RUNNING SELF-OPTIMISATION:
ACUTE AND SHORT-TERM ADAPTATIONS TO RUNNING MECHANICS
AND RUNNING ECONOMY

Submitted by Isabel Sarah Moore to the University of Exeter
as a thesis for the degree of Doctor of Philosophy in Sport and Health Sciences
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Signature: ..................................................................................................
Abstract

The intuitive link between a runner’s gait and their metabolic cost of running, or running economy (RE), has led to many trying to compare the running mechanics of economical runners to those of less economical runners. However using this approach has created controversy about whether running mechanics meaningfully contribute to RE. Additionally only a limited number of studies use a broad, explorative, inter-disciplinary approach, encompassing physiological parameters, flexibility, kinematics, kinetics and muscular activity. The purpose of this thesis was to primarily assess ‘self-optimisation’ through considering acute and short-term adaptations to running mechanics and RE. To assess the biomechanical and physiological mechanisms behind changes to RE three studies were conducted, in addition to a fourth study which investigated biomechanical familiarisation. Study one investigated whether there were any biomechanical or physiological changes in beginner runners after 10 weeks of running and whether any of these changes contributed to a change in RE. There was an 8.4% improvement in RE (224 ± 24 vs. 205 ± 27 mL·kg⁻¹·min⁻¹) and an increase in treadmill time-to-exhaustion (16.4 ± 3.2 vs. 17.3 ± 2.7 min), but no change in \(\hat{V}O_{2\text{max}}\), minute ventilation or heart rate. Several kinematic, kinetic and flexibility measures were found to change over time, but joint moments and stiffness remained similar, with knee extension at toe-off, rearfoot velocity at touch down and timing of peak dorsiflexion explaining 94.3% of the variance in change in RE.

Results from study one suggested that changes in muscular activity might have contributed to kinematic differences, and subsequently an economical gait. Specifically, as joint moments were unchanged after 10 weeks it is possible that muscular coactivation may have changed since varying levels of agonist-antagonist activation can produce the same joint moment. Consequently study two examined the relationship between muscular coactivation and the metabolic cost of running, as thus far there was conflicting evidence. Results showed that in trained, recreational runners greater thigh coactivation was associated with a greater metabolic cost of running. Furthermore, the speed of running was found to affect the level of coactivation at the shank and of the flexor-flexor muscle pair, with less coactivation reported at faster submaximal speeds.

The final part of the thesis focused on a manipulation investigation into barefoot (BFT), minimalist shod (MS) and shod (SH) running. Applying the novel findings from studies one and two to this topical area would hopefully provide new insight into the
BFT running debate. Prior to applying this knowledge of kinematic and muscular activity changes in relation to RE whilst running BFT, an investigation into the time required to become familiar with barefoot treadmill running was needed. Results revealed that barefoot familiarisation was characterised by less plantarflexion and greater knee flexion at touch down, whilst stride length appeared to be adopted instantaneously. Reliability (intra-class correlations) and accuracy (standard error of mean) of the kinematic data appeared strongest once individuals had been running for 20 mins. Furthermore there were no significant differences in the kinematics after 20 mins of running.

The final study considered how changing the levels of proprioception and cushioning (BFT, MS and SH) influenced RE and the potential running mechanics that contributed to any changes in RE. The ramifications of such changes on injury risk were also considered by investigating impact accelerations, effective mass and pronation. Additionally, the effect of naturally changing stride length from a shorter BFT stride to a longer SH stride on RE were examined. Heightened proprioception and no external cushioning (BFT running) appeared to improve RE by at least 5% regardless of stride length, when compared to SH running with a SH stride length. However less proprioception and no external cushioning (MS running) only improves RE, compared to SH running with a SH stride length, when runners run with their SH stride length, rather than their shorter BFT stride length (~2.5% shorter). Improvements in RE are attributed to a lower vertical oscillation and effective mass, greater dependency on efficient, Type I muscles i.e. tibialis anterior, and less plantarflexion at toe-off. However higher impact accelerations, earlier heel off and low pronation angles, suggest there may be an increase in injury risk.

Therefore the findings from this thesis have demonstrated that runners naturally self-optimise the way they run. This is seen both as an acute (changes in footwear) and short-term (10 weeks) response to changing running gait. Study two demonstrated that economical runners appear to use different muscular strategies, with study one and four showing they also adopt specific movement patterns that may promote efficient storage and release of elastic energy. Additionally study three found that runners can become familiar with BFT treadmill running in 20 minutes. It is also important to note that economical biomechanical adjustments do not always favour a reduction in injury risk. But the thesis findings seem to suggest that perhaps performance denominates in terms of self-optimisation, rather than injury prevention.
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Eq. 1  \[ SL = ST \times v\text{tread} \]

Eq. 2  \[ \dot{V}O_2 (mL\cdot kg^{-1}\cdot km^{-1}) = \dot{V}O_2 (mL\cdot kg^{-1}\cdot min^{-1}) \times [60 / \text{speed (km\cdot h^{-1})}] \]

Eq. 3  \[ \text{Signal Energy} = EMG_n - (EMG_{n+1}EMG_{n-1}) \]

Eq. 4  \[ \text{Coactivation} = 2 \times \left[ \frac{\min(EMG_1EMG_2)}{\int \frac{EMG_1 + \int \frac{EMG_2}{x} x 100}{} \right] \]

Eq. 5  \[ \text{Impulse} = \int_{T0}^{TD} F \, dt \]

Eq. 6  \[ \text{FootRF} = \begin{bmatrix} x_{\text{foot}}(i,x) & x_{\text{foot}}(i,y) & x_{\text{foot}}(i,z) \\ y_{\text{foot}}(i,x) & y_{\text{foot}}(i,y) & y_{\text{foot}}(i,z) \\ z_{\text{foot}}(i,x) & y_{\text{foot}}(i,y) & z_{\text{foot}}(i,z) \end{bmatrix} \]

Eq. 7  \[ d_{\text{CoM}} = CoM_{\text{foot}} - JC_{\text{ank}} \]

Eq. 8  \[ d_{\text{CoP}} = CoP_{\text{foot}} - JC_{\text{ank}} \]

Eq. 9  \[ M_{\text{GRF}} = d_{\text{CoP}} \times GRF_{\text{foot}} \]

Eq. 10  \[ M_{\text{WEIGHT}} = d_{\text{CoM}} \times m_{\text{footg}} \]

Eq. 11  \[ M_{\text{EFF}} = d_{\text{CoM}} \times m_{\text{footafoot}} \]

Eq. 12  \[ M_{\text{ank}} = -T_q - M_{\text{GRF}} - M_{\text{WEIGHT}} + M_{\text{EFF}} + I\alpha_{\text{foot}} \]

Eq. 13  \[ K_{\text{joint}} = \frac{\Delta \text{joint moment}}{\Delta \text{joint angle}} \]

Eq. 14  \[ \int_{T_2}^{T_1} F_z(t) = m_{\text{body}}(\Delta v_{\text{CoM}} + g\Delta T) = m_{\text{eff}}((v_2 - v_1) + g(T_2 - T_1)) \]

Eq. 15  \[ \int_{T_1}^{T_2} F_z(t) = \int_{T_1}^{T_2} m_{\text{eff}}(a_{\text{impact}} + g)(t) \]

Eq. 16  \[ m_{\text{eff}} = \frac{\Delta F_z}{(\Delta v + g\Delta T)} = \frac{\Delta F_z}{\Delta T(a_{\text{impact}} + g)} \]
Symbols and Abbreviations

10wkRP \( \text{ten week running programme} \)

\( \Delta F_z \) \( \text{change in vertical force} \)

\( \Delta T \) \( \text{change in time} \)

\( \Delta v \) \( \text{change in velocity} \)

\( \Delta v_{\text{CoM}} \) \( \text{change in vertical velocity of the body’s centre of mass} \)

\%LL \( \text{percentage of leg length} \)

\%\( \dot{V}O_{2\text{peak}} \) \( \text{fractional utilisation of peak oxygen consumption} \)

\%\( \dot{V}O_{2\text{max}} \) \( \text{fractional utilisation of maximal oxygen consumption} \)

\( a_{\text{foot}} \) \( \text{foot angular acceleration} \)

\( a_{\text{impact}} \) \( \text{impact acceleration} \)

\( A_z \) \( \text{vertical acceleration} \)

ATP \( \text{adenosine triphosphate} \)

BF \( \text{biceps femoris} \)

BFT \( \text{barefoot} \)

BMI \( \text{body mass index} \)

\( \text{CoM}_{\text{foot}} \) \( \text{foot centre of mass} \)

\( \text{CoP}_{\text{foot}} \) \( \text{point of force application} \)

\( d_{\text{CoM}} \) \( \text{moment arm from foot centre of mass to ankle joint centre} \)

\( d_{\text{CoP}} \) \( \text{moment arm from point of force application to ankle joint centre} \)

EMG \( \text{electromyography} \)

EMG\(_1\) \( \text{electromyography data of first muscle} \)
EMG \textsubscript{2} electromyography data of second muscle

EMG \textsubscript{TKEO} Teager-Kaiser energy operator electromyography data

EMG \textsubscript{RMS} root mean square of the electromyography data

F force component

FA floating axis

Foot \textsubscript{RF} foot reference frame

F\textsubscript{x} (F\textsubscript{ML}) medio-lateral force

F\textsubscript{y} (F\textsubscript{AP}) anterior-posterior force

F\textsubscript{z} (F\textsubscript{v}) vertical force

F\textsubscript{z}(t) time-varying vertical force

F\textsubscript{zimpact} peak impact force

g gravity

GL gastrocnemius lateralis

GLTA gastrocnemius lateralis – tibialis anterior

GRF ground reaction force

GRF\textsubscript{foot} ground reaction force exerted on foot segment

GRF\textsubscript{r} resultant ground reaction force

GXT graded-exercise test

HR heart rate

i frame number

I moment of inertia

I\textsubscript{leg} leg moment of inertia

iEMG integrated electromyography data
\( i_{\text{prox}} \) medio-lateral axis of proximal segment

