their habitual foot strike technique. Participants were then allowed as long as they needed to familiarise themselves with the imposed foot strike technique. After the warm-up participants were allowed to rest and their heart rate (Polar F1 Heart Rate Monitor, Kempele, Finland) return to within 10% of their
resting value before commencing the measurement trials. Two running trials at 4.5 ms⁻¹ (a speed commonly used in the participants’ training) were completed 1) habitual foot strike technique, 2) imposed technique, the order of which was randomized. Each measurement trial lasted for two minutes. All participants were provided with the same lightweight running footwear (Nike Lunaracer™).

Joint kinematic and kinetic data for the ankle, knee and hip were computed using three-dimensional (3D) inverse-dynamic gait analysis. 3D segment motion was recorded using an eight-camera near infrared 200 Hz Vicon MX 3D motion capture system (Oxford Metrics, Oxford UK). Segment kinematics were computed from 34 retro reflective markers attached to the pelvis, lower limbs and shoes using the lower body marker set outlined in Besier et al. [4]. Ground reaction forces from the instrumented treadmill were recorded at 2,000 Hz, and synchronized with the kinematic data using a Vicon MX-Net control box (Oxford Metrics, Oxford UK). Lower limb biomechanical modelling was performed in accordance with the UWA lower body model [4]. Ankle joint centres were defined as the mid-point between the medial and lateral malleoli anatomical landmarks. A six-marker pointer was used to identify the medial and lateral femoral condyles, with a functional knee helical axis to define knee joint centres and knee axes orientation. A functional method was also used to define the hip joint centres [4]. A custom foot alignment rig was used to measure calcaneus inversion/eversion and foot abduction/adduction to define the anatomical coordinate system of the foot segment. Joint coordinate systems were defined in accordance with ISB standards of Wu et al. [36]. All marker trajectories were filtered using a zero-lag 4th order low pass Butterworth filter with cut-off frequencies typically at 11 Hz, which was determined by a custom residual analysis algorithm for each participant (MATLAB, The MathWorks Inc., USA). Ground reaction forces were filtered at the same cut-off frequency as the kinematic data to mitigate any artefacts in joint moments arising due to un-accounted segment acceleration [5, 18].

Net hip, knee and ankle joint moments and instantaneous power from the right leg across the gait cycle were calculated using BodyBuilder software (Oxford Metrics, Oxford UK). Peak positive and negative instantaneous powers and peak moments in the sagittal, frontal and transvers planes were calculated during the stance phase and normalized to body mass.
Net positive work (J kg\(^{-1}\)) and net negative work (J kg\(^{-1}\)) for each joint were computed by integrating the positive and negative instantaneous joint power data with respect to time, respectively:

\[
W_j^{\text{pos}} = \int_{t_i}^{t_f} P_j \, dt \quad \text{for} \quad P_j > 0 \quad \text{[Equation 3.1]}
\]

\[
W_j^{\text{neg}} = \int_{t_i}^{t_f} P_j \, dt \quad \text{for} \quad P_j < 0 \quad \text{[Equation 3.2]}
\]

where \(j\) represents the joint (ankle, knee or hip), \(P\) the joint power, and \(t_i\) and \(t_f\) represent the start and end time the integration respectively.

Positive and negative work at each joint (ankle, knee and hip) was computed separately for the stance (as defined by a ground reaction force threshold of 10 N) and swing phases. The total combined limb work of the entire stride was subsequently computed as the sum of each joint’s (ankle, knee and hip) work production. The rate of work production/absorption (referred to here as average power) during running was computed by multiplying the right leg’s joint work (either positive or negative; Eq. 1 and 2) at each joint across the stride by two (assuming bilateral symmetry) to represent both legs and dividing by the participants stride time. As an index of cumulative joint loading, the average rate of joint moment production (referred to here as average moment rate) was computed as the average joint moment during stance divided by the stride time (Nm \(s^{-1}\)). We computed the average moment rate for both sagittal and non-sagittal moments. The percentage each joint contributed to total stance positive and negative work and average power was computed by dividing the individual joint work during stance by the total stance work.

3.3.3 Statistical Analyses

Five consecutive strides from the final minute of running for each participant from each trial were selected for analysis and used to compute participant mean data. These data were used to assemble group mean data for each running condition (habitual RFS, habitual FFS, imposed RFS, imposed FFS). Two-way mixed-model multivariate analyses of variance (MANOVA) were performed to determine if any significant differences in peak joint moments and instantaneous powers, work and average power, and average moment rates existed between foot strike groups (between-subjects factors) and habitual vs imposed conditions (within-subjects factors). To address our questions, a number of multivariate analyses were performed. Joint level analyses included a) peak joint moments and average moment rate (dependent variables: sagittal and non-sagittal plane moments and average moment rates in stance), b) peak
3.4 RESULTS

3.4.1 SPATIAL-TEMPORAL GAIT PARAMETERS

On average, stride and stance time were greater (4.4% and 13%, respectively) and stride frequency smaller (3.4%) in habitual RFS vs habitual FFS (Table 3.1). Of these only stance time was statistically different ($p = 0.013$) which is in line with the literature (24, 28). Swing times and duty factor (the fraction of the time spent in contact with the ground) were similar between habitual RFS and FFS. The only difference in spatial-temporal variables between habitual and imposed techniques was an increase in stride time when habitual FFS runners switched to an imposed RFS (Table 3.1; $p = 0.037$).
Table 3.1  Spatial-temporal gait parameters in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions.

<table>
<thead>
<tr>
<th></th>
<th>Habitual RFS</th>
<th>Imposed FFS</th>
<th>Habitual FFS</th>
<th>Imposed RFS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of participants</td>
<td>8</td>
<td>8</td>
<td>8</td>
<td>8</td>
</tr>
<tr>
<td>Stride time (s)</td>
<td>0.68 ± 0.03</td>
<td>0.67 ± 0.02</td>
<td>0.65 ± 0.04</td>
<td>0.67 ± 0.04*</td>
</tr>
<tr>
<td>Stance time (s)</td>
<td>0.23 ± 0.02*</td>
<td>0.21 ± 0.01</td>
<td>0.20 ± 0.02*</td>
<td>0.21 ± 0.03</td>
</tr>
<tr>
<td>Swing time (s)</td>
<td>0.45 ± 0.02</td>
<td>0.46 ± 0.03</td>
<td>0.46 ± 0.04</td>
<td>0.46 ± 0.03</td>
</tr>
<tr>
<td>Stride frequency (Hz)</td>
<td>1.48 ± 0.06</td>
<td>1.50 ± 0.04</td>
<td>1.53 ± 0.11</td>
<td>1.50 ± 0.09</td>
</tr>
<tr>
<td>Duty factor</td>
<td>0.33 ± 0.03</td>
<td>0.32 ± 0.03</td>
<td>0.31 ± 0.03</td>
<td>0.32 ± 0.04</td>
</tr>
</tbody>
</table>

Mean ± standard deviation. Duty factor refers to the fraction of the stride spent in contact with the ground. (*) indicates a significant (α < 0.05) difference to RFS habitual. (†) indicates a significant (α < 0.05) difference to FFS habitual.

3.4.2 The Effect of Habitual and Imposed Foot Strike Techniques on Peak Joint Moments and Instantaneous Power.

Peak plantar-flexion moments were greater in both habitual and imposed FFS compared with habitual and imposed RFS, although these differences were not significant following the Bonferroni adjustment (Figure 3.1 & Table 3.2). Ankle internal rotation moments (also referred to as adduction moments in clinical gait literature) were not different between habitual RFS and habitual FFS runners, however they significantly increased when habitual RFS runners switched to an imposed FFS (33% increase; p = 0.012; Figure 3.1 & Table 3.2). Similarly, peak instantaneous ankle power production increased between habitual RFS and imposed FFS running (21% increase; p = 0.005; Figure 3.1 & Table 3.2). Peak instantaneous ankle power absorption was significantly greater in habitual FFS runners vs habitual RFS runners (45% greater; p = 0.011; Figure 3.1 & Table 3.2), increased when habitual RFS runners switched to an imposed FFS (85% increase; p = 0.001), and decreased when habitual FFS runners switched to an imposed RFS (-21% decrease; p = 0.016).

Peak extension moments at the knee were not significantly different between habitual RFS and habitual FFS runners. However when habitual FFS runners switched to an imposed RFS the knee extension moment increased by 29% (p = 0.016). Knee abduction moments were significantly smaller in habitual FFS vs habitual RFS runners (-105%; p = 0.010), but when either
group changed to their imposed foot strike technique they did not alter their abduction moments (Figure 3.1 & Table 3.2). Peak instantaneous power absorption at the knee was greater in all RFS conditions compared with FFS. There were no significant differences in either peak moments or powers at the hip joint between habitual RFS vs habitual FFS or when runners switched to an imposed technique (Figure 3.1 & Table 3.2). Additional non-significant net joint moment results are provided in Appendix C.

Figure 3.1  Group mean data for (A) ankle, knee and hip joint moments and (B) instantaneous powers across the stride (both stance and swing) in habitual forefoot strike (FFS) runners (solid black line), habitual rearfoot strike (RFS) runners (solid grey line), imposed FFS (black broken line) and imposed RFS (grey broken line). Vertical dashed lines represent toe-off in habitual FFS running (black) and habitual RFS runners (grey).
Table 3.2 Instantaneous peak moments (N m kg\(^{-1}\)) and positive (produced) and negative (absorbed) peak instantaneous power (W kg\(^{-1}\)) at the ankle, knee and hip joints during stance in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions.

<table>
<thead>
<tr>
<th>PEAK MOMENTS AND POWERS</th>
<th>MANOVA Interaction Effect Results</th>
<th>RFS Habitual Runners</th>
<th>FFS Habitual Runners</th>
<th>T-test Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>Moments N m kg(^{-1})</td>
<td>p value</td>
<td>RFS</td>
<td>FFS</td>
<td>RFS hab. vs FFS hab.</td>
</tr>
<tr>
<td>Power W kg(^{-1})</td>
<td></td>
<td></td>
<td></td>
<td>p value</td>
</tr>
<tr>
<td>Ankle</td>
<td>0.004(^4)</td>
<td></td>
<td></td>
<td>0.042</td>
</tr>
<tr>
<td>Plantar-Flexion</td>
<td>0.003</td>
<td>3.65 ± 0.27</td>
<td>4.02 ± 0.39</td>
<td>4.09 ± 0.47</td>
</tr>
<tr>
<td>Internal Rotation</td>
<td>0.161</td>
<td>-0.30 ± 0.08</td>
<td>-0.40 ± 0.12(^*)</td>
<td>-0.35 ± 0.14</td>
</tr>
<tr>
<td>Knee</td>
<td>0.010(^#)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Extension</td>
<td>0.002</td>
<td>2.56 ± 0.43</td>
<td>2.18 ± 0.29</td>
<td>2.09 ± 0.44</td>
</tr>
<tr>
<td>Abduction</td>
<td>0.148</td>
<td>2.11 ± 0.91(^*)</td>
<td>2.0 ± 0.77</td>
<td>1.03 ± 0.36(^*)</td>
</tr>
<tr>
<td>Hip</td>
<td>0.316(^#)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Extension</td>
<td>0.462</td>
<td>2.67 ± 0.16</td>
<td>2.98 ± 0.52</td>
<td>2.31 ± 0.42</td>
</tr>
<tr>
<td>Abduction</td>
<td>0.269</td>
<td>2.80 ± 0.70</td>
<td>2.81 ± 0.48</td>
<td>2.34 ± 0.25</td>
</tr>
</tbody>
</table>

Data are mean ± standard deviation. Bold numbers represent significant results after Bonferroni adjustment. (\(^4\)) indicates multivariate effect results. (\(^*\)) indicates a significant difference to RFS Habitual. (\(^\#\)) indicates a significant difference to FFS Habitual. Statistical significance was set at \(\alpha < 0.025\) for MANOVAs’s and \(\alpha < 0.0167\) for t-tests with Bonferroni post hoc adjustments.
3.4.3 The Effect of Habitual and Imposed Foot Strike Techniques on Work, Average Power and Average Moment Rate at the Ankle, Knee and Hip Joints During Stance.

No significant main effect of foot strike or condition was present in either work, average power, average moment rate, or percent contribution of each joint to total work and average power. However, there were significant interaction effects in all multivariate analyses (Table 3.3 & 3.4). Significantly greater negative ankle average power (49.5%; \( p = 0.003 \); Table 3.3) was observed in habitual FFS vs habitual RFS runners. Post hoc tests also revealed a significant increase in ankle negative average power during stance in imposed FFS vs habitual RFS running (63%; \( p = 0.001 \)) and habitual FFS vs imposed RFS (49%; \( p = 0.004 \)) (Table 3.3). Positive average ankle power was not different between habitual RFS vs habitual FFS runners, however, when habitual RFS runners switched to an imposed FFS it increased by 19.3% (\( p = 0.012 \)). Similar differences in ankle work were observed to those of ankle average power (Table 3.3). The percent contribution of the ankle joint to total negative lower limb work and average power during stance was significantly lower in a RFS technique compared with a FFS technique both when comparing habitual and imposed conditions (Figure 3.2). The average plantar-flexion moment rate was significantly greater in habitual FFS vs habitual RFS (31%; \( p = 0.001 \)) and vs an imposed RFS (24%; \( p = 0.008 \)) (Table 3.4). The average ankle internal rotation moment rate was significantly greater when habitual RFS runners switched to an imposed FFS (34%; \( p = 0.011 \); Table 3.4).

Negative knee average power during stance was significantly different across all conditions. Post hoc tests revealed that habitual RFS runners had 49% (\( p = 0.003 \)) greater negative average power at the knee than habitual FFS runners and 45% (\( p < 0.001 \)) greater negative average power than when they performed an imposed FFS technique. When habitual FFS runners switched to an imposed RFS they had 40% (\( p = 0.001 \)) greater negative average power at the knee (Table 3.3). The percent contribution of the knee joint to total negative lower limb work and average power during stance was significantly greater in a RFS technique compared with a FFS technique both when comparing habitual and imposed conditions (Figure 3.2). Similar to peak knee moments, the average knee extension moment rate was significantly greater when habitual FFS runners switched to an imposed RFS (16%; \( p = 0.002 \); Table 3.4). Despite not being statistically significant after a Bonferroni adjustment (\( p = 0.028 \)), the knee abduction moment rate was 87% greater in habitual RFS vs habitual FFS runners while there was only a 3% and 16% difference (non-significant) in habitual RFS vs imposed FFS and habitual FFS vs imposed RFS, respectively (Table 3.4).
After post hoc tests, there were no statistically significant differences in work, average power or average moment rate at the hip joint between foot strike techniques or conditions (Table 3.3 & 3.4). Although, habitual RFS and habitual FFS runners produced a similar amount of positive average power during stance a trend was present for greater (82%; $p = 0.035$) average power when habitual RFS runners switched to an imposed FFS. On the contrary, when habitual FFS runners switched to an imposed RFS they produced 43% less average power ($p = 0.091$; Table 3.3). The hip contributed approximately the same percentage of total positive and negative work and average power (Figure 3.2) in both RFS and FFS techniques.

3.4.4 THE EFFECT OF HABITUAL AND IMPOSED FOOT STRIKE TECHNIQUES ON TOTAL LOWER LIMB WORK AND AVERAGE POWER.

Significant interaction effects were found for both total lower limb positive and negative average power ($p < 0.001$ and $p = 0.002$, respectively; Table 3.3). Post hoc tests revealed a significant increase in the total limb positive average power when habitual RFS runners switched to an imposed FFS (17%, $p = 0.002$; Table 3.3) and a significant decrease in the total lower limb positive average power when habitual FFS runners switched to an imposed RFS (- 10.5%, $p = 0.014$; Table 3.3). There was a significant increase in the average total negative power when habitual RFS runners switched to an imposed FFS (8.9%, $p = 0.007$; Table 3.3). Differences in total lower limb work between habitual and imposed techniques were similar to those for average power and outlined in Table 3.3.
Bonferroni post hoc adjustments.

---

### Table 3.3

#### Mechanical work (J kg\(^{-1}\)) and average power (W kg\(^{-1}\)) at the ankle, knee and hip joints and the total lower limb (sum of ankle, knee and hip across stance and swing) in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions.

<table>
<thead>
<tr>
<th>Joint</th>
<th>Average Power</th>
<th>Stance Positive</th>
<th>Work J</th>
<th>P value</th>
<th>Effect</th>
<th>MANOVA p value</th>
<th>Interaction p value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>3.11 ± 0.44</td>
<td>3.56 ± 0.56</td>
<td>0.001</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee</td>
<td>3.31 ± 0.44</td>
<td>3.56 ± 0.45</td>
<td>0.001</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip</td>
<td>3.83 ± 0.45</td>
<td>4.57 ± 0.45</td>
<td>0.001</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Data are mean ± standard deviation. Bold numbers represent significant results after Bonferroni adjustment. (*) indicates a significant difference to FFS Habitual. Statistical significance was set at α < 0.05 for MANOVAs and α < 0.017 for t-tests with Bonferroni adjustment.

---

### Notes

- Average Power: This column represents the average mechanical power output for each condition.
- Stance Positive: This column indicates the stance phase under positive loading conditions.
- Work J: This column indicates the work done in joules.
- P value: This column indicates the statistical significance of the results.
- Effect: This column indicates the effect size of the results.
- MANOVA p value: This column indicates the statistical significance of the MANOVA results.
- Interaction p value: This column indicates the statistical significance of the interaction effects.

---

#### Results

- **RFS** vs **FFS**: Significant differences were observed in the average power across all joints and the total lower limb (sum of ankle, knee and hip across stance and swing) in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions.

---

#### Conclusion

The results indicate that the imposed forefoot strike (FFS) condition significantly alters the mechanical work and average power output compared to the habitual rearfoot strike (RFS) condition. This suggests that the change in strike pattern affects the energy transfer and mechanical efficiency at the ankle, knee, and hip joints during running.
Table 3.4  Average moment rate (rate of loading) (N m kg\(^{-1}\) s\(^{-1}\)) at the ankle, knee and hip joints during stance in habitual rearfoot strike (RFS), imposed forefoot strike (FFS), habitual FFS and imposed RFS conditions.

<table>
<thead>
<tr>
<th>MOMENT RATE</th>
<th>MANOVA Interaction Effect Results</th>
<th>RFS Habitual Runners</th>
<th>FFS Habitual Runners</th>
<th>T-test Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>Moment Rate N m kg(^{-1}) s(^{-1})</td>
<td>(p) value</td>
<td>RFS Habitual</td>
<td>FFS Imposed</td>
<td>FFS Habitual</td>
</tr>
<tr>
<td>Ankle</td>
<td>0.002(^{§})</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plantar-flexion</td>
<td>0.001</td>
<td>5.68 ± 0.74(^{*})</td>
<td>7.04 ± 0.98</td>
<td>7.44 ± 0.76(^{*})</td>
</tr>
<tr>
<td>Internal Rotation</td>
<td>0.108</td>
<td>0.59 ± 0.18</td>
<td>0.78 ± 0.19(^{*})</td>
<td>0.64 ± 0.22</td>
</tr>
<tr>
<td>Knee</td>
<td>0.035(^{§})</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Extension</td>
<td>0.009</td>
<td>4.09 ± 0.60</td>
<td>3.58 ± 0.44</td>
<td>3.50 ± 0.50</td>
</tr>
<tr>
<td>Abduction</td>
<td>0.424</td>
<td>3.31 ± 1.46</td>
<td>3.30 ± 1.36</td>
<td>1.77 ± 0.81</td>
</tr>
<tr>
<td>Hip</td>
<td>0.001(^{§})</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Extension</td>
<td>0.259</td>
<td>3.90 ± 0.43</td>
<td>3.36 ± 0.49</td>
<td>3.03 ± 0.95</td>
</tr>
<tr>
<td>Abduction</td>
<td>0.695</td>
<td>3.90 ± 0.93</td>
<td>3.99 ± 0.94</td>
<td>3.41 ± 0.48</td>
</tr>
</tbody>
</table>

Data are mean ± standard deviation. Bold numbers represent significant results after Bonferroni adjustment. \(\dot{\theta}\) indicates multivariate effect results. \(\cdot\) indicates a significant difference to RFS Habitual. \(\cdot\) indicates a significant difference to FFS Habitual. Statistical significance was set at \(\alpha < 0.025\) for MANOVA’s and \(\alpha < 0.0167\) for t-tests with Bonferroni post hoc adjustments.
Figure 3.2 Distribution of positive and negative work and average power between the ankle, knee and hip joints during the stance phase in habitual rearfoot strike running (RFS), imposed forefoot strike running (FFS), habitual FFS and imposed RFS. (*) indicates significant difference to habitual RFS, (#) indicates significant difference to habitual FFS p< 0.0167. (A) Habitual RFS positive work/average power in stance. (B) Habitual RFS negative work/average power in stance. (C) Imposed FFS positive work/ average power in stance. (D) Imposed FFS negative work/ average power in stance. (E) Habitual FFS positive work/ average power in stance. (F) Habitual FFS negative work/ average power in stance. (G) Imposed RFS positive work/ average power in stance. (H) Imposed RFS negative work/ average power in stance.
3.5 DISCUSSION

The higher prevalence of FFS running among elite athletes and the claims by some of reduced injury rates (7), has led to a tendency for both high-level and recreational athletes to adopt a FFS technique despite naturally preferring RFS running. In this investigation we assessed joint- and limb-level mechanical differences both between habitual RFS and habitual FFS runners, as well as the effect of changing between foot strike techniques.

3.5.1 THE DIFFERENCE BETWEEN HABITUAL RUNNING TECHNIQUE JOINT MECHANICS

Our finding that habitual RFS and FFS runners did not differ in the amount of total lower limb mechanical work or average power when running at 4.5 m s\(^{-1}\) indicates that one technique does not offer a clear-cut mechanical advantage over the other. This corroborates the recent finding by Gruber et al. (2013) (12), that no difference in metabolic cost exists between habitual RFS and FFS runners. It may instead be the altered loading profile and distribution of work and average power between lower limb joints, resulting in varying mechanical demands on the musculoskeletal system, which explains the functional effect of foot strike technique and injury risk.

The current study reinforces the view that habitual RFS runners place more demand on the knee joint in both the sagittal and frontal planes, while habitual FFS runners place more demand on the ankle joint in the sagittal plane (13, 25, 35). In a systematic review by van Gent et al. (2007), it is reported that 19.4 - 79.3% of runners experience lower limb injuries each year with the knee being the predominant site of injury (7.2 - 50%) (31). Given that 75% of runners adopt a RFS technique (14), the high occurrence of knee injuries compared with other joints may be associated with the large mechanical demand that occurs at the knee during stance in RFS running. While the lower mechanical demand at the knee might place habitual FFS runners at a lower risk of knee injury, the higher mechanical demand at the ankle might, on the other hand, place FFS runners at a greater risk of ankle related injuries such as Achilles tendinopathy and/or rupture or eccentric loading injuries in the triceps surae group.

Although the mechanics of the stance phase itself dictates most of the differences between foot strike techniques, our study also shows, for the first time, that the rate of mechanical loading and thus the cumulative work and moment production also plays an important role. Indeed, we found a more pronounced difference in the average ankle moment rate and
average negative ankle power in FFS vs RFS runners compared with differences in peak ankle moments and negative ankle work. In contrast, differences in the average knee moment rates and average negative knee power between foot strike techniques were smaller than those observed for peak knee moments and negative work. These findings suggest that the ankle in FFS running might be especially susceptible to cumulative overload injury, although the relative importance of peak and cumulative joint loading to musculoskeletal injury remains unclear.

By extending our measurements to include positive work and average power we found that the ankle contributed equal amounts of positive work in habitual RFS and FFS running (Figure 3.2A & E) even though there is greater negative work at the joint in FFS running. It is well known that the Achilles tendon is capable of storing energy absorbed at the ankle (negative work) as elastic strain energy, and returning this energy to provide positive ankle joint work in the second half of stance \(^{3, 10, 15, 20}\). It is possible therefore, that elastic recoil provides a greater contribution to positive work at the ankle in FFS runners. However, if this is the case it does not translate to a benefit in the total metabolic cost of running \(^{12}\).