\( J_{\text{Cank}} \) ankle joint centre

\( k_{\text{dist}} \) vertical axis of distal segment

\( k_{\text{joint}} \) joint stiffness

\( k_{\text{leg}} \) leg stiffness

\( k_{\text{vert}} \) effective vertical stiffness

\( \text{LCS} \) local coordinate system

\( M_{\text{ank}} \) ankle moment

\( m_{\text{body}} \) body mass

\( m_{\text{eff}} \) effective mass

\( M_{\text{EFF}} \) moment applied by acceleration forces

\( m_{\text{foot}} \) foot mass

\( M_{\text{GRF}} \) moment applied by GRF

\( \text{min} \) minimum

\( \text{MS} \) minimalist shod

\( M_{\text{WEIGHT}} \) moment applied by the weight of the segments

\( \text{MTU} \) musculotendon unit

\( n \) \( n^{th} \) electromyography data point

\( \text{PAR-Q} \) physical activity readiness questionnaire

\( R_{\text{ank}} \) ankle joint reaction force

\( \text{RE} \) running economy

\( \text{RER} \) respiratory exchange ratio

\( \text{RF} \) rectus femoris
<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>RFBF</td>
<td>rectus femoris – biceps femoris</td>
</tr>
<tr>
<td>RFGL</td>
<td>rectus femoris – gastrocnemius lateralis</td>
</tr>
<tr>
<td>ROM</td>
<td>range of motion</td>
</tr>
<tr>
<td>RPE</td>
<td>ratings of perceived exertion</td>
</tr>
<tr>
<td>SENIAM</td>
<td>surface electromyography for the non-invasive assessment of muscles project</td>
</tr>
<tr>
<td>SH</td>
<td>shod</td>
</tr>
<tr>
<td>SL</td>
<td>stride length</td>
</tr>
<tr>
<td>SRT</td>
<td>sit-and-reach test</td>
</tr>
<tr>
<td>SSC</td>
<td>stretch-shortening cycle</td>
</tr>
<tr>
<td>ST</td>
<td>stride time</td>
</tr>
<tr>
<td>T₁</td>
<td>time at touchdown</td>
</tr>
<tr>
<td>T₂</td>
<td>time at impact transient</td>
</tr>
<tr>
<td>TA</td>
<td>tibialis anterior</td>
</tr>
<tr>
<td>TD</td>
<td>touchdown</td>
</tr>
<tr>
<td>TO</td>
<td>toe-off</td>
</tr>
<tr>
<td>T₉</td>
<td>ground reaction force torque vector</td>
</tr>
<tr>
<td>v₁</td>
<td>vertical velocity of the foot at touchdown</td>
</tr>
<tr>
<td>v₂</td>
<td>vertical velocity of the foot at impact transient</td>
</tr>
<tr>
<td>( \dot{V}CO_2 )</td>
<td>carbon dioxide output</td>
</tr>
<tr>
<td>( \dot{VE} )</td>
<td>pulmonary ventilation (expired)</td>
</tr>
<tr>
<td>VL</td>
<td>vastus lateralis</td>
</tr>
<tr>
<td>VLBF</td>
<td>vastus lateralis – biceps femoris</td>
</tr>
</tbody>
</table>
\[
\dot{V}O_2 \quad \text{pulmonary oxygen uptake}
\]
\[
\dot{V}O_{2\text{max}} \quad \text{maximum oxygen uptake}
\]
\[
\dot{V}O_{2\text{peak}} \quad \text{peak oxygen uptake}
\]
\[
\dot{V}O_{2\text{submax}} \quad \text{submaximal oxygen uptake}
\]
\[
v_{\text{tread}} \quad \text{treadmill velocity}
\]
\[
x \quad \text{medio-lateral vector component}
\]
\[
x_{\text{foot}} \quad \text{medio-lateral foot vector}
\]
\[
y \quad \text{anterior-posterior vector component}
\]
\[
y_{\text{foot}} \quad \text{anterior-posterior foot vector}
\]
\[
z \quad \text{vertical vector component}
\]
\[
z_{\text{foot}} \quad \text{vertical foot vector}
\]
Declaration, Communications and Publications

Declaration

The material contained within this thesis is original work conducted and written by the author. The following communications and publications are a direct consequence of this work.

Publications


Presentations


*Invited Speaker*

Acknowledgements

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Chapter 1 Introduction

Physiologists have extensively examined what distinguishes one long distance runner from another and what makes them a better performer. Early research often focused on investigating the maximal oxygen uptake ($\dot{V}O_{2\text{max}}$) as a primary determinant of performance (Daniels & Daniels, 1992) due to elite distance runners having generally high values (70-82 mL·kg$^{-1}$·min$^{-1}$) (Boileau, Mayhew, Riner, & Lussier, 1982; Conley & Krahenbuhl, 1980; Morgan & Daniels, 1994; Saltin & Astrand, 1967). Whilst $\dot{V}O_{2\text{max}}$ correlates well with performance in a heterogeneous group of runners (Pollock, 1977), it cannot distinguish between a group of runners with homogenous $\dot{V}O_{2\text{max}}$ who perform differently (Morgan, Baldini, Martin, & Kohrt, 1989a). This means there must be other underlying factors influencing performance.

Numerous studies have identified that one physiological variable which correlates well with performance, and to a greater degree than $\dot{V}O_{2\text{max}}$, is the rate an individual consumes oxygen at a given speed, termed running economy (RE) (Conley & Krahenbuhl, 1980; Daniels, 1985). If a runner has a good RE they can consume less oxygen whilst at the same steady-state speed as a runner with a poor RE. In a homogenous group of elite runners RE can vary by as much as 30% and it has been identified as a crucial determinant of the superior performance and dominance of Kenyan distance runners in events of 800m and upwards (Larsen, 2003). Evidence shows that these runners can run faster whilst consuming the same amount of oxygen as their Caucasian counterparts, who either exhibit similar or higher $\dot{V}O_{2\text{max}}$ scores (Larsen, 2003; Saltin et al., 1995).

Other determinants of performance, such as $\dot{V}O_{2\text{max}}$ and the ability to utilise a high fraction of $\dot{V}O_{2\text{max}}$ (%$\dot{V}O_{2\text{max}}$), have received greater research attention. Consequently, it is argued that RE is poorly understood (Foster & Lucia, 2007). However, what is known is that RE can not only discern better performers from others in a homogenous group, but also trained individuals from untrained, elite from good and male from female (Bransford & Howley, 1977; Morgan et al., 1995). Physiologists have posited that the volume of training completed is associated with RE (Jones & Carter, 2000), which although is perhaps too
general, suggests it is conceivable that this parameter is “trainable”. Consequently a variety of training interventions have been investigated to try and improve an individual’s RE, such as plyometric (Saunders et al., 2006; Spurrs, Murphy, & Watsford, 2003; Turner, Owings, & Schwane, 2003), strength and resistance (Barnes, Hopkins, McGuigan, Northuis, & Kilding, 2013; Ferrauti, Bergermann, & Fernandez-Fernandez, 2010; Guglielmo, Greco, & Denadai, 2009; Johnston, Quinn, Kertzer, & Vroman, 1997; Jung, 2003; Paavolainen, Hakkinen, Hamalainen, Nummela, & Rusko, 1999; Støren, Helgerud, Støa, & Hoff, 2008), interval (Barnes, Hopkins, McGuigan, & Kilding, 2013; Denadai, Ortiz, Greco, & de Mello, 2006; Franch, Madsen, Djurhuus, & Pedersen, 1998; Slawinski, Demarle, Koralsztein, & Billat, 2001) and altitude training (Saunders et al., 2004c; Saunders, Telford, Pyne, Hahn, & Gore, 2009). The results from such studies have, however, been equivocal (McCann & Higginson, 2008). Some studies have found RE improvements ranging from 2-8% (Johnston, et al., 1997; Paavolainen, et al., 1999; Saunders, et al., 2004c; Spurrs, et al., 2003; Støren, et al., 2008; Turner, et al., 2003), yet several studies show either no effect or a detrimental effect on RE (Ferrauti, et al., 2010; Lake & Cavanagh, 1996; Ramsbottom, Williams, Fleming, & Nute, 1989). It is likely, however, that the training status of the participants partaking is influential on the outcome of the intervention, with the best results stemming from investigations using untrained runners (Saunders, Pyne, Telford, & Hawley, 2004a).

The proposed determinants of RE are varied, ranging from anthropometric to biomechanical (see Saunders et al. (2004a) for an overview) (Figure 1.1). The mechanism proposed as accounting for inter-individual differences or intra-individual improvements in RE depends on which training intervention has been implemented or the researcher’s area of interest. As a result there are several proposed mechanisms, such as: muscle fibre distribution (Kaneko, 1990; Kyrolainen et al., 2003), myosin heavy chain composition (Kyrolainen, et al., 2003), economical movement patterns (Williams & Cavanagh, 1987) and neuromuscular activation via either efficient storage and release of elastic energy (Jones, 2002; Kyrolainen, Belli, & Komi, 2001) or coactivation of muscular activity (Heise, Shinohara, & Binks, 2008). More recently, it has been reported that the choice of, or lack of, footwear can influence an individual’s RE.
The association between footwear and RE is primarily assumed to be a result of altered running mechanics (Nigg & Enders, 2013).

![Diagram](RUNNING_ECONOMY)

**Figure 1.1** Factors affecting running economy (adapted from Saunders et al., 2004a).

The intuitive link between how an individual runs and their RE has led researchers to investigate running mechanics in conjunction with RE in the pursuit of an economical way of moving. Currently there exists much anecdotal evidence provided by running coaches on the technique they believe is the most economical for the performance of their athlete. However the empirical evidence reported by researchers regarding an economical running gait has yet to provide conclusive statements on what can be defined as an economical gait. Consequently, how to improve a runner’s economy through gait training is a question which remains unanswered (Martin & Morgan, 1992).

To try and understand the relationship between running mechanics and RE researchers have often used trained/experienced runners (Cavanagh &
Williams, 1982; Franch, et al., 1998; Slawinski, et al., 2001; Williams & Cavanagh, 1987) or recreational athletes able to run for at least 30 minutes (Lake & Cavanagh, 1996). On the one hand, such investigations have highlighted various variables (see Review of Literature I) that may be associated with RE. Yet on the other hand, they do not inform us of how individuals have developed their gait. Did they initially have an economical gait or did they adopt it through a process of self-optimisation?

The term self-optimisation reflects an individual naturally fine-tuning their running mechanics (Williams and Cavanagh, 1987). For economical self-optimisation, the aim is to minimise their metabolic cost, potentially enhancing performance (Williams and Cavanagh, 1987). It is not known how long this process takes, whether it can be seen in acute manipulations or short-term interventions (2-3 months) or whether a much longer time period is needed. Previously it has been advocated that intra-individual differences in running mechanics, through manipulations or interventions, need to be explored rather than inter-individual differences (Williams and Cavanagh, 1987); as there is evidence that certain kinematic parameters have different levels which are the most economical for one runner but not for another (Cavanagh & Williams, 1982). Investigating the strategies runners’ use whilst self-optimising has received limited attention and warrants further examination. Furthermore, it is likely that runners self-optimise their gait to prevent injury, thus risk factors for injury should also be considered when exploring running mechanics.

It must also be noted that when considering running mechanics the complexity and interactional elements of the variables should not be ignored or overlooked. Rather than focusing on the magnitude of one variable, which may be considered uneconomical, recognising the effects of such a movement on other variables may provide a greater understanding of an economical movement pattern. Therefore approaching the concept of running mechanics and self-optimisation with a broad perspective is crucial to identifying biomechanical parameters that may be associated with RE.

The purpose of this thesis is to examine the nature of biomechanical self-optimisation, in terms of acute and short-term adaptations. First, a literature
review encompassing the kinematic, kinetic and flexibility variables of the lower extremities that may contribute to a runner's RE is presented. Following on from this are reviews of the literature that detail the relationship between RE and muscular activity and the effects of running barefoot (BFT), both in terms of RE and running mechanics (a summary table of the literature is presented in Appendix A). Four experimental chapters are then outlined which address the purpose of this thesis, specifically how runners change their running mechanics to improve their RE, what muscular activation strategies do economical runners use, how long does it take for inexperienced BFT runners to become familiar with BFT treadmill running and finally, what affect does varying cushioning and proprioception have on RE and running mechanics.
Chapter 2 Review of Literature I

2.1 Kinematics and spatiotemporal variables

There are various kinematic and spatiotemporal variables that have been studied in connection with the amount of oxygen a runner consumes. Certain parameters have, however, received more attention than others, specifically vertical oscillation, stride/step length and ground contact time. These will be separately addressed, in addition to research that has investigated a large array of kinematic variables.