### 3.5.2 The Effect of Switching Technique: RFS to FFS

For switching techniques to be effective as an injury prevention/management strategy the aim is for the imposed technique to replicate the habitual mechanics. When habitual RFS runners were instructed to switch to a FFS technique they were able to replicate the sagittal plane mechanics observed during habitual FFS running. However, despite there being no difference in ankle internal rotation moments or average moment rate between habitual RFS and habitual FFS runners, an imposed FFS increased these variables by 33% and 34% respectively (\(p = 0.012 \& p = 0.011\)). Additionally, the lower abduction moments about the knee joint in habitual FFS vs RFS runners reported here and in Kulmala et al. \(^{19}\) were not replicated in imposed FFS running. Considering that knee abduction moments possibly have a stronger link to knee injuries \(^{29, 34}\), as well as the increase in ankle transverse plane loading in imposed FFS running, calls into question the advice some coaches are giving to their athletes to change from a RFS to a FFS to reduce their injury risk.

Furthermore, when habitual RFS runners switched to an imposed FFS technique, the total positive and negative average power increased by 17% and 9% respectively (Table 3.3). This increase in mechanical cost is possibly related to an increase in muscle work and power and
thus may be detrimental to running performance. This provides a possible explanation for the recent results of Gruber et al. who found oxygen consumption increased when habitual RFS runners changed to an imposed FFS [12]. The source of the increase in positive average power in imposed FFS running is primarily at the ankle and hip joints (despite a non-significant difference at the hip joint after Bonferroni correction), while the increase in negative average power is primarily at the ankle joint. It is possible that the elastic mechanisms at the ankle are not as well developed in habitual RFS runners, and therefore increased work at the hip joint is required to maintain the required work output. The prospect that the calf musculature in habitual RFS runners may limit their ability to effectively adopt a FFS technique was highlighted in a study by Williams et al. [35] where RFS runners who performed a training session using a FFS experienced significant calf fatigue and delayed onset muscle soreness.

3.5.3 THE EFFECT OF SWITCHING TECHNIQUE: FFS TO RFS

In contrast to habitual RFS runners switching technique, when habitual FFS runners adopt RFS running they are able to replicate all of the joint mechanical characteristics of habitual RFS runners with the exception that they do not increase their frontal plane knee loads (peak abduction moment and average moment rate were 62% and 61% lower than habitual RFS running, respectively). Additionally, performing an imposed RFS required 10.5% less positive mechanical average power in the limb to maintain the same running speed. These findings were surprising, suggesting that switching to an imposed RFS could prove a useful strategy for injury rehabilitation whilst not affecting mechanical performance negatively. More specifically, utilising an imposed RFS in training may be useful to lower the ankle and Achilles tendon loading in athletes with ankle instability or tendon pathology/injury while at the same time reducing the overall average power, which may help mitigate fatigue.

3.6 CONCLUSION

Contrary to popular claims by some in the running community, this study found no clear mechanical advantage of habitual FFS running over habitual RFS running. Switching between RFS and FFS running techniques may have implications for injury reduction/recovery given the altered distribution in loading between joints but should be weighed against possible overall performance decrements/improvements. Switching from a habitual RFS to an imposed FFS may be detrimental to overall performance due to an increase in both positive and negative average lower limb power, which can help explain the recent finding that imposed FFS running
also requires more metabolic energy \(^{12}\). Furthermore, considering that the high knee abduction moments were not reduced by adopting a FFS technique questions the extent that switching technique will lower knee injury susceptibility. Habitual FFS runners can replicate the joint dynamics of a habitual RFS technique without incurring high knee abduction moments whilst, surprisingly, also lowering their positive average limb power. In this last regard, FFS runners adopting an imposed RFS technique, rather than the opposite, may prove the most useful training/rehabilitation strategy given the absence of clear mechanical performance decrements. However, it should be stated that these results represent the acute effects of switching technique and further research is needed to determine if these findings hold following a training intervention. Nevertheless, this study stands in contrast to the strategy of switching from a RFS to a FFS technique which is widely promoted by some members of the running community \(^{11, 23}\).

**ACKNOWLEDGEMENTS**

The authors would like to thank the participants for taking part in this study, Dr. Ben Jackson from the University of Western Australia for his assistance with the statistics analyses, and two anonymous referees for their valuable comments and constructive suggestions. No external funding was received for this work. S.M.S. was supported through an Australian Postgraduate Award scholarship.

**CONFLICT OF INTEREST**

None of the authors involved in the present study have any conflict of interest, financial, personal or otherwise which would influence this research and the results do not constitute endorsement by ACSM.

### 3.7 REFERENCES


27. Rooney BD. Joint contact loading in forefoot and rearfoot strike patterns during running [Graduate Theses and Dissertations]. Digital Repository @ Iowa State University: Iowa State University; 2011. 39 p.


CHAPTER FOUR

SOLE SEARCHING: THE FOOT’S ARCH AND THE ENERGETICS OF HUMAN LOCOMOTION.

This manuscript has been submitted for publication in Proceedings of The Royal Society B: Biological Sciences, September 2014.


Conference abstract pertaining to this manuscript is provided in the Appendix D of this thesis.

The PhD candidate, Sarah M. Stearne, accounted for 80% of the intellectual property associated with the final manuscript. Collectively, the remaining authors contributed 20%.
LINKING STATEMENT

The previous chapter (Study One) explored the influence of running foot strike technique (rearfoot vs forefoot strike) on joint mechanics and limb mechanical energetics. Significant disparities in individual joint and total limb mechanical work were identified between foot strike groups. The following paper (Study Two) will explore another aspect of running biomechanics that differs between foot strike techniques; the mechanics of the foot’s longitudinal arch. Greater compression of the arch has been observed during stance when using a forefoot compared with a rearfoot strike technique. This has been hypothesised to provide forefoot strike runners with a metabolic advantage over rearfoot strikers, based on the long accepted theory that the arch of the foot acts as a passive-elastic energy saving spring during running. However, while it is established that the tendons and ligaments of the arch are capable of storing and returning significant amounts of elastic energy, if this capability transfers to a metabolic saving has not been tested during locomotion itself, but relied instead on in situ cadaver experiments. This study will investigate, for the first time during locomotion, the effect of arch compression and recoil on the metabolic cost of level running, walking and incline running. It will also explore whether the greater arch compression observed in forefoot strike runners translates into a measureable metabolic advantage.
4.1 ABSTRACT

Using a novel insole technique that restricted compression of the foot’s longitudinal arch, this study provides among the most direct evidence of the arch’s energy-sparing role during human locomotion. When arch compression during moderate speed (2.7 ms\(^{-1}\)) level running was restricted by ~80%, gross metabolic cost increased by 6%. Restricting arch compression by only ~60% had a similar significant effect on the metabolic cost of running. The energy sparing function of the arch was not affected by foot strike technique, despite forefoot strike runners exhibiting greater arch compression than rearfoot strike runners. Interestingly, restricting arch compression had no effect on the metabolic cost of walking or incline running (3°), suggesting that the spring energy-saving role of the arch may be limited to level running. These findings provide among the most direct support for the interpretation that arch compression and recoil significantly contributes to lowering the energy cost of running and confirms our prediction that a metabolic saving depends on utilising the end-range of the arch compression. Our results have potential implications for arch support used in orthotic prescription and footwear design, and may provide additional insight into the foot’s role in human bipedal evolution.
4.2 **INTRODUCTION**

Running in legged animals has classically been characterised by the spring-mass paradigm \(^{8,14}\). During running, gravitational potential and kinetic energy is temporarily stored as elastic strain energy (primarily in the collagen of tendons and ligaments) during the first half of stance and subsequently returned in the second half of stance, propelling the body forward and upward. This form of elastic energy exchange is understood to reduce the metabolic cost of running by sparing mechanical work otherwise required by contracting muscle tissue \(^{12,44}\).

In humans, the longitudinal arch of the foot has been identified as an elastic storage-return mechanism \(^{27}\). By simulating the loads experienced during running in cadaver feet, Ker and colleagues concluded that approximately 17% of the mechanical work of running could be stored and returned by the foot’s arch as it undergoes compression and recoil over the stance phase \(^{27}\). Their investigation was the first to propose that the elastic nature of the foot’s arch contributes to the economy of running. This theory has subsequently been adopted in numerous investigations ranging from analyses of running mechanics \(^{41}\), the evolution of human bipedalism \(^{11}\), and footwear design \(^{35}\).

Since the initial study by Ker et al. \(^{27}\) the hypothesis that the foot is an energy saving spring has, to the best of our knowledge, never been tested directly during locomotion. The extent by which compression, and subsequent storage and return of elastic energy in the foot’s longitudinal arch, affects the metabolic cost of locomotion is therefore, not well understood. For example, does restricting arch compression elevate the metabolic cost of running and what magnitude of arch compression is necessary for a metabolic energy saving? How does the energy-reducing role of the arch span other locomotor tasks, such as walking and incline running? Is the longitudinal arch equally effective at reducing metabolic cost in forefoot strike (FFS) and rearfoot strike (RFS) running techniques, which are known to affect arch compression \(^{41}\) and other aspects of running biomechanics \(^{47}\)?

The purpose of this study was to address these questions and test the general hypothesis that compression and recoil of the arch of the foot affects the metabolic energy cost of running. To achieve this, we manufactured custom orthotic insoles designed to systematically restrict arch compression. We hypothesised that when arch compression was restricted during running, the reduction in the longitudinal arch elastic function would result in a significant increase in metabolic cost. To test this question, and to determine how the amount of arch restriction...
affects the energy cost of running, two separate insoles were designed that restricted arch compression either near-maximally, or by approximately 50% (compared with unrestricted shod running). Due to the exponential nature of the arch compression-elastic energy relationship \(^{(27)}\), we predicted that restricting arch compression in both insole conditions would result in a comparable increase in metabolic cost despite the nearly two-fold restriction in arch compression between conditions.

Due to the lower loads experienced during walking \(^{(38)}\) (and thus predicted lower arch elastic energy storage), and because of the additional positive mechanical work production required to raise the centre of mass vertically during incline running, (which cannot be generated from previously stored elastic energy \(^{(46)}\)), we predicted that the energy cost of walking and incline running would be less affected by reducing arch elastic energy storage/return than level running. Finally, we hypothesised that restricting arch compression and recoil would have a greater effect on the energy cost of level running in habitual FFS runners compared with habitual RFS strike runners given the recent evidence that FFS runners have greater arch compression \(^{(41)}\), and thus possibly greater arch elastic energy storage and return. To test these questions we measured metabolic cost (oxygen consumption), arch compression using three-dimensional (3D) motion capture, and made predictions of the elastic energy stored in the arch and the total mechanical work of locomotion in both normal shod gait and when restricting arch compression.

4.3 METHODS

4.3.1 PARTICIPANTS

Eight habitual RFS and nine habitual FFS male runners were included in the study. The participants had not experienced any lower limb injuries in the six months prior to testing nor presented with any pre-existing gait abnormalities. No significant differences in measured physiological variables, as determined by series of independent t-tests, existed between foot strike groups (age- RFS 25.5 ± 4.4 years, FFS 27.6 ± 3.4 years; height- RFS 185.3 ± 6.9 cm, FFS 181.8 ± 4.8 cm; weight- RFS 79.4 ± 6.8 kg, FFS 75.7 ± 5.9 kg; weekly running distance- RFS 39.4 ± 21.1 km, FFS 42.2 ± 36.0 km; mean ± SD). Participants did not regularly wear prescriptive orthotic insoles. Included participants were deemed to have normal foot structure as determined by the Foot Posture Index (FPI) \(^{(43)}\) (RFS 1.4 ± 1.4, FFS 1.0 ± 2.8). Measured foot variables did not differ between RFS and FFS groups; foot length- RFS 277.0 ± 11.4 mm, FFS
272.4 ± 6.1 mm: resting arch height (from sole to navicular tuberosity)- RFS 50.3 ± 7.3 mm, FFS 49.5 ± 9.5 mm and Achilles tendon moment arm (horizontal distance lateral malleolus to Achilles)- RFS 45.3 ± 4.1 mm, FFS 46.8 ± 5.4 mm. All FPI and foot anthropometric measurements were taken by a single experienced clinician (I.N.). Participants provided written, informed consent prior to inclusion in the study. All procedures were approved by The University of Western Australia (UWA) Human Research Ethics Committee (Approval ID: RA/4/1/4541, Appendix A).

4.3.2 CUSTOM ARCH-RESTRICTING INSOLES

Two pairs of custom made foot insoles were manufactured for each participant from 3D scans of the participants’ feet in a non-weight bearing neutral sub-talar joint position (ScanAny, Orthotech laboratories, Blackburn, Melbourne). Both insoles were made with the following specification; four millimetre polypropylene, high density arch fill (shore value ~350 - 400), four degree intrinsic rear foot grind, a balanced fore foot, maximum arch congruency and the heel ground to less than one millimetre such that heel-toe drop was deemed negligible. One insole was designed to fill the participants arch when the foot was positioned in a neutral non-weight bearing position, theoretically allowing minimal arch compression during locomotion (full arch insole; FAI). The second insole had a peak arch height five millimetres lower than the FAI, with the aim of allowing ~50% arch compression (half arch insole; HAI). The five millimetre reduction was chosen based on pilot work and arch compression data from Perl et al. (41) and Ker et al. (27). Participants were provided the insoles two weeks prior to testing to become familiar with wearing them.

To assess the energy storage/return of the insoles themselves during running the material properties of the insoles were tested using an Instron material testing machine with a custom testing mount (Instron Model 8874, Illinois Tool Works Inc.) and Instron Dynamic Software (Wave Matrix Version 1.2). A custom Acetal plastic testing mount (diameter 50 mm) that conformed to the slope and curve of the insole arch (point of max height) was designed. Force was applied to the insoles in a manner that replicated measured RFS and FFS stance time and stride frequency (47). The Instron was programmed to displace the insoles four millimetres during running for 100 cycles and the resultant load displacement was recorded allowing measurement of the insole stiffness and energy storage/return. The value of compression was
chosen to encompass the maximum insole compression measured during pilot testing of 2.7 ms\(^{-1}\) running, measured with a high speed camera). The Instron loading rate was calculated to be \(\sim 3600 \text{ Ns}^{-1}\).

4.3.3 TESTING CONDITIONS

New Balance Minimus road MR00 shoes were provided to all participants to wear for testing (approx. weight 180 grams, zero heel-toe drop, no medial arch support and a uniform EVA midsole). Pockets filled with lead weights were affixed to the laces of both shoes in order to standardise foot weight across all shoe and insole conditions. Prior to testing, participants completed a five minute warm-up on a force-plate instrumented split belt treadmill (Bertec Corporation, Columbus OH, USA) at a slow run.

Testing comprised of the following conditions; i) shoe only walk, ii) FAI walk, iii) shoe only level run, iv) HAI run, v) FAI run, vi) shoe only incline run, and vii) FAI incline run. All trials were completed on the force-plate instrumented treadmill and the order of conditions randomised to prevent any fatigue or order effects. To further ensure fatigue and trial order were not influencing results, the first condition was repeated at the end of the testing session. Participants reported minimal discomfort and no conscious change in their running technique whilst wearing the insoles (see Appendix C for questionnaire results).

All running conditions were performed using the runner’s habitual foot strike technique as confirmed by a sagittal high speed video camera (Casio EXILIM EX-F1, Casio Computer Co. LTD., Shibuya-ku, Tokyo; 300 Hz). In accordance with the literature, a RFS was defined when the heel of the shoe made initial contact with the ground and a FFS defined when the ball of the foot made first contact\(^{15, 30}\). A standardised walking speed of 1.1 ms\(^{-1}\) (representing a comfortable speed on the treadmill and within the range of the participants’ pilot tested preferred walking speeds) was selected to minimise arch compression and subsequently elastic energy contribution. To ensure our results were not affected by walking speed, a sub-set of participants (n = 8) also performed the shoe only and FAI walk at their individually preferred walking speed (average 1.3 ± 0.1 ms\(^{-1}\)). Similar metabolic cost results were found in the preferred walking speed and the imposed 1.1 ms\(^{-1}\) walking trials. All running trials were performed at 2.7 ms\(^{-1}\) (level and incline trials were performed at the same velocity to control for any speed effects). Pilot testing on a sub-set of participants (n = 8) revealed that running at faster speeds (3.5 ms\(^{-1}\)) caused the insoles to compress and thus limited the effect of the insole.
on arch compression, likely due to the higher joint loading at this speed. During incline trials the treadmill was set at three degrees (although not specifically instructed to do so, all runners maintained their level habitual foot strike technique). This inclination was selected to increase the mechanical work and metabolic cost but within aerobic levels as faster speed/incline combinations risked reliance on anaerobic metabolic pathways (34). The chosen running speed thus represents an optimised speed to test the effect of the insoles in both level and incline conditions.

4.3.4 METABOLIC COST

Participants were asked to abstain from caffeine on the day of the testing and to not eat in the two hours prior to arriving at the laboratory. Expired gasses were collected during rest (standing) and walking/running trials. Participants were required to breathe into a two-valve mouthpiece connected via two lightweight flexible tubes to a computerised oxygen and carbon dioxide gas analysis system [Morgan ventilation monitor (Morgan, Reinham, Kent, UK); oxygen and carbon dioxide analysers (Ametek SOV S-3A11 / Ametek COV CD-3A, Applied Electrochemistry, Ametek, Pittsburgh, PA)]. The ventilometer and gas analysers were calibrated before and immediately after each test using a one litre syringe pump and reference gas mixtures, respectively (BOC Gases, Chatswood, Australia). Each treadmill condition was performed until the participant reached a steady state of oxygen consumption (approx. 4 mins) after which a further minute of data was collected for analysis. During the incline conditions, blood lactate concentration levels were determined (Lactate-Pro, Arkray, LT-1710, Kyoto, Japan) after steady state was reached to ensure participants were exercising below their anaerobic threshold (previously determined during a maximum oxygen consumption test).

4.3.5 ARCH COMPRESSION AND 3D JOINT KINMATICS AND KINETICS

We used a custom 3D kinematic foot model to estimate arch compression. First, a previously established lower body model was used to define a rearfoot segment and ankle kinematics (6). Retro-reflective markers were affixed to the feet and ankles in accordance with Besier et al. (6), with additional markers placed on the navicular tuberosity and distal phalanx of the first metatarsal (Figure 4.1). The shoe upper was modified with five marker ‘windows’ coinciding with the marker positions, allowing markers to be placed directly on the foot and remain visible. To ensure marker positions remained unchanged after removing shoes, all foot
markers were detachable by a magnetic base that did not change location. This resulted in the markers being slightly offset from the anatomical landmark. The location of the marker relative to the anatomical landmark deviated minimally within their respective rigid segments during running, and the location of the functionally relevant anatomical landmarks were identified using a six marker wand in static pointer trials and expressed in the rearfoot anatomical coordinate system. The midpoint of these virtual landmarks was used to define both a rearfoot-forefoot joint centre (metatarsophalangeal [MTP] joint centre) as well as the base of the foot (Figure 4.1). A forefoot segment was created using the virtual metatarsal markers and a third virtual hallux marker, defined using the calibrated anatomical systems technique. Where a static pointer trial is used to locate and store for later reconstruction, virtual landmarks, expressed relative to a coordinate system defined by the metatarsal and hallux markers. A foot sole plane was created by transposing the rearfoot coordinate system so that its origin was the rearfoot-forefoot joint centre (the x-z-plane of this coordinate system is oriented parallel with the ground in a neutral standing posture). A small marker was also placed on the medial aspect of the insole in line with the maximum insole height and hence the maximum arch height.

Figure 4.1 Right foot medial view illustrating foot marker positions, the sole axis and navicular displacement measure. Windows were cut in the shoe upper which allowed markers to remain visible and the positions unchanged between conditions.
Motion of the retro-reflective markers were tracked using a ten-camera near infrared Vicon T-series 3D motion capture system (T40S, 250 Hz; Oxford Metrics, Oxford, UK) and sagittal plane high-speed video (300 Hz). Six consecutive right leg strides from the final minute of data collection were selected for analysis. A continuous trace of the vertical displacement of the navicular marker relative to the new rearfoot coordinate system in the y-axis was used as a measure of arch compression (Figure 4.1). This method minimised any effects that ankle inversion/eversion had on predicting arch compression since these motions do not alter the location of the navicular marker in the rearfoot coordinate system. Arch height was defined as the distance of the navicular marker relative to the base of the foot and tracked continuously throughout the stride. Arch height at initial foot contact during the shoe only level run was used as a reference height for each condition. Arch compression was defined as the difference between the reference height and the minimum arch height during stance in each condition. Results were compared with digitised high speed video footage (navicular marker relative to top of the shoe midsole); maximal navicular compression from the two methods correlated strongly (r = 0.88).

Inverse dynamics (net joint moments and joint reaction forces) were computed for the ankle in accordance with Besier et al. (6), as well as for the MTP joint (sagittal plane only), using Vicon BodyBuilder software (Oxford Metrics, Oxford, UK). Ground reaction forces (GRFs) from the instrumented treadmill were recorded at 2,000 Hz, and synchronised with the kinematic data using a Vicon MX-Net control box (Oxford Metrics, Oxford, UK). All marker trajectories were filtered using a zero-lag 4th order low pass Butterworth filter with cut-off frequencies typically at 14 Hz, determined by a custom residual analysis algorithm for each participant (MATLAB, The MathWorks Inc., USA). GRFs were filtered at the same cut-off frequency as the kinematic data to mitigate any artefacts in joint moments arising due to un-accounted segment acceleration (7, 28). The GRFs were ascribed to the forefoot segment when the centre of pressure was anterior to the MTP joint centre and to the rearfoot when it was distal. Inverse dynamic calculations of the forefoot segment assumed that the moments generated by the mass and inertia of the segment were negligible as per Stefanyshyn and Nigg (48).
4.3.6 ARCH ELASTIC ENERGY AND TOTAL MECHANICAL WORK OF LOCOMOTION

Our hypotheses are based on the premise that the insoles impede the storage of elastic energy in the arch and its subsequent ability to contribute to total mechanical work. How these variables changed between conditions was estimated using a simple model to predict arch elastic energy storage (Figure 4.2) and a force plate approach to measure total mechanical limb work of locomotion \(^{(17)}\). The arch energy storage model was based on the compressive load-energy storage function established by Ker et al. \(^{(27)}\) and calculations of the participants' individual ankle compressive loads (inverse dynamics) and arch compression (high-speed motion capture) (Figure 4.2). First, the maximal experimental ankle compressive load during stance was estimated by summing the computed joint reaction force with an estimate of the Achilles tendon force (the trial with the highest value was used). Achilles tendon force was estimated by dividing the net ankle joint moment by the Achilles tendon moment arm taken from calliper measurements during standing. The participant’s peak arch strain energy was then predicted from the estimated compressive ankle load using the strain energy versus ankle load relationship from Ker et al. \(^{(27)}\) (their Figure 2B was digitised and used to develop the strain energy equation \((J) = 0.269 \times^2 + 1.004 \times; \text{where } \times = \text{ankle compressive load (kN)}\)). We subsequently developed subject specific arch load-displacement curves (see Figure 4.2) using two fixed points \((0,0)\) and \((\text{max load}, \text{max arch compression})\), and varied a third point so that the area under the line of best fit for the three load-displacement points [power function, \(F_{\text{comp}} = y_0 + Ax^{\text{pow}}\), where \(F_{\text{comp}}\) is the compressive force (kN), \(y_0\) is the \(y\)-intercept (0), \(x\) is arch compression (mm) and \(A\) is a constant] matched the energy storage predicted from the above strain energy equation. The optimisation procedure was performed using a root solver algorithm in MATLAB (fsolve; The MathWorks, Natick, MA). The subject-specific arch load-displacement curve was subsequently used to estimate the elastic energy stored under other amounts of arch compression in the remaining walking/running trials by integrating the area under the arch load vs displacement curve for the specified arch compression. The amount of energy returned to the runner from the estimated stored arch elastic energy was calculated using a hysteresis of 0.12, based on data from Ker et al. \(^{(27)}\).
**Figure 4.2** A) Arch elastic strain energy – ankle compressive load relationship adapted from Ker et al.\(^{27}\) used to estimate arch strain energy from the participant’s maximum ankle joint compressive load. B) Subject specific load-displacement curve used to predict stored arch elastic energy during different conditions from measured arch compression.

Total mechanical work of the limbs was calculated during walking, level running and incline running using the individual limb work force plate approach described by Donelan et al.\(^{17}\). Initially limb power (P) was calculated (for the left and right legs separately) as the dot product of two vectors: the external force acting on the limb (\(F\)) and the velocity of the centre of mass (\(v_{COM}\)).

\[
P = \vec{F} \cdot \vec{v}_{COM} = F_z v_{z,COM} + F_y v_{y,COM} + F_x v_{x,COM} \quad \text{[Equation 4.1]}\]

Centre of mass (COM) velocities \((z,y,x)\) were calculated by time-integrating the COM accelerations determined from the vector sum of the left and right leg GRFs (see Donelan\(^{17}\)). Integration constants (offsets) were applied to the calculated COM velocities: \(z\) (vertical) was set so that the average COM vertical oscillation over one step equalled 0; \(y\) (fore-aft) was set to the treadmill speed; \(x\) (med-lat) was set so that the velocity at the start and end of the step were equal but opposite. Offsets were adjusted during the incline condition: \(z = \text{treadmill speed} \times \sin 3^\circ, \quad y = \text{treadmill speed} \times \cos 3^\circ\). Limb powers were integrated with respect to time \((t_i \text{ and } t_f \text{ represent the start and end of stance phase respectively})\) to determine positive \((W_{pos})\) and negative \((W_{neg})\) mechanical work performed by the limbs on the COM.

\[
W_{pos} = \int_{t_i}^{t_f} P \, dt \quad \text{for } P > 0 \quad \text{[Equation 4.2]}\]

\[
W_{neg} = \int_{t_i}^{t_f} P \, dt \quad \text{for } P < 0 \quad \text{[Equation 4.3]}\]
In walking, mechanical work from both the leading and trailing limbs were summed, as per the individual limbs methods described by Donelan et al. Prior to computing incline mechanical work, the treadmill global coordinate system was re-defined in Vicon Nexus to account for the 3 degree incline. We acknowledge that the arch elastic energy storage methodology has limitations therefore was intended as an ancillary approach to assess the link between arch compression and the metabolic cost of locomotion.

4.3.7 STATISTICS

Two-way repeated measures split-plot ANOVAs were performed to determine the effect of the custom insoles and foot strike technique on the following variables; metabolic cost, arch compression, predicted arch elastic energy storage, and total limb mechanical work of locomotion. The between subject factors were habitual foot strike technique (RFS and FFS) and the within subjects factors were shoe only, HAI (level running only) and FAI. The significance level was $\alpha < 0.05$ for ANOVA analyses. 3 x 2 ANOVAs were conducted for level running and 2 x 2 ANOVAs for walking and incline running. An additional 3 x 2 ANOVA was performed to determine the difference in maximum arch compression during shoe only walking and shoe only level and incline running conditions. In the 3 x 2 ANOVAs the location of the significant main effect was determined using a post-hoc pairwise comparison with a Bonferroni adjustment. In all ANOVAs where significant interaction effects were found, post-hoc t-tests were performed to determine where relationships existed.

4.4 RESULTS

4.4.1 TEMPORAL PARAMETERS

The insole had no effect on any temporal parameters (stance, swing or stride time). There was a main effect of foot strike technique on stance time ($p = 0.014$) in level running and on both stance ($p = 0.019$) and swing time ($p = 0.040$) in incline running. RFS runners spent longer in the stance phase in both conditions and a shorter time in the swing phase during the incline condition compared with FFS runners. See Appendix C for additional temporal parameter data.
4.4.2 ARCH COMPRESSION

In level running the insoles had a significant main effect on the maximum arch compression (navicular displacement) \( (p < 0.001) \) but the foot strike technique did not \( (p = 0.093) \). *Post-hoc* tests revealed that compared with shoe only running, both insoles prevented arch compression; HAI 61.4 ± 33.4% \( (p < 0.001) \) and FAI 78.9 ± 24.7% \( (p < 0.001) \), (Figures 4.3 & 4.4). *Post-hoc* tests also revealed there was a significant difference between the HAI and FAI arch compression during level running \( (p = 0.032; \text{Figures 4.3 & 4.4}) \).

The insole (FAI) had a main effect on maximum arch compression during walking \( (p < 0.001) \) but foot strike technique did not \( (p = 0.226) \). Similarly, there was a main effect of the insole (FAI) during the incline running condition \( (p = 0.001) \) but again no effect of foot strike technique \( (p = 0.053) \). The FAI reduced the maximal arch compression by 82.4 ± 21.1% and 68.5 ± 30.6% (Figure 4.4) compared with the shoe only maximal arch compression condition in walking and incline running, respectively. The condition (walk vs. level run vs. incline run; shoe only) had a significant main effect on maximum arch compression \( (p < 0.001) \), whereas foot strike technique did not \( (p = 0.166) \). *Post-hoc* tests reveals that compared with both level and incline shoe only running, maximum arch compression was reduced in shoe only walking \( (p = 0.002 \text{ and } p < 0.001 \text{ respectively; Figure 4.4}) \). Arch compression between level and incline shoe only running was of a similar magnitude; level running 10.6 ± 2.6 mm, incline running 10.5 ± 4.1 mm \( (p = 0.991; \text{Figure 4.4}) \).
Figure 4.3 Average arch compression (mm) ± standard error when running in the shoe only (light grey line), half arch insole (dark grey line) and full arch insole (black line) throughout the stance phase in level running. Zero indicates arch height at initial contact in the shoe only condition. Positive values indicate a slackened state of the arch elastic structures compared to initial contact in the shoe only condition and negative values indicate stretch compared to the shoe only initial contact.

Figure 4.4 Maximum arch compression (mm) (mean + standard error) relative to arch height at shoe only level running foot contact. * indicates significantly different ($p < 0.05$) to the shoe only trial in the same condition, † indicates significant difference between the half arch insole (HAI) and full arch insole (FAI) (level running only).
4.4.3 METABOLIC COST

A main effect of insole was observed on the rate of oxygen consumption during level running ($p < 0.001$), but there was no main effect of foot strike technique ($p = 0.578$) or an interaction effect between insole and foot strike technique. Post-hoc tests revealed that compared with shoe only running (36.5 ± 2.9 ml kg$^{-1}$ min$^{-1}$), there was a significant increase in gross metabolic cost with both the HAI (38.2 ± 3.2 ml kg$^{-1}$ min$^{-1}$, $p = 0.012$; + 4.5 ± 5.0%) and FAI (38.7 ± 3.4 ml kg$^{-1}$ min$^{-1}$, $p < 0.001$; + 6.0 ± 4.2%) insoles (Figure 4.5). However, there was no statistical difference in the metabolic cost between the HAI and FAI conditions ($p = 0.211$). During walking and incline conditions there was no main effect of insole (walking $p = 0.950$, incline $p = 0.164$) or foot strike technique (walking $p = 0.680$, incline $p = 0.355$) on the rate of oxygen consumption (walk shoe only 13.2 ± 1.5 vs. FAI 13.2 ± 1.5 ml kg$^{-1}$ min$^{-1}$; incline shoe only 46.5 ± 2.8 vs. FAI 47.0 ± 3.0 ml kg$^{-1}$ min$^{-1}$; Figure 4.5). When standing (weight bearing) metabolic rates were subtracted, the net metabolic cost data exhibited the same statistical results, albeit the percent differences between the insole and shoe only conditions were modestly increased (walking FAI 2.1 ± 18.0% non-significant; level running HAI 5.5 ± 6.2% ($p<0.05$), FAI 7.4 ± 5.1% ($p<0.05$); incline FAI 1.4 ± 3.7% non-significant). Blood lactate results collected during the incline conditions indicated that, with the exception of one participant, all trials occurred aerobically (<4 mmol). The incline and level running data from this participant were removed from further analyses.

![Figure 4.5](image)

**Figure 4.5** Percent change in the rate of oxygen consumption ($\dot{V}O_2$) (mean ± standard error) from the shoe only to insole trial across walking, level running and incline running conditions. FAI = full arch insole, HAI = half arch insole. * indicates significant ($p < 0.05$) difference between raw $\dot{V}O_2$ values between the shoe only and insole trial within the same condition.
4.4.4 Arch Elastic Energy Storage and Centre of Mass Work Estimations

During level running, the insole had a main effect on estimated elastic energy storage in the arch ($p < 0.001$), but foot strike technique did not ($p = 0.612$), nor was there an interaction effect between energy storage and foot strike technique. Post-hoc tests revealed that both the HAI and FAI insoles significantly reduced elastic energy storage compared with shoe only running (both $p < 0.001$), but there was no significant difference between the HAI and FAI insoles ($p = 0.151$; Figure 4.6A). In walking and incline conditions there was also a main effect of insole (FAI) on estimated elastic energy storage (both $p < 0.001$; Figure 4.6A) but no main effect of foot strike technique (walking $p = 0.727$, incline $p = 0.963$). The insoles and foot strike technique had no main effect on total limb mechanical work of locomotion in walking ($p = 0.782$, $p = 0.689$, respectively), level running ($p = 0.121$, $p = 0.137$, respectively) or incline run.
running ($p = 0.074$, $p = 0.357$, respectively) (Figure 4.6B). Material testing on the insoles indicated that they returned less than 0.3 J of energy during running, accounting for less than 3% of the elastic energy storage that was estimated to be lost when wearing the insoles (Figure 4.7).

See Appendix C for detailed data describing the arch compression, metabolic cost, estimated arch elastic energy storage and return and total limb mechanical work.

Figure 4.7 Average insole load-displacement curve from six insoles indicating minimal energy return from the insoles.

4.5 DISCUSSION

It has been suggested that elastic energy storage and return in the foot’s longitudinal arch is an important musculoskeletal mechanism for reducing the energy cost of running \(^{27, 30}\). By examining the effect of restricting arch compression on the energy cost of locomotion, the present study provides among the most direct evidence supporting the energy-sparing spring theory of the arch. We maintain that the arch represents a key passive-elastic energy saving structure in the leg, capable of substantially lowering energy costs of running. This energy saving function of the arch is not greatly affected by running technique (RFS vs. FFS). Importantly, we also show that the arch spring may have an energy saving role that is limited to level running, with little effect on the energy cost of incline running (at a moderate speed) or walking.

4.5.1 THE ENERGY SAVING ROLE OF THE ARCH

The 6% increase in the gross energy cost of level running (7.4% increase in net cost) in the FAI condition represents a considerable increase in locomotor cost arising from manipulation of the longitudinal arch, substantiating earlier interpretations of the metabolic energy sparing
role of the arch in running. Interestingly, despite confirming that arch compression is greater in FFS compared with RFS runners in the shoe only level run (FFS 12.0 mm vs. RFS 9.1 mm; \( p = 0.022 \); see Appendix C) \(^{(41)} \), the increase in metabolic cost resulting from the insole was not statistically different between foot strike groups (HAI - FFS 4.5 ± 5.8%, RFS 4.5 ± 4.6%, \( p = 0.983 \); FAI - FFS 5.9 ± 3.8%, RFS 6.2 ± 4.7%, \( p = 0.908 \)). The estimated reduction in the elastic energy storage of the arch in the FAI condition equalled 8.7% of the total centre of mass mechanical work of running (Figure 4.6B). The relatively low estimate in the present study, when compared with those simulated by Ker et al. \(^{(27)} \) in cadaver specimens, was expected considering the slower running speed and subsequent smaller compressive loads. Interestingly, if the relationship between the lost elastic energy contribution by the arch and the increase in the metabolic cost of running is proportional, the ~17% arch elastic energy contribution predicted by Ker et al. \(^{(27)} \) at 4.5 ms\(^{-1} \) may indeed reduce the metabolic cost by a similar percentage. Unfortunately, our technique for restricting arch compression was limited to slower running speeds, thereby preventing the study from exploring the potentially larger energetic effect of the arch at faster velocities. Nevertheless, the data obtained at slower speeds suggests that the arch likely does play a primary mechanical role as an energy saving structure.

It is notable that the arch spring, unlike tendon structures (e.g. the Achilles tendon; \(^{(22)} \)), achieves elastic energy recycling largely in absence of muscle activity. Kelly et al. \(^{(26)} \) have recently shown that intrinsic foot muscles are capable of controlling arch position. However, considering their relatively small cross sectional size, it is uncertain whether they prevent arch compression/elastic energy storage during running given the high loads experienced. The distinction between the passive and active spring mechanisms is important since the later requires substantial metabolic energy to maintain tension in the spring (cost of force). In this regard, the arch spring may be amongst the most effective energy saving structures, and when compared with other energy contributing components of the lower limb, may be critical.

Because elastic energy storage increases non-linearly with arch loading [see Figure 2B from Ker et al. \(^{(27)} \)] the majority of the elastic energy is stored in the final 20% of arch compression. Our hypothesis that the FAI and HAI would lead to a similar reduction in elastic energy storage and increase in metabolic energy cost was largely supported. Despite the nearly two-fold difference in arch kinematics (FAI 10 mm reduction in compression vs. HAI 6.5 mm reduction; Figures 4.3 & 4.4), both insoles resulted in a similar increase in metabolic cost and loss of arch elastic energy storage (Figures 4.5 & 4.6A). Importantly, these data indicate that for the arch to
function as a useful energy saving mechanism it must undergo nearly full compression. Furthermore, the increase in energy cost while wearing the insoles is likely not simply a systematic effect of alternating arch kinematics _per se_, but rather points to the ensuing effect on the arch elastic function as the primary factor influencing the metabolic cost of running with restricted arch compression.

Our hypothesis that restricting arch compression would have a smaller effect on the energy cost of walking and incline running was also supported. The almost complete lack of an energetic consequence when arch compression was restricted in walking was surprising, but is possibly due, in part, to the benefit gained from increased midfoot rigidity caused by the insole. This may improve the effectiveness of the plantar-flexion torque (²⁵, ⁴⁰) and outweigh the energetic consequence of reduced arch elastic energy return (Figure 4.6A), although to what extent is unknown. While we predicted a smaller percent increase in metabolic cost when arch compression was restricted during incline compared with level running, the small (1.2 %) and statistically non-significant increase was surprising. It remains possible that the lack of increase in metabolic cost during incline running reflects a difference in the effectiveness of the arch spring on sloped surfaces. It is also possible that the force plate method for calculating mechanical work underestimates the additional muscular work required to raise the centre of mass uphill above that required in level running. Given that the additional positive mechanical work of incline running cannot be supplied through the return of stored elastic energy, this scenario could help explain the small energetic effect of the FAI when running uphill.

We do not have a clear explanation for the surprisingly small energetic effect of the FAI when running uphill. Nevertheless, the fact that there was no change in energy cost during both the FAI incline running and walking conditions strengthens the interpretation that the energetic changes observed in level running may be primarily attributed to alterations in the arch spring mechanics. If other general modifications such as co-contraction, instability, cushioning (³⁷) or discomfort (³⁷) were the primary factors leading to the increase in metabolic cost after restricting arch compression, it would be expected that they would also elevate metabolic costs during running in the moderate incline (³°) condition of this study, and possibly also during walking. Instead, the energy costs in these conditions were unaffected. Adding further support to the premise that the insoles primarily altered arch compression, peak ankle eversion during level running was unaffected (see Appendix C) and there was a negligible
difference in total limb mechanical work between the insole and shoe only conditions (Figure 4.6B).

During downhill running, elastic energy returned from the Achilles tendon and arch of the foot make up a large percentage of work done on the centre of mass (46). In fact, at a grade of approximately -9° little to no mechanical work production is required by the lower limb muscular, instead elastic energy return accounts for almost all of the necessary positive mechanical work (46). It could therefore be speculated that restricting arch compression during downhill running may also invoke a metabolic penalty. This is an important area for future research.

4.5.2 IMPLICATIONS FOR ARCH FORM AND FOOTWEAR

The results of this study have implications for individuals with altered arch function such as flat foot deformity (25) and women pre- and postpartum (45). Both are associated with a decreased medial longitudinal arch height, in part due to increased laxity of the arch supporting tendons and ligaments (25, 45). Certain medical conditions also affect the properties of tendons and ligaments, such as diabetes mellitus (1, 39) and various rheumatic diseases (e.g. gout (16)). Subsequently, it is possible that these individuals experience higher metabolic costs during running due to reduced arch elastic energy storage/return, compared with individuals with a ‘normal’ foot posture (43) and function, although this requires further investigation.

Many modern day running shoes have built in arch support aimed to reduce excessive pronation, which is sometimes linked to overuse injuries (33, 49). In addition to supportive footwear, custom or pre-fabricated orthotic insoles are often used in conjunction with conventional running shoes to provide further arch support and prevent arch compression. Whilst these corrective measures can assist the runners gait mechanics and alter tissue loading, it is possible that this added support may hinder the arch’s elastic energy storage potential and subsequently lead to an increase in energy cost. Indeed, a number of studies have reported an increase in the energy cost of running when wearing orthotic insoles (5, 12, 24), although this affect may in part be due to added weight (21). Considering that the major component of the elastic energy storage occurs at the end-range of arch compression, even restricting a small amount of compression via a shoe or orthotic insert, might lead to an energetic consequence during running. We did not specifically examine the threshold at which arch support leads to a physiologically meaningful increase in the energy cost of running,
although it is clear that restricting the last 40% of compression (HAI) will have a substantial effect. Perl et al. (41) found a significant 3% increase in metabolic cost when participants ran in traditional arch supporting running shoes compared with minimalist shoes, even after controlling for strike technique, shoe mass and stride frequency. Although, they did not measure arch compression, they hypothesised that the observed increase in metabolic cost was a result of decreased elastic energy storage arch return from the arch of the foot. The benefits of using corrective footwear and/or orthotics designed with significant arch support should therefore be weighed against their effect on running energetics. It is also intriguing to consider whether spring augmenting footwear (e.g. Adidas Springblade®) or insoles can be utilised to lower the energy cost of running. In contrast to running, our findings suggest that using rigid supportive shoes or insoles that prevent arch collapse, is likely to have little energetic consequence during walking.

4.5.3 EVOLUTION OF THE HUMAN FOOT

The evolution of the longitudinal foot arch is regarded as a key adaptation for obligate hominid bipedalism (23). Although the origin of the arch is debated (1, 4), interpretation of 1.5 million year-old footprints from northern Kenya suggest that the Homo erectus (the most likely maker of the prints) possessed a foot with predominantly modern human characteristics including a medial longitudinal arch (4). The functional significance of the longitudinal arch in the evolution of human bipedal gait has often been attributed to the rigid mid-tarsal lever system allowing effective plantar-flexion during toe-off (9, 18). Recent experimental evidence supports this theory (3), although adjustable (variable) stiffness may characterise arch function more so than a permanently stiff lever system. A complimentary theory surrounding the evolution of the longitudinal arch is that its spring properties lower the energetic cost of endurance running (11, 27). Our study provides some intriguing support for the arch functioning as both a rigid lever in walking and primarily a spring in running. Interestingly, the insoles had no effect on the metabolic cost of walking despite restricting ~80% of arch compression and an estimated 2.5 J of arch elastic energy storage and return. The absence of any energetic difference might result because the insoles enhanced the effect of midfoot rigidity in walking. On the other hand, restricting the arch’s spring function by 11 J in level running resulted in a clear increase in metabolic cost. Further, that we only observed an energetic consequence of arch restriction during level running and not incline running may offer added insight into the movement behaviour and environment of early Homo. The landscape inhabited by early Homo was invariably not limited to horizontal ground (50), although given that the arch only provided an
energetic advantage during level and not incline running may suggest that their food sources \cite{42} were predominantly available in this environment.

4.5.4 LIMITATIONS

The authors acknowledge that this study has a number of limitations. Firstly, we were restricted to assessing the effect of arch compression at a relatively slow running speed. It is predicted that the energetic saving resulting from arch elastic energy storage and return is larger during faster level running, and this might also be the case for incline running.

It is also important to note that although restricting arch compression will reduce the energy storage in the arch spring ligaments and plantar fascia, the plantar fascia can also undergo strain and energy storage due to motion at the MTP joint. In addition, restricting arch compression may also alter intrinsic foot muscle function (although these muscles likely contribute relatively little to the overall changes in energy cost given their small size). Despite these possible factors, measurements of MTP joint motion and net moments displayed minimal difference between the shoe only and FAI/HAI trials (unpublished data \cite{32}), suggesting they may have had a small effect.

A number of limitations also existed in estimating arch compression and elastic energy. A skin mounted marker placed on the navicular tuberosity was used to determine arch compression. While this is one of the most commonly used methods for assessing arch compression (e.g. \cite{10, 19, 26, 41}), it is not without errors and may have under-estimated bony midfoot motion \cite{36}. Our estimate of arch elastic energy storage was intended to only supplement our observations of locomotor energy cost and several factors can affect its accuracy. Perhaps most notably, the prediction depended on an estimate of arch compressive load from inverse dynamic calculation of ankle joint moments and Achilles tendon force that are both subject to error. The Achilles tendon moment arm was measured in a static standing position as the perpendicular distance from the mid-point of the lateral malleoli to the Achilles tendon \cite{29}. While this measurement was performed by the same assessor it is understood that this method may differ from imaging-based measurements and does not represent the active moment arm of the Achilles during locomotion \cite{31}. Thus, although it serves as a crude estimation of energy storage in the arch, the simple modelled load-displacement curve may lead to errors in estimating arch elastic energy storage with a high degree of accuracy.
4.6 CONCLUSION

By examining the effect of restricting arch compression on the metabolic cost of locomotion, this study provides among the most direct evidence supporting the energy-sparing spring theory of the arch. When 79% of arch compression was restricted, gross metabolic cost increased by 6%, supporting interpretations from cadaver studies that the spring function of the arch significantly contributes to the energy cost of running \( ^{(27)} \). However, the energy sparing role of the arch may be limited to level running, given that restricting arch compression did not result in a metabolic penalty during walking or incline running. Additionally, foot strike technique (RFS vs FFS) did not influence the arch’s effect on metabolic cost. The findings of this study not only broaden our understanding of running mechanics and energetics but also have important implications for orthotic prescription, footwear design, foot pathologies, and provide support for the arch’s role in human bipedal evolution.

ACKNOWLEDGEMENTS

The authors would like to acknowledge the following; Orthotech laboratories (Blackburn, Melbourne, Australia) for the manufacture and supply of the insoles used in this study, Dr. Robert Day for his assistance testing the insole material properties, and Tony Roby for constructing the magnetic marker set and the insole material testing rig.

ORTHOTIC CLAUSE

The authors would like to note that the foot insoles used in this study do not represent conventional prescription practices by health practitioners for symptomatic individuals in a clinical setting, but rather a tool for experimentally testing our hypotheses.

CONFLICT OF INTEREST

None of the authors involved in the present study have any conflict of interest, financial, personal or otherwise which would influence this research.
4.7 REFERENCES


32. McDonald KA. The role of arch compression and metatarsophalangeal joint dynamics in modulating plantar fascia strain in running. 2014.


CHAPTER FIVE

HOW DO WE RUN WITHOUT A SPRING IN OUR STEP? ALTERATION IN JOINT MECHANICS FOLLOWING ARCH COMPRESSION RESTRICTION

This manuscript has been prepared for submission to PLOS ONE, September 2014.

Stearne SM, Alderson JA, McDonald KA, North I, Rubenson J. How do we run without a spring in our step? Alteration in joint mechanics following arch compression restriction. PLOS ONE, intended submission September 2014.

The PhD candidate, Sarah M. Stearne, accounted for 80% of the intellectual property associated with the final manuscript. Collectively, the remaining authors contributed 20%. 
CHAPTER FIVE: STUDY THREE
HOW DO WE RUN WITHOUT A SPRING IN OUR STEP?