2.1.1 Vertical oscillation

The interest in vertical oscillation stems from the link between oxygen consumption ($\dot{V}O_2$) and the energy needed to raise and lower the body's centre of gravity during running (Dugan & Bhat, 2005). It has been advocated for many years that runners should adopt a low vertical oscillation to achieve a good RE (Anderson, 1996) and several research investigations seem to support this belief. Firstly, elite runners exhibited a lower vertical oscillation than good runners (7.6 cm and 8.0 cm respectively) (Cavanagh, Pollock, & Landa, 1977) and secondly, runners who consumed the lowest amount of oxygen (i.e. good economy) had the lowest amount of vertical oscillation compared to runners in the medium and high $\dot{V}O_2$ groups (9.1 vs 9.3 and 9.6 cm respectively) (Williams & Cavanagh, 1987). Both these results, however, were non-significant and due to the small magnitude of the differences within each study it is possible that methodological errors could explain the differences.

Gait manipulation studies have provided unequivocal evidence that running with an exaggerated vertical oscillation leads to an increased submaximal $\dot{V}O_2$ (Egbonu, Cavanagh, & Miller, 1990; Tseh, Caputo, & Morgan, 2008). Both Egbonu and colleagues (1990) and Tseh, and colleagues (2008) found that when participants increased their vertical oscillation (by 3 SDs and 4 SDs, respectively), their $\dot{V}O_2$ also significantly rose by 4.6% (Egbonu, et al., 1990) and 19% (Tseh, et al., 2008). Furthermore running with an exaggerated vertical oscillation is more influential on $\dot{V}O_2$ than the effect of changing the position of
your arms (e.g. hands behind back, hands on head). There have also been cases where successful manipulation to lower vertical oscillation has been achieved using visual and auditory feedback (Eriksson, Halvorsen, & Gullstrand, 2011), but it remains to be seen whether specific modification in this way has a beneficial effect on RE or whether individuals naturally adopt the most economical vertical displacement for their movement pattern.

Slawinski and Billat (2004) reported highly trained runners had a lower vertical oscillation and lower mechanical cost of their centre of mass vertical oscillation than non-trained runners. However this mechanical cost was not related to overall energy cost, which did not differ between groups. This led the authors to hypothesise that humans adopt a certain movement pattern based not just on an optimal metabolic cost, but also as a protective mechanism against injury. As yet the role of self-optimisation to reduce injury has been looked at in separate studies, but rarely in combination with economical gait studies. It must be noted that the energetic cost of running (as measured in Slawinski and Billat’s (2004) study) is not RE per se, but rather the amount of energy spent per unit of distance (di Prampero, 1986), hence it takes into account both aerobic and anaerobic contributions. Researchers often use RE and energy cost of running interchangeably, however a distinction will be made here, with the latter referring to the mechanical description of energy expended per unit of distance and the former to $\dot{V}O_2$ per unit distance.

A review conducted by Anderson (1996) advocates a low vertical oscillation to be beneficial to RE, yet this is contradicted by a combination of evidence from biomechanical and physiological studies. Findings suggest that females have a significantly lower vertical oscillation than males (Williams, Cavanagh, & Ziff, 1987), but are also considered to be less economical runners (Bransford & Howley, 1977). Furthermore technique changes that result in a decrease in vertical oscillation have been shown to negatively affect RE (Dallam, Wilber, Jadelis, Fletcher, & Romanov, 2005; McMahon, Valiant, & Frederick, 1987). Specifically, Dallam and colleagues (2005) instructed runners to adopt a ‘Pose technique’, whereby runners are encouraged to vertically align their body (Arendse et al., 2004), so their body mass is over the ball of their foot at touchdown (TD), and then to ‘fall forwards’ onto the other foot, using
gravitational torque (Fletcher, Bartlett, Romanov, & Fotouhi, 2008). However, this technique also caused a significant decrease in the participants’ stride lengths, which may have been detrimental to RE (see 2.1.2 Stride/step length). Additionally, McMahon et al. (1987) found that instructing runners to exaggerate their knee flexion during stance (‘Groucho running’) led to a concomitant decrease in vertical oscillation and 50% rise in RE.

A major limitation that must be acknowledged when comparing studies examining vertical oscillation is the differing methodologies used to obtain the displacement of the centre of gravity, sometimes referred to as the centre of mass (Eriksson, et al., 2011; Williams & Cavanagh, 1987). To-date researchers have used the hip marker (Arendse, et al., 2004; Tseh, et al., 2008), the top of the head (Williams, et al., 1987) and the neck (Dallam, et al., 2005; Lake & Cavanagh, 1996). Additionally, others have considered either affixing an abdominal marker (Williams & Cavanagh, 1987) or calculating a position between the shoulder and hip markers (Cavanagh, et al., 1977). The ‘gold standard’ of calculating the centre of gravity position based on individual body segments is rarely conducted in research, possibly due to laboratory limitations.

2.1.2 Stride/step length

Stride length, the distance between successive touch downs of the same foot (Dugan & Bhat, 2005), and step length, the distance from TD of one foot to TD of the opposite foot (Dugan & Bhat, 2005), have often been used interchangeably in early research. However, a distinction should be made and thus the above definitions will apply to any subsequent references to step and stride length.

Stride/step length and stride frequency are often used as the only biomechanical parameters measured when assessing running gait, possibly due to the ease of data collection (Ferrauti, et al., 2010; Franch, et al., 1998; Kerdok, Biewener, McMahon, Weyand, & Herr, 2002). This allows researchers to connect their results to biomechanical outcomes and has led to strong evidence that modification of either step or stride length can affect RE, but
neglects to address whether specific kinematic adjustments have taken place which may occur without changes in step length.

Experimentally altering stride length has led authors to conclude that well-trained runners freely chose to adopt near to/ the most optimal stride length for their metabolic cost (Cavanagh & Williams, 1982; Hogberg, 1952). Early research suggested that over-striding had a greater $\dot{V}O_2$ demand than under-striding, but both were detrimental to RE (Hogberg, 1952). Hogberg (1952) believed that over-striding causes a greater $\dot{V}O_2$ increase than under-striding because the centre of gravity is raised higher than normal. The results from Cavanagh and Williams (1982) suggest that runners with long legs and short strides find decreasing their stride length above their freely chosen stride length causes a greater increase in RE than increasing their stride length above their freely chosen stride length. Whereas, runners with long strides and short legs find the oxygen demand greater when increasing rather than decreasing their stride length. The reason for this however is not known and was not discussed by the authors. A possible explanation could however be derived from walking manipulation studies.

Several investigations into how individuals achieve a lengthening and/or shortening of step/stride length during walking have been conducted (Patla, Robinson, Samways, & Armstrong, 1989; Varraine, Bonnard, & Pailhous, 2000; Warren, Young, & Lee, 1986). During walking, under-striding is achieved by decreasing the time spent in the swing phase (Varraine, et al., 2000). However, for over-striding it is believed that walkers increase their propulsive force and their swing duration (Varraine, et al., 2000). Researchers have posited that this is due to biomechanical and energetic optimisation rather than neural processes. A thorough analysis into running over- and under-striding would need to incorporate kinematic and muscular activity measures to address whether there are universal patterns of movement or inter-individual differences, which may equate to the different $\dot{V}O_2$ responses reported by Cavanagh and Williams (1982).
2.1.3 Lower extremity kinematics

Correlational studies provide the bases for kinematic descriptors of RE, with Williams and colleagues presenting data for both elite male and female distance runners (Williams & Cavanagh, 1986, 1987; Williams, et al., 1987). A comprehensive investigation was conducted by Williams and Cavanagh (1987) as they covered at least 29 kinematic, kinetic and power variables in their initial analysis. By using factor analysis they determined which of these variables provided the strongest association with RE. Thirty-one well-trained runners participated and were split into three groups representing: good (low \( \dot{V}O_2 \)), average (medium \( \dot{V}O_2 \)) and poor (high \( \dot{V}O_2 \)) economy; this allowed the authors to not only test for significant differences but also to report observable trends in the data between groups. Three variables were found to explain 54% of the variance in RE, of these two were kinematics (shank angle at TD and plantarflexion at toe-off (TO)), with the other being net positive power. The overall kinematic findings from their data suggest that the good economy runners had less wrist excursion, a slower knee flexion velocity during swing, greater knee flexion during support, less plantarflexion at TO, a greater forward lean and had a greater shank angle to the vertical at TD (leg ahead of vertical) compared to the poor economy group.

Further running mechanics believed to be associated with a reduced \( \dot{V}O_2 \) are greater plantar flexion velocity (Williams & Cavanagh, 1986), greater horizontal heel velocity at TD (Williams & Cavanagh, 1986), greater maximal thigh extension angle with the vertical (Williams & Cavanagh, 1986), slower knee flexion velocity during swing (Williams, et al., 1987), greater dorsiflexion and faster dorsiflexion velocity during stance (Williams, et al., 1987) and less knee extension at TO (Williams & Cavanagh, 1986; Williams, et al., 1987). Thus far, the kinematics examined have been in the sagittal plane and whilst this may be the predominant plane of motion for running, movements occur in three dimensions and, in particular, there is a suggestion that the frontal plane kinematics should not be ignored. Hasegawa, Yamauchi and Kraemer (2007) concluded that higher inversion of the ankle at TD, in mid-foot and forefoot strikers, might contribute to a better RE. Whilst this may have been a rather
tenuous implication drawn from their data investigating strike patterns in road running, in which no physiological data were collected, it highlights the need to incorporate three-dimensional biomechanical data. Consequently further research explicitly investigating RE and rearfoot movement is required to help provide substantiated conclusions.

Evidence suggests that achieving a less extended leg at TO is associated with a better RE. It appears that there are two different ways of accomplishing a more flexed leg: less plantarflexion or less knee extension. Williams and Cavanagh (1987) identified plantarflexion at TO as one of three variables contributing to 54% of the variance in RE, however, its unique contribution was not reported. Runners with good economy had, on average, 6.4° less plantarflexion than poor economy runners. In support of its potential connection with RE is the evidence that elite runners have less plantarflexion at TO than good runners, suggesting a biomechanical difference between ability levels (Cavanagh, et al., 1977). Additionally male distance runners show an association between knee extension and RE, with less knee extension at TO signifying a better economy (Williams & Cavanagh, 1986). Whilst these represent kinematics specific to males, Williams, Cavanagh and Ziff (1987) investigated the running mechanics of elite female distance runners and confirmed the relationship between RE and a less extended leg through less knee extension. This kinematic variable appears to be one of the only variables associated with RE in more than one study.

Explanations of why certain variables contribute to an economical gait have not been forthcoming, with early research instead focusing on merely describing the kinematic and kinetic traits of economical runners. That said, researchers have still endorsed a number of specific running mechanics as being economical, such as less range of motion (ROM) at the ankle which results in less plantarflexion at TO (Anderson, 1996). Even with the limited scientific explanations there have been recent attempts of incorporating biomechanical principles into constructing different running forms, such as ‘Pose method’, Chi running and BFT running. There have been contrasting results regarding the success of such running techniques on improving RE, with some researchers
tending to focus on injury prevention rather than performance (Arendse, et al., 2004; Daoud et al., 2012; Lieberman et al., 2010).