LINKING STATEMENT

The previous chapter (Study Two) of this thesis provides among the most direct evidence supporting the energy-sparing spring theory of the arch. When ~80% of arch compression was restricted, gross metabolic cost increased by 6%, supporting interpretations from cadaver studies that the spring function of the arch significantly contributes to the energy cost of running \(^{(10)}\). Given the arch’s important contribution to positive mechanical work, restricting its function is likely to elicit compensatory mechanical responses in the lower limb due to the loss of propulsive work. Study Two revealed that different foot strike technique (rearfoot vs forefoot) did not alter the metabolic consequence of restricting arch compression. However, Study One identified significant disparities in lower limb kinetics between rearfoot and forefoot strike runners. It is therefore possible that runners utilising different foot strike techniques employ varied compensatory mechanisms when elastic energy storage/return in the foot’s arch is restricted. This study aims to identify the compensatory mechanics of habitual rearfoot and forefoot strike runners to restricted arch compression and by extension uncover the structures (i.e. joints) in which mechanical demand may be reduced when the arch spring functions normally, and elevated when the arch function is impaired. The findings of this study will not only broaden our understanding of the lower limb structure-function relationship, but also have important implications for orthotic prescription, footwear design, and individuals with foot pathologies.

The data presented in this chapter was collected in conjunction with Study Two. This thesis is presented as a series of stand-alone studies, we therefore acknowledge that the methodology of this chapter may appear somewhat repetitive.
5.1 ABSTRACT

**Purpose:** By employing a novel insole approach, this study aimed to discover how lower limb joint mechanics compensate when compression of the foot’s medial longitudinal arch is restricted during running, subsequently reducing the mechanical work provided by the storage and release of arch elastic energy (Study Two [27]). A secondary aim was to investigate whether rearfoot strike (RFS) and forefoot strike (FFS) runners employ the same or different compensatory mechanisms. **Methods:** Eight habitual RFS and nine habitual FFS runners ran on an instrumented treadmill at 2.7 ms\(^{-1}\) while three-dimensional force and kinematics data were collected. Two testing conditions were performed; 1) shoe only, 2) shoe with custom-made subject-specific rigid insoles that restricted arch compression during running by ~80%. **Results:** Inverse dynamic analysis uncovered differences in joint-level mechanical work when arch compression was restricted. Positive and negative mechanical work at the knee was increased (15.8 ± 24.9% and 20.2 ± 28.8% respectively), and positive and negative mechanical work at the ankle decreased (-7.8 ± 9.6% and -10.0 ± 12.7% respectively). The combined joint positive mechanical work from the hip, knee and ankle across the stride was minimally reduced (-4.2 ± 6.8%) after arch compression was restricted. **Conclusions:** Contrary to our hypothesis, restricting arch compression does not result in a compensatory increase in total joint work. Nevertheless, the switch from ankle to knee-based running likely contributes to the increase in metabolic cost observed by Stearne et al. (Study Two [27]) by requiring a larger proportion of the total positive and negative mechanical work to be performed by muscle fibres due to a reduced reliance on the arch and Achilles tendon elastic energy storage and return. Alteration of muscle efficiency may also have increased metabolic cost. Foot strike did not influence the compensatory mechanisms employed to overcome the reduced arch compression/recoil. These findings have important implications for individuals with impaired arch function and orthotic insole, shoe and prosthetic design.
5.2 INTRODUCTION

Measurements on cadaver specimens have found that the longitudinal arch of the human foot possesses spring-like qualities owing to the elastic nature of its tendons and ligaments \(^{10}\). Recently, Stearne and colleagues (Study Two \(^{27}\)) provided experimental data supporting the theory that the spring function of the arch lowers the metabolic cost of running. Using custom orthotic insoles designed to restrict compression of the arch during running, Stearne et al. (Study Two \(^{27}\)) found that the percent increase in metabolic cost during level running was comparable to the estimated loss in elastic arch energy when expressed as a percent of the total lower limb mechanical cost.

While these studies point strongly to an energy saving function of the longitudinal arch, what remains unclear is the impact returned elastic energy has on mechanical work required by other proximal lower limb joints. For example, if the spring function of the arch is disabled, how is the production of mechanical work modulated at the hip, knee and ankle, the other major joint structures powering running? The answer to this question is important if we are to obtain a fundamental understanding of how the longitudinal arch impacts running mechanics and energetics, and is of practical value for understanding how running mechanics are compensated in conditions where the arch is impaired [e.g. flatfoot deformity \(^{28}\), rigid pes cavus \(^{7}\), midfoot break \(^{15}\), plantar fasciotomy \(^{24}\) or pre- and postpartum \(^{23}\)]. Further, corrective rigid and semi rigid insoles (orthotics) may also affect arch compression and its capacity to store and return elastic energy \(^{6, 11}\). Quantifying how such modification of the foot’s arch spring mechanism might impact overall lower limb joint mechanics is important for understanding foot pathologies and corrective treatments, and can provide valuable information for footwear, orthotic and prosthetic designers.

The aim of this study was to determine how the joint mechanics of the lower limb compensate when arch compression is inhibited and subsequently reduces the mechanical work provided by the storage and release of arch elastic energy \(^{27}\). A secondary aim was to investigate rearfoot strike (RFS) and forefoot strike (FFS) runners compensatory patterns when arch function is compromised given that these foot strike techniques display varied joint mechanical work profiles (Study One \(^{26}\)). These questions were addressed using customised, subject-specific insoles, designed to restrict the majority of arch compression occurring during level running. We hypothesised that when arch compression was restricted runners would increase their total joint work production in order to overcome a loss of arch elastic work. We
further hypothesised that the ankle joint would contribute the majority of the additional positive work, due to its significant role in producing the mechanical work in running \(^8, 26\). Finally, we hypothesised that the increase in positive work at the ankle would be greater in RFS compared with FFS runners due to the already large role the ankle plays in a FFS (Study One \(^{26}\)) and we hypothesised that in addition to an increase at the ankle, FFS runners would thus also utilise the hip joint as source of additional propulsion.

5.3 METHODS

For a more detailed methodology refer to Stearne et al. (Study Two \(^{27}\)). The insole used in the current study is the full arch insole (FAI) used in Stearne et al. (Study Two \(^{27}\)).

5.3.1 PARTICIPANTS

Eight habitual RFS and nine habitual FFS recreational male runners were recruited to participate in the study. Foot strike technique was confirmed using high-speed video recording at 300 frames/sec positioned in the sagittal plane (Casio EXILIM EX-F1, Casio Computer Co. LTD., Shibuya-ku, Tokyo). The participants had not experienced any lower limb injuries in the six months prior to testing nor presented with any pre-existing gait abnormalities. No significant differences in measured physiological or fitness variables existed between foot strike groups as determined by a series of independent t-tests (age- RFS 25.5 ± 4.4 years, FFS 27.6 ± 3.4 years; height- RFS 185.3 ± 6.9 cm, FFS 181.8 ± 4.8 cm; weight- RFS 79.4 ± 6.8 kg, FFS 75.7 ± 5.9 kg; weekly running distance- RFS 39.4 ± 21.1 km, FFS 42.2 ± 36.0 km; mean ± SD). Participants did not regularly wear prescriptive orthotic insoles and were deemed to have normal foot structure as determined by the Foot Posture Index \(^{18}\) (RFS 1.4 ± 1.4, FFS 1.0 ± 2.8) measured by a single experienced clinician (I.N.). Measured foot variables did not differ between RFS and FFS groups (foot length- RFS 277.0 ± 11.4 mm, FFS 272.4 ± 6.1 mm; resting arch height- RFS 50.3 ± 7.3 mm, FFS 49.5 ± 9.5 mm; Achilles tendon moment arm- RFS 45.3 ± 4.1 mm, FFS 46.8 ± 5.4 mm). Participants provided written, informed consent prior to inclusion in the study. All procedures were approved by The University of Western Australia (UWA) Human Research Ethics Committee (Approval ID: RA/4/1/4541, Appendix A).
5.3.2 Custom Arch Restricting Insoles

A pair of custom insoles was fabricated for each participant using three-dimensional (3D) scans (ScanAny, Orthotech laboratories, Blackburn, Melbourne) of the feet in a non-weight bearing subtalar neutral position. The insoles were made from four millimetre polypropylene, possessed high density arch fill (shore value ~350-400), a four degree intrinsic rear foot grind, a balanced fore foot, maximum arch congruency and the heel was ground to less than one millimetre such that heel-toe drop was deemed negligible. The insoles were designed to completely fill the participant’s arch, when the foot was positioned in a neutral non-weight bearing position, theoretically allowing minimal arch compression. Participants were provided the insoles two weeks prior to testing to become familiar wearing them. For a description of the insole material properties refer to Study Two (27).

5.3.3 Testing Conditions

All participants wore New Balance Minimus road MR00 shoes for testing (approximate weight of 180 grams, zero heel to toe drop, no medial posting and a uniform EVA midsole). Neoprene pockets filled with lead weights were affixed to the laces of both shoes in order to standardise foot weight across all shoe and insole conditions. Prior to testing commencement, participants completed a five minute warm-up on a force-plate instrumented split belt treadmill (Bertec Corporation, Columbus OH, USA). Participants performed two concurrent data collection sessions on the instrumented treadmill at 2.7 ms⁻¹ 1) shoe only 2) shoe with custom insoles. Each data collection lasted approximately five minutes with the condition order randomised to account for fatigue effects. Six consecutive right leg strides from the final minute were selected for analysis. Following each data collection session participants completed a brief questionnaire regarding fatigue and discomfort levels (see Stearne et al. (Study Two (27) for details). Rate of oxygen consumption and arch compression data were collected as per Stearne et al. (Study Two (27)).

5.3.4 3D Joint Kinematics and Kinetics

Retro-reflective markers were affixed to the lower limbs in accordance with Besier et al. (2), with additional markers placed on the navicular tuberosity and distal phalanx of the first metatarsal. The shoe upper was modified to allow markers to be placed directly on the foot and remain visible. To ensure marker positions remained unchanged after removing shoes, all foot markers were detachable by using a magnetic base that did not change location.
Motion of the retro-reflective markers was tracked using a ten-camera near infrared Vicon T-series 3D motion capture system operating at 250 Hz (T40S model, Oxford Metrics, Oxford UK).

Joint kinematic and kinetic profiles for the ankle, knee and hip were computed using 3D inverse-dynamic gait analysis in accordance with Besier et al., \(^2\) using Vicon BodyBuilder software (Oxford Metrics, Oxford UK). Ground reaction forces from the instrumented treadmill were recorded at 2,000 Hz, and synchronised with the kinematic data using a Vicon MX-Net control box (Oxford Metrics, Oxford UK). Ankle joint centres were defined as the mid-point between the medial and lateral malleoli anatomical landmarks. A six-marker pointer was used to identify the medial and lateral femoral condyles and a mean instantaneous knee helical axis was used to define the knee joint centres and corresponding knee anatomical coordinate systems. A functional method was also used to define the hip joint centres \(^2\). A custom foot alignment rig was used to measure calcaneus inversion/eversion and foot abduction/adduction to assist in defining the anatomical coordinate system of the foot segments. Joint coordinate systems were defined in accordance with ISB standards of wu et al. \(^29\). All marker trajectories were filtered using a zero-lag 4th order low pass Butterworth filter with cut-off frequencies typically at 14 Hz, which was determined by a custom residual analysis algorithm for each participant (MATLAB, The MathWorks Inc., USA). Ground reaction forces were filtered at the same cut-off frequency as the kinematic data to mitigate any artefacts in joint moments arising due to un-accounted segment acceleration \(^3, 12\).

Net hip, knee and ankle joint moments and instantaneous power were calculated using inverse dynamics in BodyBuilder software (Oxford Metrics, Oxford UK). Data from the right leg were separated into stance and swing phases (as defined by a ground reaction force threshold of 10 N). Instantaneous joint power was normalised to body mass and integrated with respect to time to compute net positive and negative mechanical work (J kg\(^{-1}\)), respectively:

\[
W_j^{pos} = \int_{t_i}^{t_f} P_j \, dt \quad \text{for} \quad P_j > 0 \\
W_j^{neg} = \int_{t_i}^{t_f} P_j \, dt \quad \text{for} \quad P_j < 0
\]  

[Equation 5.1]

[Equation 5.2]

Where \(j\) represents the joint (ankle, knee or hip), \(P\) the joint power, and \(t_i\) and \(t_f\) represent the start and end time of the integration respectively. Positive and negative work at each joint was computed separately for the stance and swing phases of six strides and then averaged. The total positive and total negative combined limb work of the entire stride was subsequently computed as the sum of each joint’s stance and swing work production.
The percent each joint contributed to total stance positive and negative lower limb work was calculated by dividing the individual joint work during stance by the total stance work.

5.3.5 STATISTICAL ANALYSIS

Two-way repeated measures split-plot analysis of variance (ANOVA’s) were performed for each of the ankle, knee and hip joints to determine the effect of the custom insoles and foot strike technique on positive and negative mechanical work. Two-way repeated measures ANOVA’s were also performed on; arch compression, metabolic cost, mechanical stance work, swing work and total stride work. Finally, following an arcsin transformation, two-way mixed-model multivariate analyses of variance (MANOVA’s) were conducted to determine the effect of the insoles and foot strike technique on each joint’s (ankle, knee and hip) percentage contribution to total stance phase lower limb positive and negative mechanical work. For all analyses, within subject factors were shoe only and insole conditions, with between subject factors habitual foot strike group, RFS or FFS. Where significant interaction effects were found post-hoc t-tests were performed to determine the location of the relationship. Paired samples t-tests were conducted for within foot strike group comparisons and independent samples t-tests for between groups. Significance was set at \( \alpha < 0.05 \).

5.4 RESULTS

5.4.1 TEMPORAL PARAMETERS

During level running no main effect of insole on stance (\( p = 0.551 \)), swing (\( p = 0.126 \)) or stride time (\( p = 0.197 \)) was observed. There was a significant main effect of foot strike technique on stance time (\( p = 0.014 \)) with RFS runners spending longer in stance than FFS runners in both; shoe only (RFS 0.30 ± 0.03 sec, FFS 0.28 ± 0.03 sec) and insole running conditions (RFS 0.30 ± 0.03 sec, FFS 0.26 ± 0.02 sec). There was no main effect of foot strike technique on swing (\( p = 0.079 \)) or stride time (\( p = 0.512 \)).

5.4.2 ARCH RESTRICTION AND METABOLIC COST

As per the previous study by Stearne et al. (Study Two \(^{27}\)), insoles had a significant main effect on maximum arch compression (\( p < 0.001 \)) and metabolic cost (\( p < 0.001 \)). There was no main effect of foot strike technique or an interaction effect in either analysis. Arch compression was reduced by an average of 78.9 ± 24.7% (RFS 84.7 ± 22.0%, FFS 73.0 ± 27.3%; average shoe only
10.6 ± 2.6 mm, average insole 0.6 ± 6.0 mm) and the gross metabolic cost increased by an average of 6.0 ± 4.2% (RFS 6.2 ± 4.7%, FFS 5.9 ± 3.8%; average shoe only 36.5 ± 2.9 ml kg$^{-1}$ min$^{-1}$, average insole 38.7 ± 3.4 ml kg$^{-1}$ min$^{-1}$) when wearing the insole compared with shoe only running.

5.4.3 JOINT MOMENTS

The insole had a significant main effect on both ankle and knee joint moments; $p < 0.001$ and $p = 0.002$ respectively. Peak plantar-flexion moments about the ankle were significantly reduced in both foot strike groups (combined average -7.0 ± 4.1%; Figure 5.1) when arch compression was restricted. At the knee, peak extension moment was significantly increased in both groups (7.2 ± 9.7%; Figure 5.1), there was also a significant interaction effect ($p = 0.004$) due to the knee moment in FFS runners increasing by 14.5 ± 7.4% but only by 0.8 ± 6.4% in RFS runners. No differences in peak moment were identified at the hip joint (main effect $p = 0.220$).

**Figure 5.1** Ankle and knee net joint moments (N m kg$^{-1}$) during the stance phase in shoe only (solid line) and insole (dashed line) conditions. Data represents the average of rearfoot and forefoot strike groups. * indicates significant difference in peak joint moment between shoe only and insole $p < 0.05$. 
Table 5.1 Lower limb positive and negative mechanical work during shoe only and insole running conditions in rearfoot strike (RFS), forefoot strike (FFS) and both foot strike groups combined (Group).

<table>
<thead>
<tr>
<th>MECHANICAL WORK (J kg⁻¹)</th>
<th>Forefoot Strike</th>
<th>Rearfoot Strike</th>
<th>Group</th>
<th>Main Effect p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Shoe Insole</td>
<td>Shoe Insole</td>
<td>Shoe Insole</td>
<td>Insole Foot Strike</td>
</tr>
<tr>
<td>Ankle Work Stance Pos</td>
<td>1.04 ± 0.16</td>
<td>0.93 ± 0.14</td>
<td>0.87 ± 0.16</td>
<td>0.82 ± 0.14</td>
</tr>
<tr>
<td></td>
<td>± 0.15 ± 0.13</td>
<td>± 0.10 ± 0.11</td>
<td>± 0.10 ± 0.11</td>
<td>± 0.11 ± 0.13</td>
</tr>
<tr>
<td>Neg</td>
<td>-0.53 ± 0.47</td>
<td>-0.38 ± 0.35</td>
<td>-0.46 ± 0.41</td>
<td>-0.41 ± 0.41</td>
</tr>
<tr>
<td></td>
<td>± 0.15 ± 0.13</td>
<td>± 0.10 ± 0.11</td>
<td>± 0.10 ± 0.11</td>
<td>± 0.11 ± 0.13</td>
</tr>
<tr>
<td>Knee Work Stance Pos</td>
<td>0.30 ± 0.08</td>
<td>0.33 ± 0.07</td>
<td>0.31 ± 0.07</td>
<td>0.36 ± 0.09</td>
</tr>
<tr>
<td></td>
<td>± 0.09 ± 0.11</td>
<td>± 0.08 ± 0.08</td>
<td>± 0.08 ± 0.08</td>
<td>± 0.09 ± 0.10</td>
</tr>
<tr>
<td>Neg</td>
<td>-0.29 ± 0.37</td>
<td>-0.38 ± 0.42</td>
<td>-0.33 ± 0.39</td>
<td>-0.39 ± 0.39</td>
</tr>
<tr>
<td></td>
<td>± 0.09 ± 0.15</td>
<td>± 0.12 ± 0.10</td>
<td>± 0.12 ± 0.10</td>
<td>± 0.13 ± 0.13</td>
</tr>
<tr>
<td>Hip Work Stance Pos</td>
<td>0.19 ± 0.08</td>
<td>0.18 ± 0.08</td>
<td>0.18 ± 0.08</td>
<td>0.18 ± 0.08</td>
</tr>
<tr>
<td></td>
<td>± 0.06 ± 0.04</td>
<td>± 0.08 ± 0.10</td>
<td>± 0.08 ± 0.10</td>
<td>± 0.08 ± 0.08</td>
</tr>
<tr>
<td>Neg</td>
<td>-0.12 ± 0.13</td>
<td>-0.18 ± 0.19</td>
<td>-0.15 ± 0.16</td>
<td>-0.16 ± 0.16</td>
</tr>
<tr>
<td>Stance Work Total Pos</td>
<td>1.52 ± 0.21</td>
<td>1.38 ± 0.21</td>
<td>1.46 ± 0.22</td>
<td>1.41 ± 0.23</td>
</tr>
<tr>
<td></td>
<td>± 0.21 ± 0.21</td>
<td>± 0.21 ± 0.25</td>
<td>± 0.22 ± 0.25</td>
<td>± 0.23 ± 0.23</td>
</tr>
<tr>
<td>Neg</td>
<td>-0.95 ± 0.17</td>
<td>-0.94 ± 0.19</td>
<td>-0.94 ± 0.17</td>
<td>-0.97 ± 0.23</td>
</tr>
<tr>
<td></td>
<td>± 0.25 ± 0.25</td>
<td>± 0.25 ± 0.22</td>
<td>± 0.25 ± 0.22</td>
<td>± 0.26 ± 0.26</td>
</tr>
<tr>
<td>Swing Work Total Pos</td>
<td>0.50 ± 0.09</td>
<td>0.66 ± 0.25</td>
<td>0.57 ± 0.19</td>
<td>0.53 ± 0.10</td>
</tr>
<tr>
<td></td>
<td>± 0.11 ± 0.08</td>
<td>± 0.08 ± 0.08</td>
<td>± 0.10 ± 0.10</td>
<td>± 0.10 ± 0.10</td>
</tr>
<tr>
<td>Neg</td>
<td>-0.53 ± 0.52</td>
<td>-0.52 ± 0.54</td>
<td>-0.52 ± 0.53</td>
<td>-0.53 ± 0.53</td>
</tr>
<tr>
<td></td>
<td>± 0.08 ± 0.08</td>
<td>± 0.07 ± 0.07</td>
<td>± 0.07 ± 0.07</td>
<td>± 0.07 ± 0.07</td>
</tr>
<tr>
<td>Stride Work Total Pos</td>
<td>2.02 ± 0.23</td>
<td>2.04 ± 0.40</td>
<td>2.03 ± 0.31</td>
<td>1.94 ± 0.27</td>
</tr>
<tr>
<td></td>
<td>± 1.99 ± 0.26</td>
<td>± 1.88 ± 0.27</td>
<td>± 0.31 ± 0.27</td>
<td>± 0.27 ± 0.27</td>
</tr>
<tr>
<td>Neg</td>
<td>-1.47 ± 0.21</td>
<td>-1.45 ± 0.22</td>
<td>-1.46 ± 0.21</td>
<td>-1.50 ± 0.26</td>
</tr>
<tr>
<td></td>
<td>± 1.50 ± 0.29</td>
<td>± 1.49 ± 0.24</td>
<td>± 0.21 ± 0.26</td>
<td>± 0.26 ± 0.26</td>
</tr>
</tbody>
</table>

* indicates significant difference p < 0.05

5.4.4 POSITIVE MECHANICAL WORK

There was a significant main effect of insole on positive mechanical work at both the ankle (p =0.003) and knee joints (p = 0.027; Table 5.1). Positive ankle work decreased in both foot strike groups by an average of -7.8 ± 9.6% and positive knee work increased by an average of 15.8 ± 24.9% in both groups (Figure 5.2A, Table 5.1). No significant differences were found at the hip joint (insole main effect p = 0.576). There was no main effect of insole on total (sum of the ankle, knee and hip) stance (p = 0.066) or swing phase (p = 0.214) positive mechanical work (Table 5.1). However, a slight decrease in work during both stance (-3.2 ± 6.3%) and swing (-3.3 ± 20.7%) when wearing the insole resulted in a significant main effect of insole (p = 0.018; Table 5.1) on total positive joint work across the stride. During the insole condition total stride joint work was decreased by an average of -4.2 ± 6.8% in both foot strike groups compared with shoe only running (Figure 5.2C).
The insole had a significant main effect ($p = 0.031$) on the percent each joint contributed to total positive stance work. *Post hoc* tests revealed that the insole significantly decreased the ankle contribution by an average of -4.9 ± 5.9% ($p = 0.002$; Figure 5.3) and increased the contribution from the knee by an average of 20.2 ± 27.2% ($p = 0.014$; Figure 5.3). There was no difference at the hip joint nor was there a main effect of foot strike group or an interaction effect in any of the positive work variables assessed.

5.4.5 **NEGATIVE MECHANICAL WORK**

Negative mechanical work was also significantly affected by the insole with a main effect at both the ankle and knee; $p = 0.012$ and $p = 0.015$ respectively (Table 5.1). Negative work at the ankle was reduced in both foot strike groups by an average of -10.0 ± 12.7% and negative work at the knee was increased in both groups by an average of 20.2 ± 28.8% (Figure 5.2B). A significant main effect of foot strike ($p = 0.032$) was also present at the ankle joint due to FFS runners absorbing significantly more work than RFS runners in both the shoe only and insole conditions (Table 5.1), however, the percent increase in work due to the insole was similar in both groups (FFS 10.3 ± 15.0%, RFS 9.7 ± 10.6%). No significant main effect of insole or foot strike group on negative mechanical work at the hip joint or for total stance, total swing or total joint work across the stride was detected (Table 5.1). There were no significant interaction effects for any of the negative work variables assessed.

There was a main effect of both the insole ($p = 0.010$) and foot strike group ($p = 0.006$) on the percent each joint contributed to total negative stance work. *Post hoc* tests revealed that the insole caused a similar percent decrease in the ankle contribution to negative work in both foot strike groups (FFS -11.9 ± 14.4%, RFS -10.5 ± 8.9%; insole effect $p = 0.004$), but the contribution from the ankle remained larger in FFS compare with RFS runners (foot strike effect $p = 0.001$). The insole also had a significant effect on the percent contribution to negative work from the knee joint (insole effect $p = 0.006$; Figure 5.3) with both foot strike groups increasing the knee contribution. However, while RFS runners demonstrated a consistently greater reliance on the knee to absorb negative work (foot strike effect $p = 0.013$), FFS runners were more affected by the insole, increasing the percent contribution from the knee by 23.9 ± 26.1% compared with 10.3 ± 12.7% for the RFS technique. See Appendix C for supplementary foot strike percentage joint contribution results.
Figure 5.2 A) Change in positive and B) negative ankle, knee and hip mechanical work (J kg\(^{-1}\)) during stance between shoe only and insole conditions. C) Change in total positive stance, swing and stride mechanical work (J kg\(^{-1}\)) between shoe only and insole conditions. Positive values indicate an increase in work when the insole was worn compared to the shoe only condition and negative values indicate a decrease. Values are average ± standard error of rearfoot and forefoot strike groups combined. * indicates significant difference between shoe only and insole (p < 0.05).
CHAPTER FIVE: STUDY THREE

HOW DO WE RUN WITHOUT A SPRING IN OUR STEP?

5.5 DISCUSSION

Stearne et al. (Study Two \(^\text{(27)}\)) recently reported that restriction of the foot’s longitudinal arch during running resulted in a significant increase in metabolic cost. This finding provides some of the most direct evidence during locomotion to support the long-standing theory by Ker et al. \(^\text{(10)}\) that arch compression and the ensuing elastic energy storage and return in the tendons and ligaments of the foot’s arch, can reduce the metabolic cost of running. Nevertheless, how the arch spring mechanism affects joint mechanics during running remains unclear. The present study identifies that mechanical compensation occurs when arch compression is restricted, including an unexpected redistribution of mechanical work from the ankle to the knee joint. A shift from ankle to knee-based running may explain the increased metabolic cost associated with restricted arch compression (Study Two \(^\text{(27)}\)), possibly due to a reduction in Achilles tendon elastic energy storage/return and altered muscular efficiency.

Figure 5.3 Ankle (black), knee (grey) and hip (white) joint percentage contribution to total positive (A & B) and negative (C & D) mechanical work during the stance phase in shoe only (A & C) and insole (B & D) conditions. Data represents the average from rearfoot and forefoot strike runners. * indicates significantly different to shoe only (\(p < 0.05\)).
5.5.1 Total Limb Mechanical Work

Our hypothesis that total positive lower limb mechanical work would increase to compensate for the loss of elastic energy contribution from the arch was not met. Instead, gait mechanics were altered in such a way that total positive stance work remained unchanged and unexpectedly, total stride work decreased slightly by 4.2% during insole running. Lower limb elastic structures, such as the Achilles tendon and arch of the foot, contribute to positive work by recycling a large proportion of negative work into positive work \(^1\). If these elastic structures are compromised, work must be produced by muscle which comes at a greater metabolic cost. It is therefore possible for mechanical work values to remain unchanged but the cost of producing this work to increase. This concept is also supported by the work of Ortega and Farley \(^{16}\), who found that reducing centre of mass displacement during walking resulted in a significant increase in energy cost, despite net external mechanical work remaining unchanged. The results of this study suggests that rather than an increase in total work production, the greater metabolic cost observed when arch compression is restricted, may instead be attributable to changes in joint work distribution.

5.5.2 Distribution of Joint Work

When the elastic spring function of the arch was restricted, both positive and negative mechanical work was reduced at the ankle and increased at the knee. This contradicted our second hypothesis that an increase in ankle work would occur after restricting arch compression. Similarly, the percent the ankle and knee contribute to total stance work was decreased and increased respectively (both positive and negative). The reduction in mechanical work at the ankle was surprising, but may be explained by the complex interaction between longitudinal arch mechanics and Achilles tendon loading. One of the major components necessary to compress the arch is force in the Achilles tendon, which causes a large compressive force on the talus \(^{10}\). It is possible that in the insole condition high forces in the Achilles tendon are avoided since they are no longer effective in contributing to arch compression. This theory is supported by the statistically significant reduction in ankle plantarflexion moments during stance in the insole compared with the shoe only condition (Figure 5.1). It was also estimated that peak Achilles tendon force decreased by 5.5% \((p = 0.002)\) when arch compression was restricted (computed using peak ankle joint moments and measured Achilles tendon moment arm during standing). Reduced Achilles tendon load with orthotic insole intervention was also found in a recent study by Sinclair et al. \(^{25}\). The lower Achilles
force and ankle plantar-flexion joint moments may partially explain the reduction in ankle work when wearing the insoles.

Importantly, the reduction in ankle work during the insole condition may represent a reduction in reliance on elastic energy contribution to overall mechanical work. The Achilles tendon is known to function as an effective spring during gait, recycling a large component of stored negative work into positive work \(^1\). During moderate speed running the Achilles is hypothesised to store approximately 35 joules of energy, which equates to approximately 35% of the total mechanical cost of stance \(^10\). Given the elastic mechanism by which the Achilles tendon functions, any compensatory positive work produced elsewhere in the lower limb possibly incurs a greater metabolic cost since the mechanical work is likely produced by muscle tissue as opposed to storage/return of energy in tendon that can be achieved using less metabolically expensive force production.

In this study, the knee is the primary joint that compensates for the loss of arch elastic energy; when compression of the arch was restricted, both negative (during the first half of stance) and positive (during the second half of stance) mechanical work at the knee significantly increased, by 20.2% and 15.8% respectively. This increase was attributed to greater knee extension moments as joint excursion remained unchanged. The knee musculature normally plays a minor propulsive role in running (Study One \(^26\)), instead acting as the critical musculature group to absorb the impact of landing and support the body during the weight acceptance phase of stance \(^9, 26\). In comparison to the passive-elastic method of producing positive work utilised at the ankle, positive work production at the knee during the propulsion phase of running predominantly requires concentric contraction of the quadriceps musculature \(^21\). Concentric contractions not only demand a higher energy cost than isometric and eccentric contractions \(^20\) but also elicit greater fatigue \(^17\). It is therefore feasible that the metabolic cost of running increased with the shift from ankle to knee-based running when the arch spring was restricted. Adding support to this, Sawicki \(^22\) concluded that due to varied muscle fibre architecture and elastic elements, re-distribution of positive work production from distal (ankle) to proximal joints, results in an increase in the metabolic cost during walking.

The changes in knee mechanics that compensate for the reduced elastic spring function of the arch may also place a runner at greater risk of knee and quadriceps injury. The increase in negative work indicates greater eccentric muscle contractions. Eccentric contractions place the
muscle at greater risk of exercise-induced injury; micro-tears which lead to inflammation and pain (5). Positive work produced about the knee transmits force through the quadriceps/patella tendon; an increase in tendon force may place a runner at greater risk of quadriceps/patella tendonitis (14, 19) and Osgood-Schlatter’s disease (4). Restricting arch compression therefore not only increases the metabolic cost of running but also may also be detrimental to the quadriceps/patellofemoral musculoskeletal structures.

In addition to a greater reliance on muscle-tissue derived mechanical work (knee versus ankle work), the insole condition might also decrease muscle efficiency, in particular in the plantarflexors. The predicted lower forces in the Achilles tendon will result in lower tendon strain and thus potentially larger muscle fibre strain and velocity. This alteration in fibre mechanics has the potential to alter the operating length, contraction type (isometric vs eccentric vs concentric) and/or contraction velocity of the triceps surae, which are known to impact muscle efficiency (13). More accurate measurement techniques such as ultrasound are required to quantify changes within the muscle tendon unit and to what extent these alterations may affect energy expenditure. However all the above mentioned changes in muscle fibre mechanics present plausible factors contributing to the observed metabolic cost increase.

5.5.3 IMPLICATIONS FOR ARCH IMPAIRMENT

The current study provides new insight into how individuals with arch impairment may differ in their lower limb mechanics compared with runners who utilise a significant elastic energy contribution from the arch of the foot. In addition to experiencing a higher metabolic cost, runners with altered arch function such as flatfoot deformity (28), rigid pes cavus (7), midfoot break (15), plantar fasciotomy (24) and women pre- and postpartum (23), may rely more on the knee to produce positive work. It is also possible that they absorb greater negative work at the knee and offload the Achilles tendon. This is the first study to investigate the implications of reduced arch compression and elastic energy storage on lower limb mechanical work and the information gained should be considered when interpreting gait dysfunction in individuals with arch impairment.

This study presents foundation data surrounding how restricting arch compression alters lower limb mechanics, enabling runners to make informed decisions regarding the merits/detrims of arch support interventions (e.g. orthotic insoles). The implications of arch support (e.g. altered foot biomechanics) should be weighed against the possible increase in metabolic cost.
and positive and negative knee work. Runners who choose to adopt an insole intervention that significantly alters arch compression function, may benefit from first implementing a concentric and eccentric quadriceps strength training program to prepare the musculature for higher loading and reduce injury risk predisposition. It should be noted that this study examined only non-injured individuals with neutral feet, therefore the impact of restricting arch function in a population with lower limb injuries and/or varied foot/arch form may differ and requires further investigation.

5.5.4 FOOT STRIKE TECHNIQUE

RFS and FFS runners compensated in the same manner when arch compression was restricted, namely by decreasing ankle and increasing knee positive and negative work. Although, perhaps owing to their typically greater dependence on the ankle for mechanical work production and absorption (Study One [26]), the magnitude of change at both the ankle and knee was larger in FFS runners. Despite the large decreases in work production/absorption at the ankle when arch compression was restricted, FFS runners continued to absorb and produce more ankle work than RFS runners. During shoe only running RFS runners absorbed and produced more negative work at the knee, however restriction of the arch resulted in FFS runners utilising the knee to produce greater positive work. Our hypothesis that FFS runners would produce a portion of additional positive work at the hip was not supported, as no significant differences were observed at the hip joint for any assessed variable.

5.6 CONCLUSION

The arch of the foot provides an important source of elastic energy during running and its restriction results in a significant increase in metabolic cost (Study Two [10, 27]). This study investigated the impact that restricting arch compression had on lower limb mechanical work and found that the knee compensated for the loss of arch elastic energy by producing additional positive work during stance. There was also a shift in both positive and negative work from the ankle to the knee that likely contributed to an increase in metabolic cost resulting from decreased reliance on elastic work production and decreased triceps surae muscle efficiency. Both RFS and FFS runners compensated for a restricted arch spring is in a similar manner. The results of this study not only provide useful information regarding the structure-function relationship of the lower limb, but also have practical implications for individuals with impaired arch function and individuals adopting arch support insoles during running.
CHAPTER FIVE: STUDY THREE
HOW DO WE RUN WITHOUT A SPRING IN OUR STEP?

ACKNOWLEDGEMENTS

The authors would like to acknowledge Willetton Podiatry (Willetton, Perth, Australia) for allowing participants to attend their clinic for insole fitting and Orthotech laboratories (Blackburn, Adelaide, Australia) for supplying the insoles. The authors would also like to acknowledge Dr. Robert Day from The Mechanical Engineering Department at Royal Perth Hospital for his time and use of their Instron machine to determine the material properties of the insoles. Finally, the authors would like to acknowledge the technical staff, in particular Tony Roby from The School of Sport Science, Exercise and Health at The University of Western Australia, for their valuable assistance particularly in the creation of a magnetic marker set.

CONFLICT OF INTEREST

None of the authors involved in the present study have any conflict of interest, financial, personal or otherwise which would influence this research.

5.7 REFERENCES


6.1 Executive Summary

The number of people participating in distance running has significantly increased in the last 40 years \(^{(28)}\). With this comes the desire for improved running performance and a larger number of people experiencing running related overuse injuries \(\text{(the prevalence of running injuries has been reported between } 19.4-79.3\% \text{)}^{(14, 44)}\). Foot function has been identified as an important factor influencing running health and performance, possibly in part because it can affect not only the foot itself but also both lower limb joint mechanics and whole body mechanics and energetics. Two aspects of the foot that play a key role in running are the angle at which the foot makes initial contact with the ground in the sagittal plane (foot strike technique) and the function of its longitudinal arch. This research, presented as a series of papers, investigated the influence of these two aspects of foot function on the mechanics and energetics of running. The thesis explored both the individual effect of foot strike and arch mechanics as well as their integrated function. An overview and summary of the key findings and conclusions are outlined in this chapter.

Study One (Chapter Three) aimed to determine how the lower limb joint kinetics of habitual rearfoot and habitual forefoot strike runners differ and also explored the consequences of switching to an imposed (non-preferred) foot strike technique. No significant differences in the total (sum of the ankle, knee and hip) positive or negative lower limb work or average power (mechanical work rate) were observed between habitual foot strike techniques, indicating that neither offers a clear-cut mechanical advantage over the other. However, the distribution of negative work and average power between the lower limb joints varied with foot strike technique which may have implications for injury. Habitual rearfoot strike runners experienced greater: peak negative power, negative work and negative average power at the knee joint compared with habitual forefoot runners. Conversely, habitual forefoot strike runners experienced greater: peak negative power, negative work and negative average power at the ankle joint compared with habitual rearfoot strikers. No significant differences were observed at the hip.
Interestingly, contrary to the view of many running coaches, switching from a habitual forefoot to an imposed rearfoot strike may prove more beneficial than switching from a habitual rearfoot to an imposed forefoot strike. When adopting an imposed forefoot strike the total positive and negative average power were greater due to increased positive work at both the ankle and hip joint and increased negative work at the ankle. Additionally, knee abduction moments were not reduced (remaining higher than habitual forefoot strike runners) and ankle internal rotation moments were increased. Switching to an imposed rearfoot strike resulted in lower total positive average power due to reduced positive work at both the ankle and hip, while knee abduction moments remained low (relative to habitual rearfoot strike runners). This switch from a habitual forefoot to an imposed rearfoot strike may prove a useful rehabilitation strategy to reduce ankle work without an overall mechanical penalty.

The primary aim of Study Two (Chapter Four) was to explore the theoretical hypothesis that elastic energy stored via compression of the longitudinal arch of the foot contributes to lowering the metabolic cost of running. To address this question, compression of the arch was restricted both maximally and by ~60% using participant-specific custom insoles and the subsequent effect on metabolic cost was analysed. Additional aims were to investigate the impact of foot strike technique on arch mechanics and the subsequent effect on running energetics (habitual rearfoot vs habitual forefoot strike), and to explore the role of the arch in walking and incline running. When compression of the arch was restricted near maximally during level running, the gross and net metabolic cost increased by (a statistically significant) 6.0% and 7.4%, respectively. In addition, the estimated arch elastic energy contribution to total mechanical limb work decreased by a similar 8.7%. Despite the large difference in arch compression, restricting compression by ~60% resulted in a similar increase in metabolic cost compared with maximal arch restriction (~80%). This observation is explained by the exponential nature of the arch compression-elastic energy relationship, whereby the majority of elastic energy is stored in the end-range of arch compression. These findings provide the most direct evidence supporting the long standing theory proposed by Ker et al., that the arch spring plays an important role in running energetics. This has important implications for arch restricting insole and footwear design and prescription.

Despite greater absolute levels of arch compression in shoe only running in forefoot compared with rearfoot strike runners, the energetic consequence of restricting arch compression was the same in both foot strike groups suggesting that both techniques gain a similar energetic benefit from the arch spring. Interestingly, the energy saving role of the arch was limited to
level running, as restricting arch compression during walking and incline running resulted in no metabolic penalty. These findings may also provide insight into the development of the foot’s arched form in human bipedal evolution.

Expanding on the findings of Study Two, Study Three (Chapter Five) investigated changes to lower limb kinetics that occurred when the elastic energy spring function of the arch was compromised during running. Using a similar comprehensive kinetic analysis as Study One, the knee was identified as the primary joint providing additional positive mechanical work, compensating for the reduced elastic energy return from the arch. Restricting arch compression also resulted in a reduction in ankle; plantar flexion moments, and positive and negative mechanical work. Despite rearfoot and forefoot strike runners displaying different lower limb mechanics in Study One, both groups compensated for the reduction in elastic arch energy in a similar manner. The redistribution of mechanical work from the ankle to the knee joint may explain the increased metabolic cost associated with restricted arch compression (Study Two), possibly attributed to a reduction in Achilles tendon elastic energy storage/return and altered muscular efficiency. These findings not only provide useful information regarding the structure-function relationship of the lower limb, but also have practical implications for individuals with impaired arch function and individuals utilising arch support insoles during running.

6.2 Synthesis of Findings

The studies conducted as part of this thesis illustrate the important role both foot strike technique and the arched anatomy of the foot have in running. The research confirms the important role compression and recoil of the arch of the foot plays in running energetics (Study Two) and also in altering the demand on the lower limb joints (and likely also musculature) (Study Three). Foot strike technique does not appear to offer runners a performance advantage, either in terms of lower limb mechanical work performed (Study One) or elastic energy storage in the arch (Study Two). Nevertheless, foot strike technique may alter a runner’s risk of lower limb injury given the variation in joint kinetic profiles associated with rear- and forefoot strike techniques. Switching to an imposed (non-preferred) foot strike technique may have implications for both performance and potential injury risk (Study One).
6.2.1 IMPACT OF FOOT STRUCTURE AND STRIKE PATTERN ON PERFORMANCE

Study Two contains one of the first studies to link restricted elastic energy storage in the arch of the foot with increased metabolic cost of locomotion. Ker and colleagues (21) established the elastic energy storing potential of the longitudinal arch ligaments using cadaver specimens. Based on loads experienced during running, Ker et al. (21) estimated that the arch of the foot is capable of storing a significant amount of elastic energy and is therefore likely to substantially reduce the metabolic cost of running. Study Two investigated this theory during locomotion by restricting arch compression via the use of two pairs of participant-specific rigid insoles that prevented 61% and 79% of arch compression and an estimated 84% and 97% (10 and 11.5 joules) reduction in arch elastic energy storage and return during running. When wearing these insoles, the gross metabolic cost of moderate paced running increased by 4.5% and 6.0% respectively, both representing a statistically significant difference from the shoe only condition. Based on these results, we conclude that the arch of the foot plays an important role in the energetics of running and that restricting compression by even ~60% can have a marked negative effect on running performance. It remains possible, that at faster speeds elastic energy storage and return in the arch may increase and hence the contribution of the arch spring to running energetics may be even greater. Interestingly, the energy saving was isolated to level running only as arch compression restriction during walking and incline running did not affect metabolic cost. This finding has important implications for orthotic and footwear design and prescription, and may have implications for the interpretation of human bipedal evolution.

It has been proposed that due to initial ground contact on the ball of the foot, a forefoot strike technique increases elastic energy storage in the arch compared with a rearfoot strike and may offer runners an advantage (24, 32). In support, Perl and colleagues found that when running barefoot with a forefoot strike, the arch undergoes greater compression and experiences greater strain compared with a rearfoot strike technique (32). Interestingly, shod running arch compression data from Study Two supports this previous research (forefoot strike 12.0 mm vs rearfoot strike 9.2 mm; \( p = 0.022; \) Appendix C) and suggests that the disparity in arch compression between a rear- and forefoot strike is likely related to the technique itself, rather than attributed to switching from a habitual forefoot to an imposed rearfoot strike technique (the comparison conducted by Perl et al. (32)). Based on the arch compression-displacement methods described in Study Two, it was estimated that 12.3 joules of elastic energy was stored in the arch of a forefoot strike runner compared with 11.2 joules in a rearfoot runner (Appendix C). Nevertheless, despite the difference in arch compression between rear- and
forefoot strikes, Perl et al. \(^{(32)}\) failed to find a metabolic difference between techniques when running shod (they did not measure metabolic cost during barefoot running). Gruber et al. \(^{(17)}\) similarly found no change in metabolic cost when habitual forefoot strike runners performed an imposed rearfoot strike technique (the comparison conducted by Perl et al. \(^{(32)}\)) or between habitual forefoot and habitual rearfoot strike runners \(^{(17)}\). Likewise, Study Two failed to find a metabolic difference between habitual rearfoot and habitual forefoot strike runners (forefoot 29.4 ml·kg\(^{-1}\)·min\(^{-1}\) vs rearfoot 30.3 ml·kg\(^{-1}\)·min\(^{-1}\); \(p = 0.584\)).

The question then remains, what mechanism explains the lack of difference in the energy cost of habitual rear- and forefoot strike runners, despite recorded differences in arch compression and possible elastic energy storage? It is possible that the absence of a metabolic advantage may be attributable to lower arch ligament stiffness in forefoot strike compared with rearfoot strike runners, and thus they may store lower elastic energy for a given arch compression. However, data from Study One suggests that the benefit of increased elastic energy storage in forefoot strike running may be counteracted by the production of additional mechanical work. Total stride average power was \(~9\%\) greater in habitual forefoot vs habitual rearfoot runners (although this was not statistically significant, \(p = 0.102\); Study One). Furthermore, when using their habitual forefoot strike, runners produced 10.5\% more total positive average power compared with the same participants using an imposed rearfoot strike technique (Study One). Despite the arch potentially storing greater elastic energy when using a habitual forefoot compared with an imposed rearfoot strike \(^{(32)}\), the habitual forefoot strike does not provide a performance advantage (Study One, \(^{(17, 32)}\)). Indeed this research suggests that regardless of the lack of a performance advantage (Study One, \(^{(17, 32)}\)), adopting an imposed rearfoot strike may prove beneficial for habitual forefoot strike runners. The reduced lower limb mechanical work (particularly at the ankle and hip) may delay muscular fatigue and reduce the risk of overuse injury. In addition, the decreased arch compression may lower the risk of injury to associated structures (e.g. the plantar fascia). However, future research should investigate if the immediate advantages gained by switching to an imposed rearfoot strike dissipate with prolonged use of the imposed technique (i.e. habituation).

Interestingly, when the spring function of the arch was restricted with rigid insoles, the metabolic cost of level running increased by the same magnitude in both foot strike groups (Appendix C), suggesting that the role of the arch did not differ. However, it may also be possible that while this method of restricting arch compression was effective at identifying the general importance of compression and recoil of the arch in the energetics of running, it was
not sensitive enough to differentiate the role of foot strike. Recent work by Kelly, et al. \(^{(20)}\) may provide an alternative (or concurrent) explanation for the greater magnitude of arch compression observed in habitual forefoot compared with rearfoot strike runners without a decrease in metabolic cost. Kelly et al. \(^{(20)}\) found that intrinsic foot muscle activation increased in conjunction with medial longitudinal arch deformation. Greater elastic energy stored in the arch of habitual forefoot strike runners may therefore be offset by the metabolic cost of increased intrinsic foot muscle activation, although the metabolic cost associated with these small muscles is likely low.

Future research should investigate the effect of switching from a habitual rearfoot strike technique to an imposed forefoot strike on arch mechanics. Unsupported claims of improved performance and reduced injuries have resulted in some coaches recommending this change in foot strike technique to their athletes \(^{(1, 37, 49)}\). However, Study One found this change in technique resulted in an increase in both total positive and negative lower limb average power by 17% and 9% respectively. In addition, Gruber et al. \(^{(17)}\) found a concurrent increase in metabolic cost with this change in foot strike. It could be hypothesised that part of the increase in positive average power when performing an imposed forefoot strike is due to the arch of the foot not recycling as much elastic energy as a habitual forefoot strike runner. Therefore, the elastic energy which normally offsets the higher mechanical and metabolic cost of performing a forefoot strike is no longer present, resulting in a significant increase in both (Study One and Gruber et al. \(^{(17)}\)).