2.1.4 Contact time

Contact time has been associated with the metabolic demand of running in a variety of animal species (Kram & Taylor, 1990). The evidence suggests that the cost of running is primarily determined by the cost of supporting the animal’s weight and the time course of force application, with recent human studies speculating that shorter contact times contribute to a better RE (Hasegawa, Yamauchi, & Kraemer, 2007). However the correlational evidence between contact time and RE, in humans, is equivocal. Several studies have failed to find a significant relationship between the two (Kyrolainen, et al., 2001; Storen, Helgerud, & Hoff, 2011; Williams & Cavanagh, 1987), whilst Williams and Cavanagh (1986) found good RE (low $\dot{V}O_2$) to be associated with longer contact times and Nummela, Keronen and Mikkelsson (2007) found the opposite to be true, better economy was related to shorter contact times ($r = 0.49$). The mechanism which explains why longer or shorter is better is actually similar. Given that the inverse of contact time (1/contact time) is directly proportional to the metabolic cost of locomotion per unit of body weight, fast force production is suggested to involve recruiting metabolically expensive fast twitch fibres (Kram & Taylor, 1990; Roberts, Kram, Weyand, & Taylor, 1998) and therefore is not beneficial to the runner. Notwithstanding this explanation, Nummela et al. (2007) argue that short contact times are required for economical running, as force is produced quickly and runners undergo shorter braking phases which is when they lose the greatest amount of speed. Furthermore, Bushnell and Hunter (2007) demonstrated that during maximal sprinting, trained sprinters had shorter contact times than distance runners, suggesting that contact times can be modified to produce the best performance in a specific discipline, as contact time and maximal running speed are significantly related ($r = -0.52$) (Nummela, et al., 2007). Recent evidence also suggests that slower runners (positioned lower down in a half-marathon) had longer contact times (Hasegawa, et al., 2007). The authors hypothesised that contact time was therefore related to
economy, yet it must be noted that as speed increases there is a concomitant
decrease in contact time (Nummela, et al., 2007), meaning faster runners would
be expected to have shorter contact times.

Additional support for the relevance of contact time comes from Kong and de
Heer (2008) who studied Kenyan distance runners and found that they
exhibited very short ground contact times when compared to the literature. The
authors posited that this may relate to good RE since there is less time for the
braking force to decelerate the forward motion of the body and thus time spent
decelerating may contribute to their performance. This combined with evidence
that forefoot strikers have shorter contact times than rearfoot strikers, but spend
the same amount of time decelerating and have similar RE as the rearfoot
strikers (Ardigo, Lafortuna, Minetti, Mognoni, & Saibene, 1995; Perl, Daoud, &
Lieberman, 2012), implies that it may be the decelerating component of contact
time, rather than simply the time spent in contact with the ground which is
important to RE.

2.2 Kinetics

2.2.1 Ground Reaction Force

Ground reaction force (GRF) may be influential to RE because a higher GRF
may necessitate more intense muscular contributions to control segmental
movements and stabilise body position during the support phase, which could
result in greater metabolic demands from the involved muscles. The results
from Williams and Cavanagh (1987) support the above hypothesis, as they
found that runners with good economy had significantly lower vertical impact
forces ($F_{\text{zimpact}}$) than runners with poor economy. Further to this, they found that
$F_{\text{zimpact}}$ was positively correlated with RE ($r=0.56$) in male runners, meaning a
greater $F_{\text{zimpact}}$ was associated with higher $\dot{V}O_2$. However, as there was also a
trend for greater vertical oscillations in the poor RE group it is hard to determine
whether it is the influence of kinetics or kinematics/spatiotemporal variables that
influence RE. In contrast to the results found by Williams and Cavanagh (1987),
a recent study failed to find a significant relationship between RE and $F_{\text{zimpact}}$
(Adelson, Yaggie & Buono, 2005) in 35 recreational runners (gender
unspecified), thus contradicting previous evidence. However consideration of the likelihood of sustaining an overuse injury is needed, as researchers have argued that runners who have high $F_{\text{zimpact}}$ may be more likely to develop these types of injuries than those with lower values (Hreljac, 2004; Nigg, Denoth, & Neukomm, 1981). Therefore lower $F_{\text{zimpact}}$ may be a gait adaptation which minimises the risk of injury rather than specifically lowering $\dot{V}O_2$, which follows on from the earlier suggestion that an individual's running gait may develop through a process of self-optimisation considering both $\dot{V}O_2$ and injury risk factors.

Heise and Martin (2001) reported moderate correlations between RE and both total and net (above body weight line) vertical impulse ($r = 0.62$ and 0.60, respectively). The combined influence of the vertical force ($F_z$) and the time course of force application explained 38% of the inter-variability in RE. The authors interpreted this to mean that runners who are less economical have greater amounts of wasteful vertical motion. However vertical oscillation was not measured, so this cannot be verified. Another interpretation of the data would be that the greater impulse is a function of a greater change in momentum and, thus, vertical velocity change. This would mean that the less economical runners had a faster change in their vertical velocity rather than overall motion.

Previously, it has been argued that the vertical GRF was a major determinant of the metabolic cost during running (Farley & McMahon, 1992; Kram & Taylor, 1990), yet this does not always appear to be the case. Therefore, the other components of GRF should not be ignored, especially as Kyrolainen, Belli and Komi (2001) revealed that the average horizontal component ($F_y$) of the braking force was the main factor (81%) from 3D force parameters to explain $\dot{V}O_2$. Additionally, the medio-lateral force ($F_x$) has been investigated in conjunction with RE. Research has found a smaller peak $F_x$ to be associated with a better RE, yet net $F_x$ impulse was similar between RE groups (Williams & Cavanagh, 1986, 1987). The reason for this is not apparent, as theoretically a greater net $F_x$ impulse suggests that a runner is zig-zagging rather than solely generating force in the direction of progression (Lafortune, Valiant, & McLean, 2000), meaning net $F_x$ impulse rather than peak $F_x$ force would be expected to exhibit differences across RE groups.
The argument that vertical GRF is more important than the other GRF components is brought about as it opposes gravity, so runners are required to generate vertical forces to lift their body off the ground. Whilst evidence supports the notion that gravity is the primary determinant of $F_z$ during running, it is also reported to indirectly influence the generation of horizontal, anterio-posterior forces ($F_y$) (Chang, Huang, Hamerski, & Kram, 2000). Chang and colleagues (2000) altered the amount of inertial and gravitational force impeding runners and found that the combination of the two forces increased $F_y$ impulses by 28%, yet on their own inertial forces only increased $F_y$ impulses by 10%. The difference between these two conditions, of 18%, was concluded as resulting from the gravitational force. Although, theoretically, this seems counterintuitive it led Chang et al., (2000) to consider the resultant force vector. They identified that runners, whilst under various gravity and inertia conditions, tried to constantly align the resultant force vector with their leg axis. The effect on RE was not investigated, however, it was postulated that such alignment would have important metabolic and mechanical consequences. For example, such alignment would minimise muscle moment arms, which Scholz and colleagues (2008) identified as being related to RE, and consequently reduce the muscular activity needed to produce these moments and hence lower the metabolic cost (Chang, et al., 2000). Therefore, runners with a better RE could potentially have a better alignment of the resultant force vector rather than lower peak forces in each dimension. Additionally minimising moment arms, and potentially therefore the moments they create, can help reduce the risk of injury (McClay, 2000), thus strengthening the notion that runners develop an ‘optimal’ movement pattern based upon economy of motion and injury prevention.

Chang and Kram (1999) concur with the proposal that the GRF components should not be regarded as independent determinants of metabolic cost. By using wind impedance during steady state running they highlighted how metabolically expensive the generation of propulsive $F_y$ was per unit of force, even though it is not against gravity. Additional support is provided by Storen, Helgerud and Hoff (2011) who, when considering the GRF as individual components, failed to identify any relationship with RE, but when the $F_z$ and $F_y$ peak forces were summed together there was a significant relationship with
both 3 km performance \( (r = -0.71) \) and RE \( (r = 0.66) \). Consequently, the suggestion by Chang et al. (2000) of needing to consider the horizontal and vertical forces together rather than as separate, independent entities appears an appropriate course of action to take in future research. However breaking the GRF down into its three components should not be overlooked as providing important information, regarding loading, propulsion and lateral force production.

2.2.2 Stiffness

A leg-spring model was developed by McMahon and Cheng (1990), whereby the leg was represented as a simple spring during running. The maximum change in vertical position of the centre of mass during stance (compression of the spring) is calculated to determine the effective vertical stiffness \( (k_{\text{vert}}) \) and the deformation of the leg is determined as the difference between the length of the uncompressed spring (standing leg length or hip height) to maximum compression (minimum hip height during stance) \( (k_{\text{leg}}) \). Lowering \( k_{\text{vert}} \) results in a 50% increase in \( \dot{V}O_2 \) (McMahon, et al., 1987), with Heise and Martin (1998) reporting a significant negative relationship \( (r = -0.48) \) between \( \dot{V}O_2 \) and \( k_{\text{vert}} \). Therefore runners that are more compliant are less economical. However they found no relationship between \( k_{\text{leg}} \) and \( \dot{V}O_2 \) (Heise & Martin, 1998), perhaps because this parameter is kept constant during running, regardless of speed (He, Kram, & McMahon, 1991; McMahon & Cheng, 1990). In contrast, there is also evidence that when surface stiffness is substantially lowered (945.7 to 75.4 kN·m\(^{-1}\)), \( \dot{V}O_2 \) decreases (12%) and \( k_{\text{leg}} \) increases (29%) (Kerdok, et al., 2002). The authors acknowledged that from their findings mechanisms behind changes in \( k_{\text{leg}} \) could not be identified as running kinematics remained fairly stable. It is possible that joint stiffness \( (k_{\text{joint}}) \) and lower limb posture contributed to these changes (Farley & Morgenroth, 1999; Kerdok, et al., 2002), in addition to contributing to the reduction in metabolic cost.

To the author’s knowledge no studies have directly investigated the relationship between \( k_{\text{joint}} \) and RE, therefore it is difficult to decipher whether decreasing or increasing \( k_{\text{joint}} \) would be beneficial. That said, ankle stiffness appears to be a
characteristic of the activity (i.e. running or sprinting) rather than specific to an individual (Stefanyshyn & Nigg, 1998), suggesting that it may not directly influence RE. On the other hand, a stiffer joint may be able to facilitate the transmission of muscle force to bone, enhancing the efficiency of the stretch-shortening cycle (SSC) (see 2.3 Flexibility and stretch-shortening cycle) (Kubo et al., 2007). Through enhancing the SSC, the metabolic cost of running could be lowered and RE improved. Further research into joint stiffness and RE is required before conclusive statements can be made.

2.3 Flexibility and stretch-shortening cycle

Flexibility has been investigated by numerous researchers with somewhat contradictory results regarding its relationship with RE. One of the most utilised flexibility measurements is the sit-and-reach test (SRT) which assesses lower back and hamstring flexibility (Heyward, 2006). Currently, evidence suggests that those who are the least flexible have better running economies (Gleim, Stachenfeld, & Nicholas, 1990; Jones, 2002; Trehearn & Buresh, 2009). This strong relationship is evident in both elite (r = 0.68) (Jones, 2002) and trained endurance runners (r = 0.83) (Trehearn & Buresh, 2009). In contrast, several researchers have failed to find an association between RE and SRT flexibility in collegiate track athletes and well-trained and sub-elite distance runners (Beaudoin & Whatley-Blum, 2005; Craib et al., 1996; Mojock, Kim, Eccles, & Panton, 2011).