The results of Study Two may have implications for barefoot running. Pilot work revealed that the arch underwent significantly greater \((p = 0.002)\) compression during barefoot compared with shod running at 2.7 ms\(^{-1}\) (12.9 mm vs 10.6 mm). This was evident in both rearfoot and forefoot strike runners. Based on Ker et al.’s \(^{(21)}\) work, the greater compression likely results in increased elastic energy storage and theoretically provides barefoot runners a metabolic advantage. However, while studies have identified a reduced energy cost of barefoot versus shod (standard footwear) running \(^{(10, 16, 19, 40)}\), it is not of the magnitude one would expect given the absence of shoe mass (see Figure 1 Franz et al. \(^{(15)}\)) and the potential increase in arch elastic energy return. Furthermore, lightweight shoes (< 150 grams) have been shown to result in the same metabolic cost as barefoot running \(^{(10, 15, 16, 34, 40, 43)}\). It has been proposed that this may be due to the shoe cushioning returning elastic energy to the runner and the ‘cost-of-cushioning’ theory, whereby cushioning is accomplished by the actions of the leg muscles as opposed to the shoe therefore requiring metabolic energy \(^{(4, 15, 30)}\). In addition, barefoot
running training has been shown to strengthen foot muscles \(^{(26)}\) and greater arch compression is associated with increase intrinsic foot muscle activation \(^{(20)}\). Hence, despite the arch of the foot potentially storing greater elastic energy when running barefoot compared with shod, the metabolic advantage gained may be counteracted by the cost of cushioning and the cost of increased intrinsic foot muscle activation, although these hypotheses require further investigation.

Interestingly, in Study Two when arch compression was restricted by 82% during walking, there was on average no metabolic penalty. This suggests that even though an estimated 2.6 joules of elastic energy was stored in the arch, it does not significantly contribute to overall metabolic cost. Alternatively, it is possible that the cost of walking was unaffected due to the insoles increasing midfoot rigidity and hence enhancing the foot’s function as a lever during push off \(^{(2, 11, 13)}\). This may have counteracted the disadvantage of reduced elastic energy storage and return. Nevertheless, these results suggest that orthotic intervention does not systematically affect the metabolic cost of walking and therefore provides a viable treatment option without metabolic detriment for individuals with musculoskeletal conditions requiring correction by arch support (e.g. plantar fasciitis).

### 6.2.2 Effect of Foot Strike Technique on Lower Limb Joint Mechanics

While neither a habitual rearfoot strike nor a habitual forefoot strike technique appear to offer runners a performance advantage (Studies One, Two and Gruber et al. \(^{(17)}\)), Study One revealed significant lower limb joint kinetic differences between the techniques which may have injury risk implications. It has been promoted in the lay media that a rearfoot strike technique places a runner at greater risk of injury than a forefoot strike \(^{(29)}\). However, to date only a small sample size study by Daoud and colleagues \(^{(9)}\) has provided support for this claim. A larger study conducted by Walther \(^{(45)}\) found no significant difference in the injury incidence between foot strike groups but did find variation in the injury location. Combining data from Daoud et al. \(^{(9)}\) and Walther \(^{(45)}\) it was reported that rearfoot strikers experience a higher incidence of injuries to the: tibialis posterior, knee, hip and plantar fasciitis, and forefoot strikers suffer more Achilles tendon injuries. While this thesis did not directly investigate injury rates, Study One conducted a comprehensive analysis of lower limb joint kinetics, providing insight into the forces placed on internal structures and subsequent injury risk predisposition. The key difference identified between habitual rearfoot and habitual forefoot strike techniques was an altered distribution of negative work, which may partially explain the different injury locations.
identified by Daoud et al. \(^9\) and Walther \(^{45}\). Rearfoot strike runners utilised the knee to a greater extent to absorb negative work (~40%) while forefoot strike runners absorbed ~60% at the ankle (Studies One and Three). It was also identified that habitual rearfoot strikers have significantly greater knee abduction moments than habitual forefoot runners (Study One and Kulmala et al. \(^{23}\)). Negative work often involves eccentric muscle fibre contractions which have been documented to cause greater muscle damage and fatigue than concentric contractions \(^8\) and high knee abduction moments have been linked to common knee injuries such as patellofemoral pain syndrome \(^{41}\) (the most prevalent lower limb injury \(^{42}\)) and degenerative conditions such as osteoarthritis \(^{46}\). Therefore, the differences identified in joint kinetics between foot strike techniques may place rearfoot strike runners at greater risk of knee injury and forefoot strike runners at increased risk of ankle injury. This conclusion is supported by injury location data reported by Daoud et al. \(^9\) and Walther \(^{45}\).

A combination of factors (predominantly unsupported claims by the lay media) have led some coaches to instruct their habitually rearfoot athletes to change to an imposed forefoot strike technique with the belief that it will improve running performance and reduce injury \(^1, 37, 49\). As discussed previously, switching to an imposed forefoot strike increases both the metabolic cost of running \(^{17}\) and total positive average power production (17%; Study One), therefore the switch does not appear to elicit the perceived performance advantage. Notwithstanding, it could be hypothesised that the risk of injury to the knee may be reduced by adopting an imposed forefoot strike, based on the differences observed between habitual techniques (Study One, \(^9, 45\)). However, Study One revealed that when switching to an imposed forefoot strike, not all joint kinetics of the habitual foot strike are initially adopted. For example, while high knee positive and negative mechanical work was reduced when a habitual rearfoot strike runner switched to an imposed forefoot strike, high knee abduction moments were not (Study One). In support, Kleindienst et al. \(^{22}\) also found no difference in peak knee abduction moments when runners performed a habitual rearfoot strike versus an imposed forefoot strike technique. Switching to an imposed forefoot strike may therefore not reduce knee injury risk by as much as hypothesised in the lay arena. It is also possible that the switch may increase the risk of ankle injury as negative ankle work was increased to be in line with a habitual forefoot strike technique (Study One). Load placed on the metatarsal bones when switching from a rearfoot to an imposed forefoot strike technique is most likely also increased, placing the runner at greater risk of metatarsal stress fracture, although this is yet to be measured directly.
Although not often considered by coaches and athletes, switching from a habitual forefoot strike to an imposed rearfoot strike may prove a more useful strategy than the more commonly recommended switch from a habitual rearfoot to an imposed forefoot strike. Habitual forefoot strike runners were able to almost exactly replicate the joint mechanics of habitual rearfoot runners when adopting the imposed rearfoot strike technique, with only a few minor and favourable exceptions. Positive and negative mechanical work were decreased at the ankle and increased at the knee, likely altering the potential for injury. Nevertheless, knee abduction moments remained low after the technique switch; in that they did not increase to match a habitual rearfoot strike. Consequently, while the risk of injury to the knee is possibly higher in the imposed rearfoot strike condition compared with a habitual forefoot strike, it is likely not as high as that of a habitual rearfoot strike runner. In addition, the risk of injury to the ankle is potentially decreased. This change in technique may provide a useful rehabilitation strategy for recovery from foot and ankle injuries such as Achilles tendonitis and plantar fasciitis given that loads about the ankle (Study One) and arch compression \(^{32}\) (plantar fascia strain) are reduced, without incurring an overall mechanical (Study One) or metabolic \(^{17}\) penalty. However, as previously mentioned, future research should investigate if training leads to complete adoption of the imposed foot strike technique kinematics and kinetics.

Interestingly, despite the significant differences in lower limb kinetics identified between habitual rear- and forefoot strike runners (Study One), when arch compression was restricted, both groups compensated for the loss of arch elastic energy in a similar manner. Positive and negative mechanical work at the knee was increase and positive and negative work at the ankle decreased (Study Three). Nonetheless, rearfoot strike runners continued to use the knee to absorb majority of negative work and forefoot strike runners the ankle (Appendix C). At the knee, both positive and negative mechanical work was increased by a greater percentage in forefoot strike compared with rearfoot strike runners, although the absolute magnitude of both remained larger in the rearfoot strike group. This has the potential to increase the risk of knee injury in both foot strike groups; rearfoot strike runners due to the large magnitudes and forefoot strike runners due to the unfamiliarity of performing work using the knee musculature. Common knee injuries resulting from high repetitive knee loads are quadriceps/patellar tendonitis \(^{25, 36}\) and Osgood-Schlatters disease \(^{7}\), hence restricting arch compression (e.g. arch support orthotics) may well result in an increased risk of these injuries for both forefoot and rearfoot strike runners.
Conversely, restricting arch compression may be beneficial for preventing and/or treating ankle injuries. A reduction in Achilles tendon peak force was estimated when arch compression was restricted (Study Three). Corroborating this finding, Sinclair et al. (38) also found that Achilles tendon force was reduced when running with orthotic insoles. In Study Three, there was also a reduction in peak plantar-flexion moments and ankle mechanical work (both positive and negative) during the insole condition. These ankle variables were decreased in both foot strike groups but more so in fore- than rearfoot strike runners. Given that forefoot runners typically experience high loads at the ankle (Studies One and Three) and are prone to Achilles tendon injuries (9,45), insole intervention may be beneficial. However, while the insoles reduced positive and negative ankle work by a 10.5% and 10.3% respectively in forefoot strike runners, a more effective way to reduce ankle work would be to switch to an imposed rearfoot strike technique. Changing from a habitual forefoot to an imposed rearfoot strike resulted in a 15% and 32% reduction in positive and negative ankle work respectively (Study One). Additionally, the rigid insoles lead to a ~6% increase in metabolic cost (Study Two), while switching to an imposed rearfoot strike did not incur a metabolic penalty (17) and reduced total positive lower limb average power by 10.5% (Study One). However, switching to an imposed rearfoot strike technique did increase positive and negative knee work by 38% and 40% respectively, while the use of the rigid insoles only brought about a 25% and 28% increase positive and negative knee joint work respectively. Therefore the magnitude of decrease in ankle work should be weighed against the increase in knee work and the overall effect on metabolic cost.

Neither habitual rearfoot strike nor habitual forefoot strike runners were able to completely adopt the lower limb kinetics of an imposed foot strike technique (Study One). It is therefore possible, that when performing the imposed foot strike they may respond differently to restriction of the spring function of the arch. This has important implications for runners who wish to alter their foot strike technique, perhaps due to injury concerns, but have impaired arch function (e.g. rigid orthotics). Further research is required in this area, combined with further investigation of varied arch function and form (e.g. pes planus, pes cavus and excessive pronation) on foot strike technique and the role of the arch in running. Williams and colleagues (47) identified significant differences in lower limb mechanics between high and low arched habitual rearfoot strike runners, which could, in part, be attributed to varied elastic contribution from the arch of the foot. Further research is warranted in this area as this information has important implications for orthotic and footwear prescription for individuals with varied arch shape.
6.2.3 Arch Support During Locomotion

Studies Two and Three confirm the important role the arch of the foot plays during running. Restricting compression of the arch resulted in a significant increase in metabolic cost, and mechanical work absorption and production at the knee. While the insoles utilised in this thesis are not reflective of those typically recommended by clinicians, the findings do have important implications for orthotic insole design and prescription. Clinicians should be cautious recommending orthotic insoles with rigid arch support to runners with already high knee loads. Based on the results of Study Three, rigid arch support intervention may place the runner at an increased risk of overloading the knee structures. However, as previously noted, orthotic intervention may be beneficial for individuals with ankle overload injuries such as Achilles tendonitis and plantar fasciitis.

From a performance perspective, during competition athletes should consider the potential increase in metabolic cost associated with restricting arch compression, against the benefits of improved foot biomechanics that orthotic insoles may offer. Of note, the increase in metabolic cost when wearing the insoles in Study Two was potentially underestimated because the effect of insole weight was controlled for by adding the equivalent insole mass to the foot in the shoe only condition. The insoles each weighed an average of 60 grams which would be expected to increase the metabolic cost by an additional 0.6% \(^{(16)}\). However, it should be highlighted again that the insoles used in Studies Two and Three are not typical of those prescribed by clinicians. They were extremely rigid, only returning ~0.3 joules of energy, while typical orthotics are often constructed of more flexible materials and return a greater amount of elastic energy.

The performance implications of restricting arch compression not only apply to orthotic insoles but also to supportive footwear. Footwear does not contain the same level of arch restriction as the insoles employed in this thesis. However, due to the non-linear stress-strain properties of tendons and ligaments, the majority of elastic energy is stored in the final stages of arch compression \(^{(21)}\). As demonstrated in Study Two, restricting arch compression by ~60% (HAI) resulted in a significant increase in metabolic cost. Perl et al. \(^{(32)}\) also came to the same conclusion after finding a 3% increase in metabolic cost when wearing traditional arch supporting shoes compared with minimal footwear. Therefore, while supportive footwear may benefit certain individuals (for example by controlling excessive rearfoot pronation), even mildly restricting arch compression may negatively impact runners by increasing metabolic cost (Study Two) and by placing a greater demand on the knee musculature (Study Three).
Restricting arch compression had minimal effect on the metabolic cost of incline running (Study Two). It was hypothesised that the metabolic penalty would be lower in incline compared with level running when the arch spring was restricted, due to the smaller percentage of total positive mechanical work that is provided by elastic energy in uphill running\(^{(39)}\). However, the non-significant difference in metabolic cost was surprising.

6.3 LIMITATIONS AND DELIMITATIONS

This thesis has a number of limitations which may limit the generality of the findings. Certain aspects were intentionally delimited to define a workable research problem while others were difficult to specifically control.

6.3.1 TREADMILL AND SPEED

All trials were performed on a motorised treadmill, limiting the results applicability to over ground running. However, the use of a treadmill allowed for speed to be controlled, oxygen consumption to be feasibly collected and steady state running to be achieved. Additionally, the findings are only directly applicable to the speeds assessed: 4.5 ms\(^{-1}\) in Study One, 1.1 ms\(^{-1}\) (walking) and 2.7 ms\(^{-1}\) (level and incline running) in Study Two, and 2.7 ms\(^{-1}\) in Study Three. It is acknowledged that the running speed imposed during Studies Two and Three are slower than many running related studies which may have influenced the results. However, this speed was carefully selected after pilot testing for a number of reasons: vertical loads were low enough to prevent excessive insole compression, discomfort caused by the insoles was minimised and it allowed the same speed to be used during level and incline conditions without risking transition to anaerobic metabolic pathways.

6.3.2 PARTICIPANT CHARACTERISTICS

All participants who took part in this research were male Caucasians, aged 18 - 35 years, injury free at the time of testing and did not wear orthotic insoles. Only participants with ‘normal’ feet as defined by the Foot Posture Index\(^{(35)}\) were included in Studies Two and Three. This may limit the applicability of the findings to females, other populations and individuals with varied arch morphology such as pes planus or pes cavus. Future research should investigate these populations.
6.3.3 Foot Strike Classification

Foot strike group was categorised using high speed video collected at 300 frames per second. Alternative approaches have been employed in the literature to classify foot strike techniques including; i) the Strike Index (Cavanagh) which compares the centre of pressure at initial contact to the length of the foot segment \(^6\), ii) the presence or absence of a vertical ground reaction force impact peak and, iii) the ankle angle at initial ground contact. Gruber et al. \(^{18}\) compared these three techniques and found that they each produced varying foot strike classifications for mid- and forefoot strike runners. Each method has limitations and it remains unclear which, if any, offers the most accurate classification. It was therefore concluded that given the absence of a midfoot strike group in this thesis, the high speed camera classification technique utilised was suitable. The same experienced assessor classified the foot strike technique of each participant after viewing no fewer than 20 strides. However, it remains possible that forefoot strike runners used a midfoot strike for a small percentage of their running strides.

6.3.4 Foot Strike Familiarity

While participants were given as much time as they needed in the laboratory to familiarise themselves with the imposed foot strike and all runners reported previous basic experience with the technique, the kinetic differences observed in Study One represent the acute effects of switching technique. It is possible that with prolonged training using the imposed technique, runners will reflect lower limb kinetics consistent with those found in habitual foot strike techniques. This would prove beneficial for a habitual rearfoot to an imposed forefoot strike technique switch, given it may reduce: total positive and negative average power, knee abduction moments, ankle internal rotation moments, ankle positive and negative mechanical work, and hip positive work. However, in the case of a habitual forefoot strike runner adopting an imposed rearfoot strike technique it may produce some unfavourable outcomes given the lower total positive average power and knee abduction moments reported when adopting the imposed rearfoot foot strike pattern.
6.3.5 Arch Compression Measurement

Arch compression was recorded via a skin mounted marker on the navicular tuberosity. While this is a common method for externally measuring arch mechanics, (e.g. (3, 12, 20, 32)) it may not exactly represent internal skeletal motion and may have led to an under-estimation of midfoot motion (27).

The novel insole approach employed in Studies Two and Three, effectively reduced arch compression during running and resulted in a significant increase in metabolic cost. The metabolic penalty when running in the insoles was the same in rear- and forefoot strike runners despite greater maximum arch compression during shoe only running in forefoot compared with rearfoot strike runners. It may also be possible that while this method of restricting arch compression was effective at identifying the general importance of compression and recoil of the arch in the energetics of running, it was not sensitive enough to differentiate the role of foot strike.

6.3.6 Orthotic Discomfort

The increase in metabolic cost observed during running in Study Two with the use of the rigid insoles was attributed to reduced elastic energy storage in the arch of the foot. However, arch elastic energy was not measured directly but estimated from arch compression, joint compressive force and a model prediction. When expressed as a percentage of centre of mass mechanical work, estimated restricted elastic energy closely matched the measured increase in metabolic cost. It could be suggested, instead, that the increase in metabolic cost was attributable to the uncomfortable nature of the rigid insoles, given the findings by Nigg (31) that metabolic cost increased in conjunction with perceived footwear discomfort. However, participants reported the discomfort the insole caused during level and incline running as minimal (Appendix C). Given these low pain ratings it is reasonable to suggest that the level running results are not attributed to discomfort. The lack of difference in metabolic cost during both the walking and incline running conditions also strengthens the interpretation that the energetic changes observed in level running are due primarily to alterations in the arch mechanics. If the insoles elicited other general modifications such as co-contraction, instability, or cushioning effects (15), an elevation in metabolic cost would also be expected (and perhaps magnified) during incline running and perhaps also walking. Instead, the metabolic cost in these conditions was unaffected.
6.3.7 **Clinical Orthotic Intervention**

The insoles utilised in this thesis are not reflective of those typically prescribed by clinicians. They were constructed using extremely rigid materials (shore value ~350-400) and did not attempt to alter foot motion other than preventing the medial longitudinal arch from compressing. While the findings provide insight into the consequences of restricting arch compression and may have implications for orthotic design and prescription, they cannot be applied directly to general orthotic intervention.

6.3.8 **Multi-Segment Foot Model**

The use of multi-segment foot model such as the Oxford foot model \(^{(5)}\) may have allowed for further quantification of the foot’s function during gait, however the feasibility of using a multi-segment multi-marker approach is limited when attempting to assess shoe foot motion. We believe that the multi-segment foot model method used in Studies Two and Three to quantify arch compression and foot kinematics was suitable to answer the research questions posed. Assuming that an appropriate methodological protocol can be developed that does not fundamentally affect the foot/shoe interaction, or the function of the shoe itself, an interesting area of future research would be to investigate how foot function varies between foot strike techniques using a model that accounts for tri-axial rearfoot (hindfoot), forefoot and hallux motion.

6.3.9 **Continuous Data Analysis**

In Studies One and Three, discrete data points from kinetic variables (typically the maximum or minimum values) were selected for analysis. It is possible that analysing discrete selected events in the gait cycle obscures more subtle time-series affects. Future research should investigate the use of continuous data analysis techniques. However, despite not employing these statistical methods, both studies identified numerous significant differences between conditions.
6.4 PRACTICAL APPLICATION

Many of the findings arising from this thesis will be of interest to running coaches and athletes given the topical content that specifically examines the foot’s role in running performance and injury risk. Therefore, the applied knowledge that can be disseminated from the findings of this research is as follows;

- The arch of the foot acts as an important energy saving spring during level running.
- Habitual forefoot strike runners do not appear to have a clear performance advantage over habitual rearfoot strikers. Subsequently, changing foot strike technique does not improve running performance; runners should stick to what comes naturally.
- During competition, athletes should weigh the benefits of improved foot biomechanics that orthotic insoles may offer against an increase in metabolic energy cost.
- Adopting arch supporting insoles may be a useful rehabilitation strategy for runners with plantar fasciitis or Achilles tendinosis. However, load is shifted to the knee and its surrounding musculature which may subsequently increase the knees susceptibility to injury.
- Finally, changing foot strike technique may be a useful strategy for shifting lower limb load and altering injury risk. However, rearfoot runners may be at greater risk of developing knee injuries with forefoot strike runners likely at greater risk of developing an ankle injury. Transition to a new foot strike technique should be undertaken gradually and under supervision.


6.5 REFERENCES


APPENDICES

APPENDIX A — ETHICS

APPENDIX B — PARTICIPANT FORMS

APPENDIX C — ADDITIONAL DATA

APPENDIX D — PUBLICATIONS & MEDIA  NOTE: Removed due to copyright restrictions.
APPENDIX A – ETHICS

APPENDIX A 1 – UNIVERSITY ETHICS APPROVAL

The University of Western Australia
Achieving International Excellence

Research Ethics and Biobehavioural Health
Research Services
Our Ref: RA/4/1/4541
07 February 2011

Assistant Professor Jacqueline Alderson
Sport Science, Exercise & Health (School of)
MBDP: M408

Dear Professor Alderson

HUMAN RESEARCH ETHICS APPROVAL - THE UNIVERSITY OF WESTERN AUSTRALIA
Role of musculoskeletal passive-elastic mechanisms in the mechanics and energetics of running

Student(s): Sarah Michelle Stoner

Ethics approval for the above project has been granted from 04 February 2011 to 01 February 2012 in accordance with the requirements of the National Statement on Ethical Conduct in Human Research (National Statement) and the policies and procedures of The University of Western Australia.

You are reminded of the following requirements:

1. The application and all supporting documentation form the basis of the ethics approval and you must not depart from the research protocol that has been approved.
2. The Human Research Ethics Office must be approached for approval in advance for any requested amendments to the approved research protocol.
3. The Chief Investigator is required to report immediately to the Human Research Ethics Office any adverse or unexpected event or any other event that may impact on the ethics approval for the project.
4. The Chief Investigator must inform the Human Research Ethics Office as soon as practicable if a research project is discontinued before the expected date of completion, providing reasons.

Any conditions of ethics approval that have been imposed are listed below:

Special Conditions
None specified

The University of Western Australia is bound by the National Statement to monitor the progress of all approved projects until completion to ensure continued compliance with ethical standards and requirements.

Please note that the maximum period of ethics approval for this project is five (5) years from the date of this notification. However, ethics approval is conditional upon satisfactory progress reports being received by the designated renewal date for continuation of ethics approval.

The Human Research Ethics Office will forward a request for a Progress Report approximately 60 days before the due date. A further reminder will be forwarded approximately 30 days before the due date.

If your progress report is not received by the due date for renewal of ethics approval, your ethics approval will expire, requiring that all research activities involving human participants cease immediately.

If you have any queries please do not hesitate to contact the Human Research Ethics Office (HREO) at hreo.research@uwa.edu.au or (08) 6488 3703.

Please ensure that you quote the file reference – RA/4/1/4541 – and the associated project title in all future correspondence.

Yours sincerely

[Signature]
Peter Johnstone
Manager
Human Research Ethics Committee
APPENDIX A 2 – ETHICS AMENDMENT APPROVAL 2011

Our Ref: RA/4/1/4541

Assistant Professor Joans Rubenson
Sport Science, Exercise & Health (School of)
MBDP: M408

Dear Professor Rubenson

HUMAN RESEARCH ETHICS OFFICE – AMENDMENT REQUEST APPROVED
Role of musculo-skeletal passive-elastic mechanisms in the mechanics and energetics of running

Student(s): Sarah Michelle Stearne

I confirm receipt of your correspondence requesting an amendment to the protocol for the above project.
Approval has been granted for the amendment as outlined in your correspondence and attachments (if any) subject to any conditions listed below.
Any conditions of ethics approval that have been imposed are listed hereunder:
1. Foot strike technique
If you have any queries, please do not hesitate to contact Kate Kirk on (08) 6488 3703.
Please ensure that you quote the file reference RA/4/1/4541 and the associated project title in all future correspondence.

Yours sincerely

[Signature]

Peter Johnstone
Manager
Human Research Ethics Committee
Our Ref: RA/4/1/4541

Assistant Professor Joans Rubenson
Sport Science, Exercise & Health (School of)
MBDP: M408

Dear Professor Rubenson

HUMAN RESEARCH ETHICS OFFICE – ETHICS APPROVAL RENEWED
Role of musculo-skeletal passive-elastic mechanisms in the mechanics and energetics of running

Student(s): Sarah Michelle Steane

Thank you for submitting your Progress Report for the above project. The report is satisfactory and ethics approval for the project has been renewed.