Researchers have also assessed numerous other lower extremity and upper body rotations and flexions/extensions, in addition to the SRT (see Craib et al., (1996) for descriptions). Of the measures used by Craib et al. (1996) only dorsiflexion and external hip rotation flexibility were significantly related to RE, such that being less flexible was associated with having a better RE, and further analysis revealed that these two measures explained 47% of the variance in RE. Yet based on similar measures Beaudoin and Blum (2005) did not find any association between flexibility and RE. Further research that failed to find an association between flexibility and RE have shown that a static stretching (Allison, Bailey, & Folland, 2008; Hayes & Walker, 2007; Mojock, et al., 2011), a
progressive stretching (Hayes & Walker, 2007) or a dynamic stretching (Hayes & Walker, 2007) protocol immediately prior to running or a 10 week chronic stretching protocol (Nelson, Kokkonen, Eldredge, Cornwell, & Glickman-Weiss, 2001) does not change RE, but does result in an increased SRT flexibility. Nevertheless, recent results suggest that runners should be advised against performing static stretches immediately prior to running, as it can in fact be detrimental to performance and energy cost (Lowery et al., 2013; Wilson et al., 2010). This appears to be the case even when both hip flexion and hip extension stretches are performed, contradicting the argument posed by Godges and colleagues (Godges, Macrae, Longdon, Tinberg, & Macrae, 1989; Godges, MacRae, & Engelke, 1993), which states that end-range stretch in the same plane that the muscles will be utilised in may help RE. It must also be acknowledged that studies tend to focus on hip and/or thigh flexibility, with very few investigating calf flexibility even though it undergoes great tensile stress and is a primary plantarflexor muscle contributing to the push-off phase of running (Kibler, Goldberg, & Chandler, 1991).

Saunders et al. (2004a) suggest that there is an optimal level of flexibility whereby RE can benefit. Similarly, Nelson et al. (2001) proposed that there may be a certain threshold or percentage change in flexibility that needs to be reached before improvements in RE would be found. From the various results presented above it is possible that this level or threshold may be habitually achieved in some participants whereas others may require stretching immediately prior to exercise to reach this optimal level. Previous studies have only looked at manipulating the level of flexibility and monitoring RE, but have not looked at whether a training intervention to improve RE also influences flexibility.

Throughout the literature there is a common suggested mechanism that is provided as an explanation regarding why inflexibility may be more beneficial than flexibility for RE. Researchers suggest that if individuals are less flexible then they have a more efficient elastic energy storage in their muscles and tendons during the eccentric, absorption phase of ground contact, and thus, more elastic energy can be released during the concentric, propulsive phase of
ground contact (Jones, 2002). This storage and release of elastic energy by the muscles and tendons is known as the SSC (Komi, 1984).

The SSC consists of three phases: preactivation, eccentric and concentric. Its primary purpose is to enhance the performance of the final phase, the concentric contraction. The SSC ultimately helps achieve a greater force production in the muscles, compared to a purely concentric action (Komi, 2000). Research has demonstrated that by including a short run-up to a drop jump there is an increase in the muscular activity during the preactivation phase, which contributes to a greater eccentric muscular activity and potentially to the higher power output during the concentric contraction (Ruan & Li, 2010). Relating specifically to running, performing a one mile run after a static stretching protocol resulted in higher muscular activity of the gastrocnemius (Lowery, et al., 2013). The authors believed the change in muscle activity was due to a decrease in energy efficiency, which is argued to be associated with an increase in flexibility. Efficiency, and potentially RE, is believed to be greatest when runners are less flexible, as this can lead to the musculotendon unit (MTU) being stiffer and therefore it can yield a greater amount of stored elastic energy (Nelson, et al., 2001). Although this theory has merits and has been applied by many researchers to explain their results, few have incorporated any form of measurement to assess the SSC. This is primarily because accurate and situation-specific measurements are difficult to obtain, as yet information can only be inferred from counter-movement and standing jumps or isokinetic dynamometer tests. It may be possible to infer detail about the SSC from the timing of joint angles during running. For example, peak dorsiflexion would identify the switch between eccentric and concentric contraction of the gastrocnemius. Thus, timing of peak angles may have some relevance in considering an economical running gait and require consideration.

2.4 Summary

The present research will investigate running self-optimisation and look to understand the influence of lower extremity running mechanics on short-term changes in RE. Currently, little is known regarding the development of an
individual’s running gait due to a lack of research conducted using beginner runners. This is required to understand running mechanics at the other end of the spectrum to elite/trained runners. Moreover, previous research has primarily focused on correlational evidence from inter-individual comparisons rather than intra-individual comparisons. Therefore, the first study will aim to analyse the gait of runners over a 10-week period at a detailed level, involving kinematics, kinetics and flexibility measures, in combination with various physiological factors. This is needed to build a comprehensive picture of how individuals develop their gait and what aspects of such modifications are economical or uneconomical.

2.5 Aims I

The aim of this first part of the thesis was to understand the underlying running mechanics that may contribute to short-term changes in RE, specifically whether short-term RE improvements are due to biomechanical or physiological mechanisms. It follows therefore that the specific aims are to:

1) Address whether increases in training volume will improve RE in untrained, beginner runners.

2) Determine whether short-term changes in RE can be attributed to running mechanics or running physiology.

2.6 Hypotheses I

There were no specific hypotheses formulated in chapter 6 due to its exploratory nature.
Chapter 3 Review of Literature II

After conducting the first experimental study (see Chapter 6) analysis of the data highlighted the need to also consider the role of muscular activity with regards to RE. Whilst the kinematics and kinetics could explain, in part, the short-term changes in RE, it was apparent that to further the understanding of an economical running pattern the underlying muscular activity required consideration. It is possible that changes in such activity contributed to the slower rates of movement and/or the alterations in the leg angle at toe-off. Prior to examining the relative contributions of muscular activity and kinematics to RE, it was first necessary to examine the muscular activity strategies that are adopted when running.

3.1 Muscular Activity and Oxygen Consumption

Electromyography (EMG) can identify the muscular activity of specific muscles used in dynamic movements, such as running. Through using EMG the temporal profile and amplitude of muscular activation can be determined. Furthermore, by examining the role and activations of multiple lower limb muscles an inference about the active stiffness of joints in the lower limb can be made. Physiologists have also utilised EMG data in trying to investigate the mechanism responsible for the \( \dot{VO}_2 \) slow component (Green et al., 2010; Saunders et al., 2000; Scheuermann, Hoelting, Noble, & Barstow, 2001). When an athlete performs exercise above their lactate threshold, producing a sustained lactic acidosis, their \( \dot{VO}_2 \) rises to levels above that predicted from the \( \dot{VO}_2 \) – work rate relationship performed in the sub-lactate threshold domain (Marsh, Ellerby, Carr, Henry, & Buchanan, 2004; Scheuermann, et al., 2001). This is known as the \( \dot{VO}_2 \) slow component, the onset of which is not immediate but delayed. Mechanisms behind the \( \dot{VO}_2 \) slow component remain elusive; however promising evidence has recently been reported connecting muscular activity and \( \dot{VO}_2 \) (Lewek, et al., 2012).

In a review paper, Saunders et al. (2004a) covered many factors affecting RE, suggesting an interdisciplinary approach is required to understand the area.
However, whilst they comprehensively analysed many factors (Figure 1.1), they failed to acknowledge the role that muscular activity may play. Instead they incorporated only snippets of information regarding EMG data, relating it to kinematic and kinetic variables. The activity of the muscles, in particular the lower limb muscles, deserve consideration when trying to understand what makes one running gait economical and another uneconomical, as evidence concerning its contribution to $\dot{V}O_2$ is growing. Further to this, reports specifically suggest that muscular activity plays a role in gearing the leg for impact, through preactivation prior to TD, termed ‘muscle tuning’ (Boyer & Nigg, 2004, 2007). Thus it is possible that kinematic changes occurring at TD that may contribute to an improved RE, such as those found in chapter 6, are a result of muscle tuning.

Findings have suggested that higher muscular activity is a contributory factor in the greater $\dot{V}O_2$ during both fatiguing runs and running at faster speeds (Abe, Muraki, Yanagawa, Fukuoka, & Niihata, 2007; Kyrolainen, et al., 2001). Abe et al. (2007) concluded that an increased muscular activation during the concentric phase of the SSC contributed to a higher energetic cost (gross $\dot{V}O_2$ above resting/ running velocity) during a 90 min fatiguing run. This suggests that the SSC had become less efficient as the muscles generated more activity in the concentric phase without a concomitant increase of eccentric activity. The eccentric-concentric ratio can help explain inter-individual variations in, and is significantly related to, the energy cost of running (Bourdin, Belli, Arsac, Bosco, & Lacour, 1995; Modica & Kram, 2005) and therefore potentially to $\dot{V}O_2$. Thus, it seems $\dot{V}O_2$ and muscular activity could be associated and it is possible that muscle activity is a determinant of $\dot{V}O_2$.

In support of the promising link between muscle activity and $\dot{V}O_2$, Green and colleagues (2010) managed to not only identify specific muscles as contributing to the $\dot{V}O_2$ slow component exhibited in cyclists during heavy intensity cycling, but also suggested that the onset activation and temporal profiles of muscles are crucial to the muscle’s contribution to the slow component phase. Their results imply that the vastus lateralis (VL) muscle contributes to the development of the slow component whereas the muscles acting around the
ankle (tibialis anterior (TA), soleus, gastrocnemius lateralis (GL) and medialis) may be particularly important to initiation of the $\dot{V}O_2$ slow component. Biomechanically, further supporting evidence is provided by research investigating the effect of surface stiffness on $\dot{V}O_2$ (Pinnington, Lloyd, Besier, & Dawson, 2005). Pinnington et al. (2005) attributed the increase in $\dot{V}O_2$ found in previous studies for sand running compared to running on a firm surface, such as grass or carpeted wooden floor, to a change in the magnitude of muscle activation. Their results suggest that it is primarily increased activation of muscles involved in greater hip and knee range of motion i.e. quadriceps and hamstrings, that translates into higher $\dot{V}O_2$ levels.

There have also been significant correlations reported between performance and lower limb EMG, with the authors hypothesising that the neuromuscular capacity to produce force may be a determinant of distance running success (Nummela et al., 2006). The influence of wearing an ‘unstable’ shoe, whereby the soles are rounded in the anterior-posterior direction, on muscular activity and $\dot{V}O_2$ has provided some interesting results. Recently, Koyama and colleagues (2012) reported a 3-5% increase in walking $\dot{V}O_2$ when wearing unstable shoes compared to traditional trainers. They attributed the higher metabolic demand to the increase in calf muscular activity and longer step lengths. It is also likely that the 1% increase in shoe mass affected $\dot{V}O_2$ and muscle activity (Koyama, et al., 2012), a factor that must be considered when assessing metabolic demand during running in different footwear.

In contrast to these promising results Saunders et al. (2000) did not find a significant relationship between significant increases in $\dot{V}O_2$ and VL muscle activity during constant-rate high-intensity cycling. The authors posited that small effect sizes meant there were inadequate statistical powers to identify relationships. Yet, similar findings support Saunders et al. (2000) showing that muscular recruitment patterns remain unchanged during the $\dot{V}O_2$ slow component exhibited during heavy-exercise (Scheuermann, et al., 2001). Furthermore, recent research observed neuromuscular alterations during a prolonged run whilst wearing orthoses compared to running without, however RE remained unchanged between the conditions (Kelly, Girard, & Racinais,
The researchers speculated that the neuromuscular changes might be too small to alter the metabolic demand (Kelly, et al., 2011). Conversely, it has been argued that increases in muscular activity may not actually be a factor in the increase in $\dot{V}O_2$ (Gonzales & Scheuermann, 2008). This argument was based on the fact that although both $\dot{V}O_2$ and muscle activity increased during a moderate cycling exercise, after a bout of fatiguing heavy exercise, the two variables were not correlated.

### 3.2 Muscular Coactivation and Oxygen Consumption

Muscular coactivation, or cocontraction, primarily concerns the simultaneous contraction of the antagonist and agonist muscles, however, sometimes research considers the simultaneous contraction of two muscles performing similar actions i.e. extension of the lower limb. The movement pattern determines whether coactivation is detrimental or in fact beneficial to economy. For movement patterns such as cycling, a low percentage of coactivation of the rectus and biceps femoris (RF and BF, respectively) is believed to be economical and a requisite of a good technique (Candotti et al., 2009), as the simultaneous contraction fails to produce a net movement (Winter, 2009). On the other hand, during running coactivation may stabilise a joint, such as the ankle (Winter, 2009) by increasing the joint stiffness (Enoka, 2008; Humphrey & Reed, 1983), stabilising the ankle joint (Falconer & Winter, 1985) and making the runner’s storage of elastic energy more efficient (Heise, et al., 2008). That said, only two studies (by the same group) have investigated muscular coactivation in adults whilst running (Heise, Morgan, Hough, & Craib, 1996; Heise, et al., 2008) and they disagree with several studies that have considered walking economy (Frost, Dowling, Dyson, & Bar-Or, 1997; Hortobagy, Finch, Solnik, Rider, & DeVita, 2011; Mian, Thom, Ardigo, Narici, & Minetti, 2006; Peterson & Martin, 2010) and RE in children (Frost, et al., 1997).

Heise and colleagues (1996; 2008) observed a negative correlation between the coactivation of the biarticular leg muscles RF and gastrocnemius (both involved in extending the lower leg) and RE in male and female runners, when running at a self-selected speed. This coactivation across multiple joints had a
stronger relationship with RE than a single muscle, suggesting such muscular coordination is more metabolically beneficial (Heise, et al., 2008). The importance of muscular coactivation supports earlier work demonstrating trends between RE and the coactivation of biarticular muscles (RF, medial hamstrings, lateral hamstrings, and gastrocnemius) in well-trained runners. Yet, the correlations were not significant across any of the muscle pairs (Heise, et al., 1996). Conversely, Frost and colleagues (1997) stated they had ‘little doubt’ (p.186) that the greater coactivation found in younger children was associated with the higher \( \dot{V}O_2 \) during running.

Strong evidence from walking investigations seems to support this argument that an increase in muscular coactivation will increase the metabolic cost of locomotion. Such research has primarily considered different age groups of either adults (Hortobagyi, et al., 2011; Mian, et al., 2006; Peterson & Martin, 2010) or children (Frost, et al., 1997). There is a general consensus that higher lower limb coactivations, usually of the thigh, are observed in the most uneconomical age group either young children (Frost, et al., 1997) or elderly adults (Hortobagyi, et al., 2011; Marques et al., 2013; Mian, et al., 2006; Peterson & Martin, 2010). Frost and colleagues (1997) attributed this to energy being wasted when unnecessary muscular activations are performed. However any measurement of coactivation and metabolic cost will assess both necessary and unnecessary activations and their total effect on \( \dot{V}O_2 \). It is likely that coactivation does provide some protection to joint ligaments, such as prevention of tibial anterior displacement during knee extension via thigh coactivation that protects the anterior cruciate ligament (Draganich, Jaeger, & Kralj, 1989). This may be an injury prevention mechanism that could potentially increase \( \dot{V}O_2 \). Other such mechanisms that may occur as a result of coactivation are a greater, more equally distributed pressure across the articular surfaces, regulating mechanical joint impedance and the maintenance of joint stability (Baratta et al., 1988). It is thought that such dynamic protection of the joints by the muscles may be of greater importance for children and the elderly, possibly due to immature and/or weak muscular systems respectively (Frost, et al., 1997; Mian, et al., 2006). Additionally, it may represent an effort to sacrifice economy for safety and stability via neuromuscular adaptations (Frost, et al.,
It has been hypothesised that there is an increase in the metabolic cost of walking with an increase in antagonist coactivation because the agonist needs to produce greater force to oppose the antagonist force. This requires more muscle fibres to be recruited and thus, a greater metabolic demand (Hortobagyi, et al., 2011; Mian, et al., 2006). It is possible that such a hypothesis can be extended to the relationship between the metabolic cost of running and muscular coactivation.

The discrepancies between the walking and running investigations from a physiological point of view seem illogical, given the compelling evidence that greater coactivation is associated with a greater metabolic cost (or reduced RE). Conversely, the more biomechanical argument, of greater stability/stiffness and efficiency of the SSC meaning greater coactivation is associated with a lower metabolic cost (or improved RE) could have merit, as coactivation could play a crucial role during running. Critical examination of the methodologies used shows differences in how coactivation was calculated and the speed used for testing. In regards to this latter point, all investigations into walking and children running used a standardised test speed, whilst the adult running investigation allowed participants to self-select their speed based on ratings of perceived exertion (RPE) (Heise, et al., 2008). The magnitude of lower limb muscular activity is known to be speed dependent, as are the EMG temporal profiles (Gazendam & Hof, 2007; Higashihara, Ono, Kubota, Okuwaki, & Fukubayashi, 2010; Kyrolainen, Avela, & Komi, 2005; Silder & Thelen, 2010), with differences being reported between slow jogging (2.5 m·s\(^{-1}\)) and running (3-4.5 m·s\(^{-1}\)), perhaps due to changes in neuromuscular requirements and effort levels (Kuitunen, Komi, & Kyrolainen, 2002; Stirling, von Tscharner, Kugler, & Nigg, 2011). If the muscular activity profiles, both in terms of amplitude and timing are affected by speed, then it is likely that muscular coactivation will vary as a result. Such was the case in the youngest children in the investigation conducted by Frost and colleagues (1997), whereby the highest coactivation was recorded at the fastest speed. Thus, further investigations are needed to understand the influence of muscular coactivation on the metabolic cost of
running at standardised speeds to examine whether this influences the relationship found.

3.3 Relationship between Muscular Activity, Kinematics and Oxygen Consumption

Kellis, Zafeiridis and Amiridis (2011) examined the role of muscular coactivation before and after a fatiguing protocol and observed changes in both the quadriceps-hamstrings and gastrocnemius–TA ratios during the impact phase of running. According to the authors these adjustments helped preserve lower limb stability, meaning further muscular changes (after the first 50 ms of ground contact) were not necessary, as stabilisation had already been achieved. It is also likely that the changes reported by Kellis and colleagues (2011) to the preactivation of the vastus medialis and BF contributed to an observed greater knee flexion at TD. This kinematic change has consistently been found in fatigue studies (Derrick, Dereu, & McLean, 2002; Mizrahi, Verbitsky, Isakov, & Daily, 2000b) and was suggested as resulting in an increased metabolic cost (Derrick, et al., 2002). This highlights the need to interpret the mechanics of running in light of their effects and affecters. Whilst $\dot{V}O_2$ was not measured in this study, being in a fatigued state implies $\dot{V}O_2$ was greater than pre-fatigue. Therefore the preparatory movements and impact phase could be crucial to RE as the muscles act to preserve the initial stability of the lower limb, by maintaining $K_{leg}$ and consequently, could be a feature of economical running. Theoretically, it is possible that the magnitudes of muscular activation may influence the kinematics, which in turn affects RE, and the muscular coactivation ratios affect RE through enhanced or reduced $K_{leg}$. In principle this is plausible; however, there is still some debate as to whether muscular activity alterations affect either the kinematics of running or the economy of the movement pattern or both.

Hausswirth, Brisswalter, Vallier, Smith and Lepers (2000) observed an increase in the level of VL activity over the course of a prolonged run, yet this was not associated with any significant change in $\dot{V}O_2$. Furthermore, Bonacci et al.
(2010) found muscular activity levels to change in 53% (8 out of 15) of their triathletes when comparing a bout of running to a bout of cycling followed by running. Of these eight, there were only five who showed significant changes in $\dot{V}O_2$. Additionally, seven of the triathletes also exhibited kinematic changes, suggesting that muscular activation strategies are connected to movement patterns. In contrast to this apparent close relationship presented by Bonacci and colleagues (2010) between muscle activation and kinematics, are the findings by Chapman, Vicenzino, Blanch, Dowlan and Hodges (2008). They demonstrated, using similar conditions as Bonacci et al. (2010), that kinematics were similar between run and cycle-run conditions for each participant. However, EMG results of the TA showed 5 of the 14 triathletes altered their muscle recruitment patterns in the cycle-run condition. Additionally, evidence has shown that the relationship between walking kinematics and muscular coactivation in healthy, young individuals is weak (Arias, Espinosa, Robles-Garcia, Cao, & Cudeiro, 2012) and changes in muscular activity and $\dot{V}O_2$ due to different mechanical characteristics of the heel of a shoe result in subject specific alterations (Nigg, Stefanyshyn, Cole, Stergiou, & Miller, 2003). Together these results present a mixed picture, implying that neuromuscular alterations need to be considered on an individual basis rather than as a collective group (Bonacci, et al., 2010; Nigg, et al., 2003). Individually profiling a person’s movement pattern may help in identifying kinematic, kinetic and muscular activation combinations that are economical and uneconomical. However a general consensus is required first before individual assessments can be made. By purely considering kinematics and their contribution to RE, muscular adaptations can be missed and hence the biomechanical and physiological involvement in RE could be overlooked or underestimated.

3.4 Summary

The direct link between muscular activation and RE seems to be hard to determine, possibly due to the influence of running kinematics and kinetics. Without considering both running mechanics and activation of separate muscles the relationship between running gait and RE may remain elusive. Furthermore, conclusive evidence regarding the relationship between muscular coactivation
and the metabolic cost of running is needed as currently there are conflicting results in terms of the association between these two variables.

### 3.5 Aims II

The overall aim of this part of the thesis is to assess levels of muscular coactivation whilst running. In particular the specific aims are:

1) To establish the relationship between RE and muscular coactivation.

2) To identify whether there is a running speed effect on muscular coactivation.

### 3.6 Hypotheses II

This part of the thesis addresses the following hypotheses:

1) That muscular coactivation and RE are positively related to one another, meaning greater coactivation is associated with a higher oxygen cost (i.e. reduced economy) of running.

2) That greater coactivation occurs at higher running speeds.
Chapter 4 Review of the Literature III

The final part of the thesis focused on a manipulation investigation into BFT, minimalist shod (MS) and shod (SH) running. Applying the exciting findings from experimental chapters 6 and 7 to this topical area could provide new insight regarding the BFT running debate, particularly given the lack of research regarding the relationship between muscular activity and RE during BFT running (see 4.4 Muscular Activity).

4.1 Background regarding the interest in barefoot running

Whilst BFT running has often been used by researchers as a control condition during footwear investigations, there has recently been a wave of academic interest investigating the possibly beneficial effects of being BFT on performance and injury prevention. One of the main instigating factors of this current rise in BFT studies stems from the notion that humans were 'born to run [BFT]'. Evolutionary biologists believe that one of most crucial aspects of natural selection was based around being successful hunters (Lieberman, 2012). Therefore we naturally evolved to be economical BFT runners, with the most successful hunters being those who could persistently hunt prey over long periods of time. Empirical evidence suggests that individuals who are habitual BFT runners often forefoot strike (Lieberman, et al., 2010), and thus several biomechanical and physiological investigations have been conducted to analyse firstly, the gait characteristics of BFT running and, secondly, whether these result in an improved RE.