You will receive a request for your next progress report approximately one month before the next renewal date of 01 March 2013. If you have any queries, please do not hesitate to contact the Human Research Ethics Office (HREO) on (08) 6488 5703.

Please ensure that you quote the file reference – RA/4/1/4541 – and the associated project title in all future correspondence.

Yours sincerely

[Signature]

Kate Kirk
Secretary
Human Ethics Research Committee
Our Ref: RA/4/1/4541

18 January 2013

Assistant Professor Jonas Rubenson
School of Sport Science, Exercise & Health
MBDP: M408

Dear Professor Rubenson

HUMAN RESEARCH ETHICS OFFICE – AMENDMENT REQUEST APPROVED

Role of musculo-skeletal passive-elastic mechanisms in the mechanics and energetics of running

Student(s): Sarah Michelle Stearne

I confirm receipt of your correspondence requesting an amendment to the protocol for the above project. Approval has been granted for the amendment as outlined in your correspondence and attachments (if any) subject to any conditions listed below.

Any conditions of ethics approval that have been imposed are listed hereunder:

1. Approval is granted to:
2. Lower the recruitment age to 17 years.
3. Expand recruitment to include the Perth athletics community.
4. Take voluntary finger-tip capillary blood samples from participants.

If you have any queries, please contact the HREO at hreo-research@uwa.edu.au.

Please ensure that you quote the file reference RA/4/1/4541 and the associated project title in all future correspondence.

Yours sincerely

[Signature]

Peter Johnstone
Manager, Human Research Ethics
APPENDIX B – PARTICIPANT FORMS

APPENDIX B 1 – PARTICIPANT INFORMATION SHEET

Role of Musculo-Skeletal Passive-Elastic Mechanisms in the Mechanics and Energetics of Running

Information Sheet

Chief Investigators: Dr. Jacqueline Alderson PhD
Assistant Professor, School of Sport Science, Exercise & Health, The University of Western Australia (UWA)
Phone: (08) 6488 5327

Dr. Jonas Rubenson, PhD
Assistant Professor, School of Sport Science, Exercise & Health, UWA
Phone: (08) 6488 5333

Mr Cyril Jon Donnelly MSc
Lecturer, School of Sport Science, Exercise & Health, UWA
Phone: (08) 6488 3919

Co Investigators: Sarah Steane BSc(Hons)
PhD Candidate, School of Sport Science, Exercise & Health, UWA
Phone: (08) 6490 1305

Benjamin Green BSc
School of Sport Science, Exercise & Health, UWA
Phone: 0422 890 021

Purpose of the Study
It is known that the Achilles tendon and ligaments in the arch of the foot recycle significant amounts of passive energy and therefore play a vital role in running. However, the effect of these passive-elastic mechanisms on running mechanics and energetics remains unknown. This research aims to investigate the effect the Achilles tendon and the ligaments in the arch of the foot have on joint, muscle and metabolic work during running. We also aim to gain an insight into how joint, muscle and metabolic work might be influenced by foot strike technique (rear foot vs forefoot), changes in running speed and changes in ligament and tendon elasticity via the use of custom shoe orthoses and an external ankle orthosis (exoskeleton).

The portion of the foot that initially makes contact with the ground during running is one variable where runners exhibit variability. Some runners will employ a rear foot strike (where they land on their heels) and others a forefoot strike (landing on the ball of the foot). The effect of these two foot-strike postures on loading at the ankle is not well understood. This research will investigate how different foot strike postures affect the moment arms of the muscles that cross the ankle joint as well as the activation patterns of the muscles in the posterior compartment of the leg (calf muscles). This research also aims to establish how a natural vs. imposed foot strike posture in runners will affect the moment arms at the ankle and activation of the leg muscles.
What is required of you?

You will be asked to attend the School of Sport Science, Exercise and Health at The University of Western Australia on four or five occasions depending on the test group you are allocated to. Three or four visits will be to the Sports Biomechanics Laboratory and one to the Exercise Physiology Laboratory. The first session will take approximately one hour and involve some initial screening tests (e.g. to determine your foot strike technique and to test for excessive foot pronation). At the completion of this session you will be allocated to one of two test groups. Depending on allocation you will then be asked to return to the school a further three or four times, for testing sessions approximately two hours each.

During the initial screening session you will be required to run on a split belt motorised treadmill for approximately ten minutes at speeds ranging from 10-16km/hr whilst foot-strike technique is videoed. You will also be given the opportunity to familiarise yourself with breathing into a mouth piece while running and will you also be shown equipment and procedures that will be used in subsequent sessions.

The second session will take place in the Exercise Physiology laboratory where your maximal aerobic capacity will be assessed. This visit will involve running on a motorised treadmill at incrementally faster speeds, while breathing into a mouthpiece, up to the fastest speed you can sustain. The protocol used to assess your maximal aerobic capacity will follow the American College of Sports Medicine guidelines for exercise testing and is used routinely in our School. You will be required to meet a minimal VO2 fitness level of 50ml/kg/min.

The third session will involve running on a treadmill at 8, 10, 12, 14 and 16km/hr for approximately four minutes per speed. At the higher speed you will be asked to run using your natural foot strike posture (whether rear foot or fore foot) and an imposed foot strike posture. Whilst running joint movement, oxygen consumption, muscle activation and muscle-tendon length change will be measured. Joint movements will be recorded using specialised infra-red cameras. Twenty-six retro-reflective markers will be placed on your lower limbs and trunk using low allergenic double-sided tape. The markers will be tracked using cameras and converted into a three-dimensional image for computer analysis. To measure oxygen consumption you will be required to breathe into a mouth piece whilst running. Muscle activation will be recorded through small adhesive surface electrodes which will be placed on seven lower limb muscles. Muscle and tendon length change will be determined using a small ultrasound probe strapped to the calf. Ultrasound is a completely safe imaging analysis modality with no known side effects.

At the fourth testing session you will be required to run on a treadmill at 12km/hr for 4x 4minutes, each time with a different shoe insole. Insoles will range from extremely soft to completely stiff, similar to orthotics made by podiatrists and you will be provided with a shoe to wear by the researcher. The same measures described above in testing session two will be recorded.

Only participants who are identified as rear-foot strikers will be required to attend the final testing session (session five). These participants will run on a treadmill at 12km/hr for 5x 4mins whilst wearing an ankle/foot orthosis (exoskeleton). The orthosis will consist of a brace that is worn on your shin and attached to your shoes using straps. A different elasticity spring will be attached to the orthosis for each exercise bout. The same measures as previous testing sessions will be recorded. Participants will be given the opportunity to familiarise themselves with running in the exoskeleton prior to the commencement of this session.
Possible Benefits
Your participation in this research will act to advance the current state of our scientific knowledge surrounding the role of passive-elastic mechanisms during locomotion as well as how the type of foot strike posture employed when running influences loading at the ankle joint. We hope this knowledge will contribute to providing a framework for determining how a runner should strike the ground in relation to how they naturally run, as well as investigating how the Achilles tendon responds to a natural and imposed running posture. Your participation will allow us to gain a significant insight into the function of the Achilles tendon and the ligaments comprising the arch of the foot during running. Improving our basic understanding of these passive-elastic mechanisms may have significant benefits for injury prevention, rehabilitation and performance enhancement for athletic and disabled populations. Your participation will also assist in the development of a new rehabilitation and performance enhancing device, an ankle-foot exoskeleton.

Possible Risks/Discomforts
Some people are allergic to specific adhesive tapes, however the tape used to attach the markers is low-allergenic, and a very small number of people experience a reaction. Any irritation experienced however, will by relatively minor and short term in duration.

On one occasion during session three you will require you to run with a foot strike technique that is not natural to you, however the researchers believe the small duration of the testing (four mins) will not pose any adverse response to you as it will represent a small percentage of your normal volume of weekly distance running training.

During session four, you will be required to wear a range of shoes insoles and one will be made of semi-rigid plastic preventing the arch of the foot from collapsing. This may cause slight discomfort, however it is no different to the corrective insoles commonly prescribed by podiatrists.

You may also feel some discomfort and awkwardness running with the custom foot and limb orthosis, however we will provide you with time to familiarise yourself with this equipment prior to testing.

If at any point during the testing sessions you wish to discontinue due to discomfort you may do so without prejudice.

What to Bring
You will be provided with a pair of running shoes for all testing sessions, however if preferred you may wear your own footwear for the VO2 max test and pre-testing warm ups. You are advised to wear sports socks to reduce the possible discomfort of new shoes and orthotics. To all testing sessions you are required to wear running shorts/short sports shorts as retro-reflective markers will be placed on your legs and feet to track movement.

Investigator Responsibilities
All video data will not show your face. All recordings will also be de-identified during collection and stored in a password protected hard drive for later analysis. Names and personal information will be kept strictly confidential. This includes digital or computerised information. Your anonymity will be preserved in any publications of data collected.

Withdrawal
Participation in this study is voluntary and you are free to withdraw from the research at any time without reason or justification. In such cases, your records will be destroyed unless you agree that the researcher may retain and use the information obtained prior to your withdrawal. If you wish to
remove yourself from the study, please contact the Chief Investigator at the earliest possible convenience. If you withdraw from the study and you are an employee or student at the University of Western Australia this will not prejudice your status and rights as employee or student of UWA.

Approval to conduct this research has been provided by The University of Western Australia, in accordance with its ethics review and approval procedures. Any person considering participation in this research project, or agreeing to participate, may raise any questions or issues with the researchers at any time.

In addition, any person not satisfied with the response of researchers may raise ethics issues or concerns, and may make any complaints about this research project by contacting the Human Research Ethics Office at The University of Western Australia on (08) 6488 3793 or by emailing to freo-research@uwa.edu.au.

All research participants are entitled to retain a copy of any participant information form relating to this research project.
APPENDIX B 2 – STUDY ONE PARTICIPANT CONSENT FORM

Role of Musculo-Skeletal Passive-Elastic Mechanisms in the Mechanics and Energetics of Running

Consent Form

Chief Investigators:
Dr. Jonas Rubenson, PhD
Assistant Professor,
School of Sport Science,
Exercise & Health
The University of Western Australia (UWA)
Phone: (08) 6488 5533

Dr. Jacqueline Alderson PhD
Assistant Professor,
School of Sport Science,
Exercise & Health, UWA
Phone: (08) 6488 5827

Co Investigators:
Sarah Stearne BSc Hons
PhD candidate,
School of Sport Science,
Exercise & Health, UWA
Phone: (08) 6488 1383

Benjamin Green BSc
School of Sport Science,
Exercise & Health, UWA
Phone: 0422 990 021

I __________________________ have read the information provided and any questions I have asked have been answered to my satisfaction. I agree to participate in this activity, realising that I may withdraw at any time without reason and without prejudice.

I understand that all information provided is treated as strictly confidential and will not be released by the investigator unless required to by law. I have been advised as to what data is being collected, what the purpose is, and what will be done with the data upon completion of the research.

I agree that research data gathered for the study may be published provided my name or other identifying information is not used.

__________________________  __________________________
Participant                          Date

Approval to conduct this research has been provided by The University of Western Australia, in accordance with its ethics review and approval procedures. Any person considering participation in this research project, or agreeing to participate, may raise any questions or issues with the researchers at any time.
In addition, any person not satisfied with the reasons of researchers may raise ethical issues or concerns, and may make any complaints about this research project by contacting the Human Research Ethics Office at The University of Western Australia on (08) 6488 5705 or by emailing hrecresearch@uwa.edu.au.
All research participants are entitled to retain a copy of any Participant Information Form and/or Participant Consent Form relating to this research project.
APPENDIX B 3 – STUDY ONE PARTICIPANT RUNNING QUESTIONNAIRE

Running Questionnaire

Please answer the following questions based on your current and previous involvement in distance running.

Number of years you have been involved in distance running? ______________________

Average number of weeks you train per year? ______________________

Average number of runs you complete per week? ______________________

Average number of km you run per week? ______________________

Preferred competitive distance running events (e.g. 1500m, 5k, Half Marathon)

________________________________________________________________________

$\text{VO}_2\text{max} \, (\text{if known - e.g. } 65 \, \text{ml.kg.min}^{-1}) \quad ______________________

10km time (PB time and most recent time) ______________________

Have you sustained any major overuse injuries as a result of distance running in the past 5 years? (Yes/No)
If yes, please state what the injury was, how long it took to resume to the training level you were at prior to sustaining the injury and if you have suffered a repeat of the same/similar injury

________________________________________________________________________

Have you sustained any major overuse injuries as a result of distance running in the past 6 months? (Yes/No)
If yes, please state what the injury was, how long it took to resume to the training level you were at prior to sustaining the injury and if you have suffered a repeat of the same/similar injury

________________________________________________________________________

Print Name ______________________ Signature ______________________ Date (dd/mm/yy) ______________________

Sarah Stearne and Ben Green
UWA Biomechanics Group
APPENDIX B 4 – STUDY TWO AND THREE PARTICIPANT RECRUITMENT FLYER

Is Barefoot Running Better?
Runners Needed!

When the majority of people run barefoot they land on the ball of their foot first (fore-foot strike), in shoes, most people land on their heel first (rear-foot strike). It has been suggested that fore-foot strike running is more efficient than rear-foot running, one of the reasons being that the technique allows you to better utilise the spring properties of the arch of your foot. The tendons and ligament in the arch of your foot are capable of recycling significant amounts of energy, therefore reducing the work required from lower limb muscles. We would like to determine whether there is any truth to the claim that runners who fore-foot strike are able to better utilise the spring in the arch of the foot and we are looking for enthusiastic highly trained runners to do this.

Reasons you might choose to participate:
- Find out your maximal aerobic capacity by performing a VO2 max test (gold standard for determining aerobic fitness). - OPTIONAL
- Discover your lactate threshold. - OPTIONAL
- Find out if you use less energy using a rear-foot or fore-foot running technique.
- Have your running technique recorded in 3D (the most accurate movement analysis technique).
- Find out how using store bought orthotics effects your energy cost during running.
- Increase scientific knowledge on the function of the arch of the foot.

We’re looking for people who fit the following criteria:
- Male 17-35 years.
- Run more than 30km a week (the study will require you to run approx. 9km (not continuously) with minimal fatigue at 3.0ms⁻¹ pace).
- No significant lower limb injuries in the last 3 months.
- Relatively neutral feet and orthotics not required for pain free running.
- Wear a size US10, 11 or 12 running shoe.
- Available for 3 hours one day in February to attend the University of Western Australia for testing.

If you fit the above criteria and would like more information about the study or to be involved please contact Sarah Stearne.

Contact person
Sarah Stearne
PhD Candidate
The University of Western Australia
School of Sport Science, Exercise and Health
Phone: 0431 937 482
Email: stears01@student.uwa.edu.au

Chief Investigator
Dr Jonas Rubenson
Assistant Professor
The University of Western Australia
School of Sport Science, Exercise and Health
Phone: 6488 5533
Email: Jonas.rubenson@uwa.edu.au
Ethics approval # RA/4/1/4541
Role of the Arch of the Foot in the Mechanics and Energetics of Running

Information Sheet

Chief Investigators: Dr. Jonas Rubenson, PhD
Assistant Professor, School of Sport Science, Exercise & Health
The University of Western Australia (UWA)
Phone: (08) 6488 5533

Dr. Jacqueline Alderson, PhD
Associate Professor, School of Sport Science, Exercise & Health, UWA
Phone: (08) 6488 5827

Co-Investigator: Sarah Stearne, BSc Hons
PhD Candidate, School of Sport Science, Exercise & Health, UWA
Phone: +61 431 937 482
Email: stearde1@student.uwa.edu.au

Purpose of the Study
Majority of runners use a rear-foot strike running technique, meaning that they strike the ground with the heel of their foot first. However, compared recreational runners, a larger percentage of elite runners use a fore-foot strike running technique, meaning they make initial ground contact with the ball of their foot. It has been suggested that fore-foot strike running is more efficient than rear-foot running. One of the reasons being that the technique allows you to better utilise the spring properties of the arch of your foot.

The arch of the foot is made up of a number of tendons and ligaments which act as springs during running. Tendons and ligaments are similar to elastic bands, when the arch flattens during running they stretch and store energy, when the foot lifts off the ground they recoil returning their stored energy and helping to propel the runner off the ground. This reduces the work required from the lower limb muscles and therefore lowers the energy cost of running. However, it is unknown to exactly what extent the arch helps to reduce the energy cost of running and how foot strike technique plays a role.

This research aims to investigate the function of the arch of the foot by eliminating its ability to collapse through the use of orthotics and measuring the subsequent effects on energy cost and lower limb movement patterns. Habitual rear-foot strike and fore-foot strike runners will be recruited to determine the effect of foot strike technique on arch contribution to running.

What is required of you?
You will be required on two or three (the maximal aerobic fitness test is voluntary) separate occasions; 1) to get orthotics fitted for your feet, 2) for biomechanics testing, 3) to perform a maximal aerobic fitness test (optional).

Orthotics
During the first occasion you will be asked to visit Ian North at Willetton Podiatry Clinic to get two pairs of custom orthotics made for your feet. Your feet will be scanned using ScanAny technology and various
other standard podiatry foot measurements taken, the procedure will take approximately 15 minutes. Your scans will be sent to Orthotech laboratory in Victoria and constructed into orthotics, this will take approximately two weeks.

**Maximal Aerobic Test (optional)**

The maximal aerobic fitness test (VO2 max test) to determine your level of fitness will take place at The University of Western Australia (UWA) Exercise Physiology laboratory in the School of Sport Science, Exercise and Health. The protocol will follow the American College of Sports Medicine guidelines for exercise testing and is used routinely within our school to assess metabolic capacities. You will begin running at an easy pace and the treadmill speed will be increased incrementally until you reach exhaustion. Each speed will be performed for three minutes followed by a one minute rest, during the rest minute heart rate and blood lactate levels will be taken to determine your lactate threshold. A small sample of capillary blood from your fingertip is needed in order to determine your blood lactate levels. Only 5μL (less than a drop) of blood is required each time. To collect the blood your fingertip will be pricked gently with a needle. Due to the small amount of blood required, the prick will be very small and any discomfort experienced should subside within a day. All blood samples will be collected by an experienced researcher and once lactate levels have been demined blood samples will be disposed of in a biohazard waste disposal bin. The maximal aerobic test should take no longer than an hour.

**Biomechanics Testing**

When your orthotics have been constructed, this should be approximately the beginning of February, you will be asked to attend the Sports Biomechanics Laboratory at the School of Sport Science, Exercise and Health at UWA for biomechanics testing. This testing will take approximately three hours during which you will be required to complete a number of different trials on a treadmill. Trials will involve 9 x 5 minute and 3 x 2 minute bouts of running at 3.0ms⁻¹ and 2 x 5 minute bouts of walking, unlimited rest will be given between exercise bouts. In total you will run approximately 9km and walk approximately 700m. It is important that your fitness level is adequate for you to perform this distance with minimal fatigue.

You will be asked to complete all conditions using your natural foot-strike technique and up to four conditions using your non-preferred foot-strike technique.

In order to test the function of the foot arch, you will be asked to run in a number of different conditions: barefoot, in running shoes (provided), with an off the shelf orthotic, a low arch custom made orthotic and a high arch custom made orthotic. All condition will be performed on a level treadmill and three (barefoot, running shoes and the high arch orthotic) conditions will be repeated on an inclined treadmill.

Whilst you are running, lower limb (including foot) movement and oxygen consumption will be measured. Movement will be recorded using specialised infra-red cameras. Twenty-six retro-reflective markers will be placed on your feet, lower limbs and trunk using low allergenic double-sided tape. The markers will be tracked using cameras and converted into a three-dimensional image for computer analysis. High speed video will be used to assess foot movement. To measure oxygen consumption you will be asked to breathe into a mouth piece whilst running (the same mouth piece as the one used during the max test).

On three occasions during the incline shoe only condition a fingertip capillary blood sample will be collected to determine your blood lactate levels. The collection of blood will follow the same procedure as during the max aerobic test and should cause minimal discomfort due to the small amount of blood required. It is important for us to know your blood lactate levels during this trial to ensure you are still exercising aerobically and that the increase in incline has not substantially increased the running effort.
APPENDIX B: PARTICIPANT FORMS

Possible Benefits
Your participation in this research will act to advance the current state of scientific knowledge surrounding the function of the tendons and ligaments comprising the arch of the foot during running. Your participation will also increase understanding of how the arch of the foot is affected by different foot strike techniques and level vs. incline running. Improving our basic understanding of the arch of the foot may have significant benefits for injury prevention, footwear prescription, rehabilitation and performance enhancement for athletic and disabled populations.

As an athlete, you will learn valuable information regarding your running aerobic fitness, technique, and efficiency. The VO2 max test performed in session two is the gold standard method for determining aerobic fitness and will allow you to compare your fitness level to other athletes or to use it as a tool to track your training progress. During the same testing session you will find out your lactate threshold, this is important information for determining race pace and training intensities. You will be provided with a report detailing your VO2max and lactate threshold results and how to interpret them.

During the biomechanics testing in session three, your running technique will be measured using a 3D motion capture system which is currently considered the gold standard method for accurately assessing movement. You will also learn if you use more energy running with a rear-foot or fore-foot strike running technique. Following the conclusion of the study I will provide a personalised running biomechanics report for interested participants. The report will include information on lower limb joint angles and forces when running with different foot strike techniques and orthotic conditions. It will also include corresponding metabolic cost data.

Possible Risks/Discomforts
The maximal aerobic test requires you to exercise to voluntary exhaustion, you may find this level of exhaustion unpleasant. However, this test is commonly used in our laboratory and worldwide with athletes and the general population including children. You are allowed to conclude the test at any time.

Some people find the blood lactate procedure to be unpleasant, as it requires the fingertip to be pin pricked. Only a very small amount of blood is required therefore the pin prick will be very shallow and any discomfort or swelling that may result should subside within a day. This is a standard procedure is routinely performed in our laboratory.

Some people are allergic to specific adhesive tapes, however the tape used to attach the markers is hypoallergenic, and a very small number of people experience a reaction. Any irritation experienced however, will be relatively minor and short term in duration.

On three occasions you will be asked you to run with a foot-strike technique that is not natural to you, however the researchers believe the short duration of the imposed foot-strike condition trials (3 x 5 minutes) will not pose any adverse response to you.

During the testing you will be required to wear a range of shoes insoles,these are designed to prevent the arch of the foot from collapsing. Two of the insole will be made from semi-rigid plastic therefore may cause slight discomfort, however they are no different to the corrective insoles commonly prescribed by podiatrists.

If at any point during the testing sessions you wish to discontinue due to discomfort, you may do so without prejudice.
APPENDIX B: PARTICIPANT FORMS

What to Bring
Maximal Aerobic Test
Please wear lightweight sports clothes and running shoes and bring with you a water bottle and sweat towel. Please do not consume caffeine on the day of the test or eat in the 2 hours prior to testing.

Biomechanics Testing
All you need to bring for the biomechanics testing is a water bottle, sweat towel and wear a pair of short sport shorts. It is preferable that they are short because retro-reflective markers will need to be placed on your thighs to track leg movement. You will be asked to remove your shirt during the testing because markers will also need to be placed on your trunk and hips. You will be provided with a pair of running shoes for the biomechanics testing, however if preferred you may wear your own footwear during pre-testing warm up.

As with the max test please abstain from drinking coffee on the day and do not eat in the 2 hours prior to testing as this will affect your metabolic cost data.

Investigator Responsibilities
All video data will not show your face. All recordings will also be de-identified during collection and stored in a password protected hard drive for later analysis. Names and personal information will be kept strictly confidential. This includes digital or computerised information. Your anonymity will be preserved in any publications of data collected.

Withdrawal
Participation in this study is voluntary and you are free to withdraw from the research at any time without reason or justification. In such cases, your records will be destroyed unless you agree that the researcher may retain and use the information obtained prior to your withdrawal. If you wish to remove yourself from the study, please contact the Chief Investigator at the earliest possible convenience. If you withdraw from the study and you are an employee or student at the University of Western Australia this will not prejudice your status and rights as employee or student of UWA.

Approval to conduct this research has been provided by The University of Western Australia, in accordance with its ethics review and approval procedures. Any person considering participation in this research project, or agreeing to participate, may raise any questions or issues with the researchers at any time.

In addition, any person not satisfied with the response of researchers may raise ethics issues or concerns, and may make any complaints about this research project by contacting the Human Research Ethics Office at The University of Western Australia on (08) 6488 3703 or by emailing to humanresearch@uwa.edu.au

All research participants are entitled to retain a copy of any Participant Information Form relating to this research project.
APPENDIX B 6 – STUDY TWO AND THREE PARTICIPANT CONSENT FORM

Role of the Arch of the Foot in the Mechanics and Energetics of Running

Consent Form

Chief Investigators:  
Dr. Jonas Rubenson, PhD  
Assistant Professor,  
School of Sport Science, Exercise & Health  
The University of Western Australia (UWA)  
Phone: (08) 6488 5533

Dr. Jacqueline Alderson PhD  
Associate Professor,  
School of Sport Science, Exercise & Health  
The University of Western Australia (UWA)  
Phone: (08) 6488 5827

Co-Investigator:  
Sarah Stearne BSc Hons  
PhD Candidate,  
School of Sport Science, Exercise & Health, UWA  
Phone: (08) 6488 1385

Kirsty McDonald BSc  
PhD Candidate,  
School of Sport Science, Exercise & Health, UWA  
Phone: (08) 6488 1385

I __________________________ have read the information provided and any questions I have asked have been answered to my satisfaction. I agree to participate in this activity, realising that I may withdraw at any time without reason and without prejudice.

I understand that all information provided is treated as strictly confidential and will not be released by the investigator unless required by law. I have been advised as to what data is being collected, what the purpose is, and what will be done with the data upon completion of the research.

I agree that research data gathered for the study may be published provided my name or other identifying information is not used.

______________________________  
Participant Signature  
______________________________  
Date

Approval to conduct this research has been provided by The University of Western Australia, in accordance with its ethics review and approval procedures. Any person considering participation in this research project, or agreeing to participate, may raise any questions or issues with the researchers at any time. In addition, any person not satisfied with the response of researchers may raise ethics issues or concerns, and may make any complaints about this research project by contacting the Human Research Ethics Office at The University of Western Australia on (08) 6488 3703 or by emailing hrecresearch@uwa.edu.au. All research participants are entitled to retain a copy of any Participant Information Form and/or Participant Consent Form relating to this research project.
APPENDIX B 7 – STUDY TWO AND THREE TESTING QUESTIONNAIRE

Testing Questionnaire

Name: ___________________________ Date: ___________________________

Natural foot-strike technique (please circle): Rear foot-strike / Fore foot-strike

Blue Orthotic

Please answer the following questions after completing the rigid half arch custom orthotic condition.

Rating of Perceived Exertion (see last page for scale description)

6 7 8 9 10 11 12 13 14 15 16 17 18 19 20

How painful did you find the orthotics?

No Pain 0 1 2 3 4 5 6 7 8 9 10 Worst pain possible

Did you feel you changed your running technique due to the orthotics?

Yes / No (If Yes please explain how below)

Rate your level of overall fatigue.

Not Fatigued 0 1 2 3 4 5 6 7 8 9 10 Very Fatigued

Can no longer participate

Black Orthotic – Walking

Please answer the following questions after completing the rigid half arch custom orthotic condition using your habitual/natural foot strike technique.

Rating of Perceived Exertion (see last page for scale description)

6 7 8 9 10 11 12 13 14 15 16 17 18 19 20

How painful did you find the orthotics?

No Pain 0 1 2 3 4 5 6 7 8 9 10 Worst pain possible

Did you feel you changed your running technique due to the orthotics?

Yes / No (If Yes please explain how below)

Rate your level of overall fatigue.

Not Fatigued 0 1 2 3 4 5 6 7 8 9 10 Very Fatigued

Can no longer participate

Black Orthotic - Habitual/Natural Foot Strike Technique

Please answer the following questions after completing the rigid full arch custom orthotic condition using your habitual/natural foot strike technique.

Rating of Perceived Exertion (see last page for scale description)

6 7 8 9 10 11 12 13 14 15 16 17 18 19 20

How painful did you find the orthotics?

No Pain 0 1 2 3 4 5 6 7 8 9 10 Worst pain possible

Did you feel you changed your running technique due to the orthotics?

Yes / No (If Yes please explain how below)

Rate your level of overall fatigue.

Not Fatigued 0 1 2 3 4 5 6 7 8 9 10 Very Fatigued

Can no longer participate
**APPENDIX B: PARTICIPANT FORMS**

---

**Black Orthotic – Incline/Uphill Running**

Please answer the following questions after completing the rigid full arch custom orthotic condition using your habitual/natural foot strike technique.

**Rating of Perceived Exertion** (see last page for scale description)

<table>
<thead>
<tr>
<th>Rating</th>
<th>0</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
<th>10</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>No Pain</strong></td>
<td>0</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>4</td>
<td>5</td>
<td>6</td>
<td>7</td>
<td>8</td>
<td>9</td>
<td><strong>Pain possible</strong></td>
</tr>
</tbody>
</table>

**How painful did you find the orthotics?**

**Did you feel you changed your running technique due to the orthotics?**

Yes / No  (If Yes please explain how below)

---

**Rate your level of overall fatigue.**

<table>
<thead>
<tr>
<th>Rating</th>
<th>0</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
<th>10</th>
</tr>
</thead>
</table>
| **Not Fatigued** | 0 | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | **Very Fatigued** /
| **Can no longer participate** | 10 |

---

**Rating of Perceived Exertion**

**Borg’s 15 point scale**

Using the scale below rate how intense you found the exercise.

6
7 Very, Very Light
8 Very Light
9
10 Fairly Light
11 Somewhat Hard
12
13 Hard
14
15
16
17 Very Hard
18
19 Very, Very Hard
20 Maximal Exertion

Appendix C – Additional Data

Appendix C 1 – Study One – Joint Moments

Figure A.1 Average ankle frontal, knee transverse and hip transverse moments (N kg$^{-1}$) for habitual forefoot strike (FFS) (black solid), habitual rearfoot (RFS) (grey solid), imposed FFS (black dashed) and imposed RFS (grey dashed) conditions for one complete stride (from foot contact to foot contact). Vertical lines represent toe off in RFS (grey) and FFS (black). Positive values indicate ankle inversion and knee and hip internal rotation. Negative values indicate ankle eversion and knee and hip external rotation. No significant differences in any peak moment data exist (p > 0.05).
APPENDIX C 2 – RETRO-REFLECTIVE MARKER PLACEMENT

Front view

Side view

Front view lower leg

Side view lower leg

Medial and lateral right foot

Magnetic foot markers
### APPENDIX C 3 – STUDY TWO – QUESTIONNAIRE RESULTS

**Table A.1** Questionnaire results.

<table>
<thead>
<tr>
<th>question</th>
<th>group</th>
<th>walk</th>
<th>level run</th>
<th>incline run</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>full arch insole (FAI)</td>
<td>half arch insole (HAI)</td>
<td>full arch insole (FAI)</td>
</tr>
<tr>
<td>How painful did you find the insoles?</td>
<td>RFS</td>
<td>1.4 ± 0.8</td>
<td>2.4 ± 1.8</td>
<td>2.1 ± 1.4</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>1.1 ± 1.1</td>
<td>1.7 ± 1.6</td>
<td>2.0 ± 1.3</td>
</tr>
<tr>
<td></td>
<td>average</td>
<td>1.3 ± 0.9</td>
<td>2.1 ± 1.7</td>
<td>2.1 ± 1.3</td>
</tr>
<tr>
<td>Borg’s Rating of Perceived Exertion</td>
<td>RFS</td>
<td>6.9 ± 0.8</td>
<td>9.8 ± 2.0</td>
<td>9.9 ± 0.8</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>6.8 ± 1.2</td>
<td>10.6 ± 1.3</td>
<td>10.6 ± 1.3</td>
</tr>
<tr>
<td></td>
<td>average</td>
<td>6.8 ± 1.0</td>
<td>10.1 ± 1.7</td>
<td>10.2 ± 1.4</td>
</tr>
<tr>
<td>Did the insoles cause you to alter your technique?</td>
<td>RFS</td>
<td>0.1 ± 0.4</td>
<td>0.0 ± 0.0</td>
<td>0.1 ± 0.3</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>0.0 ± 0.0</td>
<td>0.3 ± 0.5</td>
<td>0.3 ± 0.5</td>
</tr>
<tr>
<td></td>
<td>average</td>
<td>0.1 ± 0.3</td>
<td>0.2 ± 0.4</td>
<td>0.2 ± 0.4</td>
</tr>
</tbody>
</table>

*No statistically significant results were identified. RFS = rearfoot strike, FFS = forefoot strike*
Table A.2: All data from Study Two

<table>
<thead>
<tr>
<th>Condition</th>
<th>Habitual foot strike</th>
<th>Arch Compression (mm)</th>
<th>% Arch Compression restricted</th>
<th>( \dot{V}O_2 ) (ml kg(^{-1}) min(^{-1}))</th>
<th>% ( \dot{V}O_2 ) diff from shoe only</th>
<th>Net ( \dot{V}O_2 ) (ml kg(^{-1}) min(^{-1}))</th>
<th>Estimated energy stored in the arch (J)</th>
<th>To total limb energy contribution to total limb work</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>WALK</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Shoe only</td>
<td>RFS</td>
<td>5.9 ± 2.3</td>
<td>42.2 ± 17.6</td>
<td>13.0 ± 1.5</td>
<td>n/a</td>
<td>6.3 ± 1.2</td>
<td>2.8 ± 2.5</td>
<td>n/a</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>7.4 ± 5.3</td>
<td>47.3 ± 29.6</td>
<td>13.3 ± 1.1</td>
<td>n/a</td>
<td>6.6 ± 0.8</td>
<td>2.4 ± 2.7</td>
<td>n/a</td>
</tr>
<tr>
<td>Average</td>
<td>RFS</td>
<td>6.7 ± 4.1</td>
<td>44.9 ± 24.2</td>
<td>13.2 ± 1.5</td>
<td>n/a</td>
<td>6.5 ± 1.0</td>
<td>2.6 ± 2.5</td>
<td>n/a</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>7.7 ± 4.2</td>
<td>48.1 ± 26.3</td>
<td>13.4 ± 1.0</td>
<td>n/a</td>
<td>6.7 ± 0.8</td>
<td>2.4 ± 2.7</td>
<td>n/a</td>
</tr>
<tr>
<td>Full Arch Insole (FAI)</td>
<td>RFS</td>
<td>-1.1 ± 5.1 *</td>
<td>84.6 ± 22.4</td>
<td>13.1 ± 1.6</td>
<td>1.3 ± 10.3</td>
<td>6.4 ± 1.1</td>
<td>0.3 ± 0.6 *</td>
<td>13.3 ± 14.4</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>2.2 ± 4.1 *</td>
<td>80.5 ± 21.1</td>
<td>13.2 ± 1.4</td>
<td>-0.6 ± 6.2</td>
<td>6.5 ± 0.6</td>
<td>0.2 ± 0.5 *</td>
<td>7.3 ± 22.6</td>
</tr>
<tr>
<td>Average</td>
<td>RFS</td>
<td>0.7 ± 4.8 *</td>
<td>82.4 ± 21.1</td>
<td>13.2 ± 1.5</td>
<td>0.3 ± 6.2</td>
<td>6.5 ± 0.8</td>
<td>0.2 ± 0.5 *</td>
<td>10.0 ± 18.8</td>
</tr>
<tr>
<td><strong>LEVEL RUN</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Shoe only</td>
<td>RFS</td>
<td>9.2 ± 2.1</td>
<td>8.8 ± 9.0</td>
<td>37.0 ± 3.1</td>
<td>n/a</td>
<td>30.3 ± 2.7</td>
<td>112 ± 3.9</td>
<td>n/a</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>12.0 ± 2.4</td>
<td>5.4 ± 7.6</td>
<td>36.1 ± 2.7</td>
<td>n/a</td>
<td>29.4 ± 2.5</td>
<td>123 ± 6.0</td>
<td>n/a</td>
</tr>
<tr>
<td>Average</td>
<td>RFS</td>
<td>10.6 ± 2.6</td>
<td>7.1 ± 8.2</td>
<td>36.5 ± 2.9</td>
<td>n/a</td>
<td>29.9 ± 2.5</td>
<td>118 ± 4.9</td>
<td>n/a</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>11.6 ± 2.7</td>
<td>6.1 ± 7.6</td>
<td>36.1 ± 2.8</td>
<td>n/a</td>
<td>29.9 ± 2.5</td>
<td>118 ± 4.9</td>
<td>n/a</td>
</tr>
<tr>
<td>Half Arch Insole (HAI)</td>
<td>RFS</td>
<td>2.6 ± 4.2 *</td>
<td>63.5 ± 33.0</td>
<td>38.6 ± 3.1 *</td>
<td>4.5 ± 4.6</td>
<td>31.9 ± 2.7</td>
<td>1.7 ± 2.2 *</td>
<td>6.1 ± 4.2</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>5.7 ± 6.1 *</td>
<td>59.3 ± 35.8</td>
<td>37.8 ± 3.4</td>
<td>4.5 ± 5.8</td>
<td>31.8 ± 2.7</td>
<td>2.2 ± 4.8 *</td>
<td>10.8 ± 7.4</td>
</tr>
<tr>
<td>Average</td>
<td>RFS</td>
<td>4.1 ± 5.3 *</td>
<td>61.4 ± 33.4</td>
<td>38.2 ± 3.2 *</td>
<td>4.5 ± 5.0</td>
<td>31.9 ± 2.6</td>
<td>1.9 ± 3.6 *</td>
<td>8.4 ± 6.3</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>2.5 ± 5.8 *</td>
<td>73.0 ± 27.3</td>
<td>38.2 ± 2.8 *</td>
<td>5.9 ± 3.8</td>
<td>31.5 ± 2.6</td>
<td>0.6 ± 1.3 *</td>
<td>10.2 ± 8.1</td>
</tr>
<tr>
<td>Full Arch Insole (FAI)</td>
<td>RFS</td>
<td>-1.3 ± 6.1 *</td>
<td>84.7 ± 22.0</td>
<td>39.3 ± 4.0 *</td>
<td>6.2 ± 4.7</td>
<td>32.6 ± 3.5 *</td>
<td>0.2 ± 0.4 *</td>
<td>7.1 ± 6.3</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>2.5 ± 5.8 *</td>
<td>73.0 ± 27.3</td>
<td>38.2 ± 2.8 *</td>
<td>5.9 ± 3.8</td>
<td>31.5 ± 2.6</td>
<td>0.6 ± 1.3 *</td>
<td>10.2 ± 8.1</td>
</tr>
<tr>
<td>Average</td>
<td>RFS</td>
<td>0.6 ± 6.0 *</td>
<td>78.9 ± 24.7</td>
<td>38.7 ± 3.4 *</td>
<td>6.0 ± 4.2</td>
<td>32.1 ± 3.1 *</td>
<td>0.4 ± 1.0 *</td>
<td>8.7 ± 7.2</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>2.5 ± 5.8 *</td>
<td>73.0 ± 27.3</td>
<td>38.2 ± 2.8 *</td>
<td>5.9 ± 3.8</td>
<td>31.5 ± 2.6</td>
<td>0.6 ± 1.3 *</td>
<td>10.2 ± 8.1</td>
</tr>
<tr>
<td><strong>INCLINE RUN</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Shoe only</td>
<td>RFS</td>
<td>9.1 ± 3.4</td>
<td>11.0 ± 18.5</td>
<td>47.4 ± 3.3</td>
<td>n/a</td>
<td>40.7 ± 2.7</td>
<td>108 ± 4.4</td>
<td>n/a</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>11.9 ± 4.5</td>
<td>10.4 ± 11.2</td>
<td>45.6 ± 2.1</td>
<td>n/a</td>
<td>38.9 ± 1.2</td>
<td>111 ± 5.8</td>
<td>n/a</td>
</tr>
<tr>
<td>Average</td>
<td>RFS</td>
<td>10.5 ± 4.1</td>
<td>10.7 ± 14.8</td>
<td>46.5 ± 2.8</td>
<td>n/a</td>
<td>38.9 ± 1.2</td>
<td>111 ± 5.8</td>
<td>n/a</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>11.9 ± 4.5</td>
<td>10.4 ± 11.2</td>
<td>45.6 ± 2.1</td>
<td>n/a</td>
<td>38.9 ± 1.2</td>
<td>111 ± 5.8</td>
<td>n/a</td>
</tr>
<tr>
<td>Full Arch Insole (FAI)</td>
<td>RFS</td>
<td>-0.3 ± 7.7 *</td>
<td>76.2 ± 33.2</td>
<td>47.5 ± 3.5</td>
<td>0.2 ± 1.5</td>
<td>40.8 ± 2.8</td>
<td>1.3 ± 2.5 *</td>
<td>4.3 ± 6.1</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>5.4 ± 4.9 *</td>
<td>60.8 ± 27.6</td>
<td>46.6 ± 2.5</td>
<td>2.1 ± 4.2</td>
<td>39.8 ± 2.7</td>
<td>0.9 ± 1.7 *</td>
<td>6.9 ± 6.1</td>
</tr>
<tr>
<td>Average</td>
<td>RFS</td>
<td>2.3 ± 6.8 *</td>
<td>68.5 ± 30.6</td>
<td>47.0 ± 3.0</td>
<td>1.2 ± 3.2</td>
<td>40.3 ± 2.7</td>
<td>1.1 ± 2.1 *</td>
<td>5.6 ± 6.1</td>
</tr>
<tr>
<td></td>
<td>FFS</td>
<td>5.4 ± 4.9 *</td>
<td>60.8 ± 27.6</td>
<td>46.6 ± 2.5</td>
<td>2.1 ± 4.2</td>
<td>39.8 ± 2.7</td>
<td>0.9 ± 1.7 *</td>
<td>6.9 ± 6.1</td>
</tr>
</tbody>
</table>

Rearfoot strike (RFS), Forefoot strike (FFS), Average (average of RFS and FFS), Half Arch Insole (HAI), Full Arch Insole (FAI). Statistical analysis was only conducted on raw values not percentages; arch compression (navicular vertical displacement [mm]), \( \dot{V}O_2 \) (ml kg\(^{-1}\) min\(^{-1}\)), Net \( \dot{V}O_2 \) (ml kg\(^{-1}\) min\(^{-1}\)), Estimated energy stored in the arch (J). * significantly different from shoe only condition p<0.05, § significant difference between HAI and FAI p<0.05.
Table A.3 Temporal parameters during walking, level running and incline running across shoe only, half arch insole (HAI) and full arch insole (FAI) in rearfoot strike (RFS) and forefoot strike (FFS) groups.

<table>
<thead>
<tr>
<th>Main Effect</th>
<th>Insole</th>
<th>Foot Strike</th>
<th>Shoe Only</th>
<th>Rearfoot Strike (RFS)</th>
<th>Forefoot Strike (FFS)</th>
<th>Half Arch Insole (HAI)</th>
<th>Rearfoot Strike (RFS)</th>
<th>Forefoot Strike (FFS)</th>
<th>Full Arch Insole (FAI)</th>
<th>Rearfoot Strike (RFS)</th>
<th>Forefoot Strike (FFS)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>stance time</td>
<td>swing time</td>
<td>stance time</td>
<td>swing time</td>
<td>stance time</td>
<td>swing time</td>
<td>stance time</td>
<td>swing time</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.142</td>
<td>0.856</td>
<td>0.151</td>
<td>0.014*</td>
<td>0.070</td>
<td>0.612</td>
<td>0.004*</td>
<td>0.040*</td>
</tr>
<tr>
<td>p value</td>
<td>0.044</td>
<td>0.427</td>
<td>0.014*</td>
<td>0.070</td>
<td>0.612</td>
<td>0.004*</td>
<td>0.040*</td>
<td>0.708</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.30</td>
<td>0.44</td>
<td>0.74</td>
<td>0.04*</td>
<td>0.06*</td>
<td>0.05</td>
<td>0.04*</td>
<td>0.06*</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.28</td>
<td>0.46</td>
<td>0.73</td>
<td>0.04*</td>
<td>0.06*</td>
<td>0.05</td>
<td>0.04*</td>
<td>0.06*</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.26</td>
<td>0.49</td>
<td>0.75</td>
<td>0.04*</td>
<td>0.06*</td>
<td>0.05</td>
<td>0.04*</td>
<td>0.06*</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.77</td>
<td>0.45</td>
<td>1.22</td>
<td>0.04*</td>
<td>0.06*</td>
<td>0.05</td>
<td>0.04*</td>
<td>0.06*</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.74</td>
<td>0.45</td>
<td>1.19</td>
<td>0.04*</td>
<td>0.06*</td>
<td>0.05</td>
<td>0.04*</td>
<td>0.06*</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.76</td>
<td>0.44</td>
<td>1.20</td>
<td>0.04*</td>
<td>0.06*</td>
<td>0.05</td>
<td>0.04*</td>
<td>0.06*</td>
</tr>
</tbody>
</table>

† indicates significant ANOVA main effect (p < 0.05). Where a significant main effect was detected, t-tests were run to determine the location of the effect. * indicates significant t-test difference (p < 0.05) between habitual rearfoot strike (RFS) and habitual forefoot strike (FFS) runners within condition.
Appendix C: Additional Data

Appendix C 6 – Studies Two and Three – Rearfoot Inversion/Eversion

Figure A.2 Rearfoot inversion/eversion angle across the stance phase of running (from foot contact to toe off) during the shoe only (grey) and Full Arch Insole (FAI) (black) conditions of Chapters Four and Five. Data represents average of forefoot strike and rearfoot strike groups ± one standard deviation. No statistically significant ($p < 0.05$) difference was observed between peak eversion angles.
### Table A.4 Percentage each joint contributes to positive and negative lower limb mechanical stance work during shoe only and insole running conditions in rearfoot strike (RFS), forefoot strike (FFS) and both foot strike groups combined (Group).

<table>
<thead>
<tr>
<th>PERCENT JOINT CONTRIBUTION TO TOTAL STANCE WORK</th>
<th>Forefoot Strike</th>
<th>Rearfoot Strike</th>
<th>Group</th>
<th>MANOVA Univariate p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Shoe</td>
<td>Insole</td>
<td>Shoe</td>
<td>Insole</td>
</tr>
<tr>
<td>% of Stance Positive Work</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td>68.2</td>
<td>± 5.5</td>
<td>63.0</td>
<td>± 5.3</td>
</tr>
<tr>
<td></td>
<td>63.1</td>
<td>± 5.4</td>
<td>61.5</td>
<td>± 4.5</td>
</tr>
<tr>
<td>Knee</td>
<td>19.8</td>
<td>± 5.9</td>
<td>23.9</td>
<td>± 4.1</td>
</tr>
<tr>
<td></td>
<td>24.9</td>
<td>± 6.2</td>
<td>25.7</td>
<td>± 4.5</td>
</tr>
<tr>
<td>Hip</td>
<td>21.1</td>
<td>± 4.0</td>
<td>13.0</td>
<td>± 5.3</td>
</tr>
<tr>
<td></td>
<td>12.0</td>
<td>± 3.7</td>
<td>12.8</td>
<td>± 6.0</td>
</tr>
<tr>
<td>% of Stance Negative Work</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td>55.9</td>
<td>± 8.0</td>
<td>40.3</td>
<td>± 7.2</td>
</tr>
<tr>
<td></td>
<td>48.7</td>
<td>± 7.8</td>
<td>35.9</td>
<td>± 6.4</td>
</tr>
<tr>
<td>Knee</td>
<td>30.6</td>
<td>± 7.9</td>
<td>40.5</td>
<td>± 6.6</td>
</tr>
<tr>
<td></td>
<td>37.1</td>
<td>± 7.9</td>
<td>44.2</td>
<td>± 5.4</td>
</tr>
<tr>
<td>Hip</td>
<td>12.0</td>
<td>± 3.7</td>
<td>12.8</td>
<td>± 6.0</td>
</tr>
<tr>
<td></td>
<td>14.2</td>
<td>± 6.0</td>
<td>19.9</td>
<td>± 8.4</td>
</tr>
</tbody>
</table>

* indicates significant difference p < 0.05