In line with the evolution theory, but with a greater focus on injury prevention comes the work from Robbins and colleagues in the late 80s (Robbins, Gouw, & Hanna, 1989; Robbins & Hanna, 1987; Robbins, Hanna, & Gouw, 1988). Their arguments were based on somewhat questionable anecdotal evidence that reported lower rates of lower extremity, over-use injuries in BFT populations (Robbins & Hanna, 1987). However their rationale had a strong scientific framework. They hypothesised that BFT running would lead runners to utilising the foot structure to dampen impact force. As the plantar surface of the foot, in terms of receptors, is similar to the palmer surface of the hand, it can sense
surface characteristics extremely well (Robbins, et al., 1989). Their first work demonstrated that after prolonged exposure to running and walking BFT the medial longitudinal arch span shortened in the majority of subjects (Robbins & Hanna, 1987). This suggests that the arch could deform to a greater degree during the shock absorption phase of running. Such adaptations are believed to be possible due to the heightened proprioception (high level of somatosensory feedback). Shoes on the other hand do not allow runners to mechanically adjust to attenuate the impact shock, as they create a rigid foot structure throughout stance and have a protective, often cushioning layer, between the plantar surface and the ground. Following on from this work, Robbins et al. (1988) looked to quantify surface avoidance behaviour by recording the load placed through the foot when loads were applied to the knee in a flexed position. This experimental set-up simulated the foot impact during initial ground contact whilst running. Findings showed that avoidance behaviour was similar between SH and BFT conditions (Robbins, et al., 1988). If a protective layer, such as that of a trainer, is extremely detrimental to motor control feedback that moderates impact load than the SH condition would be expected to a have a much lower avoidance behaviour.

A recent American study of 785 runners investigated the growing interest from the running community in BFT running and reported that 75.5% of runners were at least somewhat interested in running BFT or MS (Rothschild, 2012). There was also a low, but significant correlation between level of interest and (self-perceived) running level (r = .079), suggesting that self-described elite runners were more interested in MS or BFT. Furthermore, 20.8% expected running BFT to help improve their performance and 34.3% expected it to prevent injury. The study also reported that approximately 85% of the runners surveyed would continue with or attempt MS or BFT given they received sufficient instructions. However, empirical evidence into not just how to run BFT but also the benefits of such running is, thus far, inconclusive (Jenkins & Cauthon, 2011; Nigg & Enders, 2013).

Therefore whilst BFT running is not a new concept, it is growing in popularity within the running community. As with any mode of running, to fully understand an individual’s BFT running gait both performance and injury implications must
be considered. Consequently, whilst the literature review that follows will focus on BFT running and factors which influence RE, given the applied nature of such work the effect that such gait adjustments may have on injury risk will also be given consideration.

4.2 Performance benefits

Improving an individual’s RE by 1% can allow the runner to increase their speed per unit oxygen cost by 0.049 m·s⁻¹ (Hanson, Berg, Deka, Meendering, & Ryan, 2011), potentially having huge significance in distance running events. The relationship between RE and MS and BFT, however, is not clear-cut. Some researchers report that SH running has a greater metabolic cost than MS and BFT (Hanson, et al., 2011; Lussiana, Fabre, Hebert-Losier, & Mourot, 2013; Perl, et al., 2012; Squadrone & Gallozzi, 2009; Warne & Warrington, 2012). Yet a number of studies refute this claim (Burkett, Kohrt, & Buchbinder, 1985; Divert et al., 2008; Franz, Wierzbinski, & Kram, 2012). Recently however, Nigg and Enders (2013) argued that asking whether SH or BFT/MS is better for performance is the wrong question. Rather running in a way that feels comfortable, probably when the energy demand is lowest, is the best way to run. That said there are several factors that many researchers agree influence the RE of an individual when running BFT, such as added mass, stride length and strike pattern.

4.2.1 Added mass

It has been argued that the added mass of the shoes could cause the metabolic cost of running to be greater in SH rather than MS or BFT (Divert, et al., 2008; Franz, et al., 2012; Frederick, Clarke, Larsen, & Cooper, 1983). According to Frederick et al. (1984) for every 100g added to the foot, there is a 1% increase in \( \dot{V}O_2 \). Therefore investigations into MS and BFT need to control for the added mass of the shoe (Franz, et al., 2012), which Hanson et al. (2011) and Lussiana, Fabre, Hebert-Losier and Mourot (2013) failed to do. Their overall RE improvement of 3.8% and 1.3% respectively, could potentially be accountable for by the mass of the shoes. Importantly, Divert et al. (2008) made a distinction
between the effect of shoes and the effect of mass on RE. It appears that it is due to the added mass, as the effect on economy was similar for SH as it was for socks loaded with a mass equivalent to the mass of the shoe. That said, Perl et al. (2012) reported running in MS (Vibram FiveFingers) to be 2-3% more economical than SH running, after controlling for shoe mass. Contrastingly however, Franz et al. (2012) demonstrated that RE was better (i.e. lower $\dot{V}O_2$) in minimalist footwear (Nike, Mayfly) than for BFT with added mass (equating to the mass of the shoe) strapped to the top of the foot. The relevance of basing performance implications of BFT/MS running on investigations whereby extra mass has been strapped to the foot is however highly questionable. Such studies seemingly ignore the effect that such added mass would have on running mechanics. The leg moment of inertia ($I_{\text{leg}}$) is lower when BFT than when SH, therefore realistic BFT running may not be performed. Also when adding 0.5 kg to the foot individuals increase their stride length (Martin, 1985), therefore running mechanics may be more closely matched to those performed during SH running rather than those performed naturally during BFT running when individuals generally shorten their stride length (see 4.2.3 Stride length).

4.2.2 Cushioning

When making direct comparisons between the study conducted by Franz et al. (2012) and the one conducted by Perl et al. (2012) the difference in MS needs to be considered. The former research group used a cushioned shoe, whereas the latter used a shoe without cushioning. This is an important distinction that needs to be highlighted, as soft-soled shoes can decrease the metabolic cost of running by 1-2% compared to hard-soled shoes (Frederick, et al., 1983). To the author’s knowledge only one study to-date has compared RE during BFT, MS (with no cushioning) and SH running (Squadrone & Gallozzi, 2009). Their findings suggest that MS running has the greatest metabolic advantage (i.e. a better RE than SH, yet similar to BFT). Yet, both cushioning (SH) and a thin external layer (MS) insulate the foot against sensory feedback. This feedback governs the impact experienced during foot contact with the ground (Robbins, et al., 1989; Robbins & Hanna, 1987). It is possible therefore that when BFT, runners may adopt a different running style to attenuate the mechanical
stresses placed upon the feet due to the lack of an external/cushioning layer. Whilst advocates of BFT believe this sensory feedback is useful during running to attenuate impact forces, they neglect to address the possibility that there may be a metabolic cost associated with cushioning the body during BFT running (which may also be apparent during MS running). This “cost of cushioning” hypothesis was first proposed by Frederick et al. (1983) and later supported by Franz et al. (2012), who found lower metabolic power when running in shoes with increasing mass than compared to BFT with similar mass. Whilst Divert et al. (2008) found BFT running was more mechanically efficient than SH as greater work was done for the same metabolic cost. So rather than a direct detrimental impact of cushioning the body on metabolic cost, there is a greater mechanical cost meaning efficiency is enhanced.

If the cushioning properties of shoes can affect RE, a case can be made that there may be an optimal surface, with appropriate properties, which elicits a metabolic advantage by decreasing $\dot{V}O_2$. Pinnington and Dawson (2001a, 2001b) measured the $\dot{V}O_2$ of recreational runners and elite iron-men and found that running BFT on sand (iron-men: 47.6 and recreational runners: 43.3 mL·kg$^{-1}$·min$^{-1}$) had a higher metabolic cost than running BFT on grass (iron-men: 32.2 and recreational runners: 32.5 mL·kg$^{-1}$·min$^{-1}$). This suggests that a firm surface, rather than a compliant one, may provide BFT runners with the biggest metabolic advantage. Additionally, increasing the compliancy of a surface may have a direct detrimental effect upon the functioning of the foot when BFT. It is estimated that the arch of the foot can recover 17% of the mechanical energy generated per step (Kerr, Bennett, Bibby, Kester, & Alexander, 1987), yet this may only be possible whilst running on a stiff surface, allowing the foot to act like a natural spring. It must be acknowledged that Kerr and colleagues (2009) did not report the strike pattern that this estimation was based on, which is likely to influence the total mechanical energy recovered in the foot arch. However there is evidence that a more compliant surface reduces the metabolic cost of running due to the elastic rebound of the surface (Kerdok, et al., 2002). This may in fact lead to a more efficient running form than the release of stored elastic energy in the foot structures. Consequently, Kerdok et
al. (2002) argue that there is a material stiffness that elicits the lowest metabolic cost.

Regardless of the surface stiffness, BFT runners may use sensory feedback to modify their gait. Researchers have argued that this modification may only be necessary during multiple steps i.e. treadmill running for a length of time, rather than a limited number of steps i.e. laboratory running over a force plate, as runners may be able to sustain higher impacts when only running for a few steps (Divert, Baur, Mornieux, Mayer, & Belli, 2005a). This means that the running gait analysed may be unnatural during a test period of insufficient time. However the time taken to become accommodated to BFT running on a treadmill has not been investigated. Evidence suggests a minimum of 6-9 minutes is needed for participants to familiarise themselves with treadmill running (Lavcanska, Taylor, & Schache, 2005), yet the ability to run consistently while BFT on a treadmill, especially for habitually shod runners, requires examination particularly if investigations include experimental manipulations. It is possible that if only one adjustment is being made, such as only adjusting to a treadmill belt or only adjusting to BFT running, then similar familiarisation times may be recorded. Therefore future investigations into BFT treadmill familiarisation may look to include participants that are already familiar with treadmill running, to minimise the confounding variables.

4.2.3 Stride length

Franz et al. (2012) hypothesised that stride length may play a role in determining the metabolic cost of running BFT or in shoes. They found that runners decreased their stride length by 3.3% during BFT running compared to during SH. It was apparent from their research that this was an effect of the removal of shoes rather than running with added mass. Further to this, SH running was found to be 3-4% more economical than BFT running. However, when controlling for stride frequency, and thus stride length, Perl et al. (2012) reported that MS running was more economical than SH running. This is further supported by Squadrone and Gallozzi (2009), who found runners exhibited similar stride lengths during MS and SH running, but MS running was more economical than SH. They also reported that both of these conditions had
longer stride lengths than BFT running, but no differences in RE were found. It has commonly been reported that BFT running results in shorter stride lengths and higher stride frequencies (Burkett, et al., 1985; Divert, et al., 2008; Franz, et al., 2012; Squadrone & Gallozzi, 2009). It is possible that such a gait adjustment is based on several factors, but the influence the shoe mass or added mass to feet has on stride length (see 4.2.1 Added Mass) is a limitation disregarded by previous researchers (Franz, et al., 2012; Perl, et al., 2012). They added lead weights to the feet of runners, raising the question of whether MS running would actually be more economical if lead weights were not added. Additionally, Squadrone and Gallozzi (2009) did not adjust mathematically for the difference in shoe mass suggesting that the BFT, and to a lesser degree MS, $\ddot{V}O_2$ is not comparable to the SH $\ddot{V}O_2$. Therefore the effects of the shorter strides adopted during BFT running on $\ddot{V}O_2$ have not been empirically examined. To-date no investigation has considered mathematically adjusting absolute $\ddot{V}O_2$ values to account for the difference in shoe mass rather than adding mass. Further to this, no experimental manipulation has examined whether strategically changing stride length during different footwear conditions has any affect upon RE.

With regards to the findings of Squadrone and Gallozzi (2009), it seems possible that the small layer between the foot and ground in the MS condition dampens the sensory feedback received by the runner. This enables runners to keep their running stride similar to that exhibited during the SH condition. Shorter stride lengths adopted during BFT running may be a mechanism to reduce impact magnitudes (Mercer, Devita, Derrick, & Bates, 2003b) and thus, may protect individuals from possible bony injuries. However recent evidence suggests that a 10% increase in stride frequency (i.e. decrease in stride length) does not actually reduce $F_{\text{impact}}$ (Giandolini et al., 2013a). Therefore the 3% shortening of stride lengths to lower the amount of force experienced may also have to be accompanied by a change in foot strike pattern to successfully lower the magnitude of $F_{\text{impact}}$. 
4.2.4 Strike pattern

A rearfoot strike pattern is characterised by the heel being the first part of the foot to make contact with the ground, whereas a forefoot strike is, often, characterised in BFT studies as the forefoot striking the ground first followed by the heel (Perl, et al., 2012). Another strike pattern is midfoot, whereby the heel and forefoot simultaneously make contact with the ground. Whilst Perl et al. (2012) hypothesised that the strike pattern when BFT may be crucial to determining whether BFT is more economical than SH running, their results, along with Gruber, Umberger, Braun and Hamill’s (2013), did not show a difference in RE between rearfoot and forefoot striking. The belief is that forefoot striking increases shock attenuation, as the plantarflexor muscles contract eccentrically absorbing the energy of the low frequencies of impact. High frequencies on the other hand, demonstrated by the appearance of $F_{\text{impact}}$, are attenuated by passive mechanisms, such as bones, heel fat pat, ligaments and cartilage. This elastic energy produced through the eccentric contraction can be stored and then released during propulsion, making forefoot striking more efficient than rearfoot striking. However the evidence does not seem to support this. Forefoot striking would cause the foot to be in a plantarflexed position upon initial contact with the ground. To prevent the heel from contacting the ground at TD ankle stiffness would increase (Hamill, Russell, Gruber, & Miller, 2011). Greater lower extremity stiffness has been associated with poorer RE (Kerdok, et al., 2002), possibly due to greater muscular coactivation (Chapter 7).

The majority of SH runners rearfoot strike (Hasegawa, et al., 2007) and it has been hypothesised that this is a result of having a heel lift, because habitual BFT runners use a forefoot strike (Lieberman, et al., 2010). Recently, however, this assumption has been contradicted. Results have shown that the majority of habitual BFT runners use a rearfoot strike during endurance running speeds and switch to midfoot/forefoot strikes with increasing speeds (Hatala, Dingwall, Wunderlich, & Richmond, 2013). Additionally, Hamill and colleagues (2011) demonstrated that a forefoot strike pattern is more likely to be an effect of being unshod, as increasing an individual’s heel lift by 4 mm had no effect upon their foot strike pattern. Furthermore, it now appears that it is the surface
characteristics that hugely influence foot strike when BFT (Gruber, et al., 2013). Gruber and colleagues (2013) found 20% of their participants used a forefoot strike pattern when on a soft surface, but on a hard surface this increased to 65%. It was likely that participants in this study changed foot strike patterns to moderate impact forces (Robbins, et al., 1989) and pain levels. Therefore it was also possible that stride length was altered, due to its ability to attenuate shock (Mercer, et al., 2003b), however this was not measured.

Whilst it appears that there may not be a performance benefit to adopting a different strike pattern during BFT it is possible that this gait alteration could influence the likelihood of injury.

**4.3 Injury prevention**

If there are performance implications of BFT running, even if they are only small, many runners will pursue this mode of running. Yet performance improvements are only worthwhile if runners can stay injury free. Therefore particular gait changes need to be considered in terms of the relationship to injury risk.

**4.3.1 Strike pattern**

Generally, runners will adopt either a midfoot or forefoot strike pattern when running BFT, even if they are rearfoot strikers during SH running (Hamill, et al., 2011; Lieberman, et al., 2010). Lieberman et al. (2010) believe that this is a protective mechanism, as it may potentially reduce the incidence of bony injuries. However empirical evidence supporting this statement is lacking, with the exception of a retrospective study by the same group which suggested forefoot strikers had a lower incidence of repetitive stress injuries (Daoud, et al., 2012). Additionally, Hamill and colleagues (2011) observed greater ankle stiffness during stance and a midfoot strike with a plantarflexed ankle at TD when comparing BFT to SH running. The change from dorsiflexion to plantarflexion at TD occurs through greater preactivation of the GL (Giandolini, et al., 2013a) and produces a greater ankle plantarflexor moment and negative power in early stance (Paquette, Zhang, & Baumgartner, 2013). Although this
could translate to greater stability at the ankle, it may also place strain on the plantarflexor muscles to produce greater amounts of eccentric activity (Paquette, et al., 2013), producing greater loads through the Achilles tendon (Almonroeder, Willson, & Kernozek, 2013; Kulmala, Avela, Pasanen, & Parkkari, in print). Additionally, striking the ground in a more anterior position on the foot may increase the likelihood of certain bony injuries (Goss & Gross, 2012; Ridge et al., 2013). A recent study investigated the effect of a 10 week transition period to MS running on bone marrow edema, an indication of bone stress injury (Ridge, et al., 2013). It was evident that after this transition period the majority of runners in the MS group (10 out of 16) were classified as injured (i.e. had greater bone edema) compared to only 1 in the group that had only trained in traditional running trainers for 10 weeks. However it is possible that a structured transition programme, incorporating strength and stretching exercises for foot muscles, may help in safely transitioning to MS/BFT running. Whilst during the Ridge et al. (2013) study, only volume of time spent running in MS was increased, results have shown that tailoring training so that the number of ‘push-offs’ are increased, can strengthen the toe flexor muscles (Goldmann, Potthast, & Bruggemann, 2013), which may protect the bony structures of the foot.

4.3.2 Impact peaks and loading rates

Supporters of BFT running with a forefoot strike often do not consider the implications of greater ankle stiffness; rather they focus on the absence of a vertical impact force peak which they argue is the basis for the injury protective mechanism of BFT running (Lieberman, et al., 2010). Even adopting a midfoot strike pattern during BFT results in a lower impact force than SH rearfoot striking (Hamill, et al., 2011). Additionally, Squadrone and Gallozzi (2009) found SH running had the greatest foot angle (indicating a rearfoot strike) and greater $F_{z\text{impact}}$ than both BFT and MS running. However recent work analysing the frequency components of the vertical force show visible low frequency impacts during forefoot striking, rather than the high frequencies associated rearfoot striking (Gruber, Davis, & Hamill, 2011). The authors believe that this could lead
to a greater incidence of muscular injuries, as the muscle tissue absorbs these low frequencies.

The impact shock (or impact acceleration) that the lower limb is exposed to during running, often measured as tibial acceleration (McNair & Marshall, 1994; Mercer, Bates, Dufek, & Hreljac, 2003a; Mercer, Vance, Hreljac, & Hamill, 2002), has been reported to be higher during BFT running than SH (McNair & Marshall, 1994; Sinclair, Greenhalgh, Brooks, Edmundson, & Hobbs, 2013a). This is an important variable because it can provide a direct estimate of the tibial load during impact (Milner, Ferber, Pollard, Hamill, & Davis, 2006), whereas impact force represents the net force as a result of the acceleration of the whole body's centre of mass acting on the ground at impact. Furthermore it has also been associated with various overuse injuries in runners (Milner, et al., 2006; Zifchock, Davis, Higginson, McCaw, & Royer, 2008). There were several possible mechanisms behind this increase in impact acceleration, with external and internal factors being influential. Firstly, the removal of an external cushioning layer that acts as a shock absorber, i.e. a typical running shoe, would mechanically increase the impact acceleration experienced. Secondly, BFT running is characterised by greater knee flexion at TD (De Wit, De Clercq, & Aerts, 2000) and a flatter, more anterior strike of the foot on the ground (De Wit, et al., 2000; Lieberman, et al., 2010; Paquette, et al., 2013). Both these characteristics lower the effective mass (m_{eff}) of the lower limb (Derrick, 2004; Lieberman, et al., 2010) and if the m_{eff} dominates this relationship (F = ma), then impact acceleration upon TD would increase as a result; the m_{eff} being the portion of the whole body’s mass that is being accelerated (Derrick, et al., 2002). Furthermore the removal of a shoe mass would also decrease the m_{eff}, even without any changes in gait. The effect that this would have on RE however has not been previously investigated.

The causal link between GRF variables and injury is debatable, but loading rates rather than impact force seem to have the greater support (Milner, et al., 2006). The high, rapid rate of loading the lower limb is a characteristic of both BFT and MS running rather than SH running (De Wit, et al., 2000; Paquette, et al., 2013; Sinclair, et al., 2013a). However this is only discernible when a distinguished impact peak is present. When this is not the case, such as when
forefoot striking, Lieberman and colleagues (2010) estimated SH rearfoot striking to exhibit similar loading rates to BFT forefoot striking. Nevertheless the lack of a distinguishable impact peak hampers the accuracy of impact force and/or loading rates calculations. Therefore conclusive statements regarding injury, based on these kinetics, are unlikely. Consequently, a more suitable measurement during BFT running may be impact acceleration.

4.4 Muscular activity

Most researchers are in agreement that BFT/MS running affects running mechanics, in comparison to SH running (De Wit, et al., 2000) and also in comparison to each other (Bonacci et al., 2013). Yet the changes that occur at a muscular level have received little attention. In fact, many researchers instead tend to hypothesise about the likely consequences of kinematic and kinetic adjustments on muscular activity (Paquette, et al., 2013). Yet recent research suggests it should be given consideration. Miller et al. (2012) generated computer simulations to investigate whether runners prioritise minimising metabolic cost or try to minimise some other quantity. They concluded that the need to minimise the total muscular activity of the lower limb, in addition to the amount of oxygen consumed, is a crucial factor in economical running. This study highlights the importance of directly measuring the activation of lower limb muscles during running.

4.4.1 Activation and coactivation

As actual measurements of muscular activity are lacking in many BFT studies, inferences based on their results can be made about the level of muscular activation. For example, both Coyles, Lake and Lees (2001) and Hamill et al. (2011) found significantly higher ankle stiffness when running BFT. These changes may have been elicited by greater muscular coactivation surrounding the joint, leading to changes in the metabolic cost of running, which could be beneficial (Heise, et al., 2008) or detrimental (Chapter 7). Additionally, it has been suggested that the greater knee flexion during SH running requires greater contraction of the quadriceps to control knee flexion, thereby increasing
the metabolic cost of running (Perl, et al., 2012). However, this suggestion by the authors implies that kinematic changes occur in conjunction with predictable changes in muscle activity, however this is not necessarily the case (Bonacci, et al., 2010; Chapman, et al., 2008).

One study has directly assessed muscular activity changes in the TA during SH and BFT running (von Tscharner, Goepfert, & Nigg, 2003). Prior to impact there was less TA activity in the BFT condition. This was probably due to greater plantarflexion at TD when BFT running (De Wit, et al., 2000) as switching from a rearfoot strike to a midfoot strike also decreases TA activity (Giandolini, et al., 2013a). However during stance there was a higher level of activity when BFT compared to SH (von Tscharner, et al., 2003). The authors stated that the positioning of the foot prior to TD is a slow event, but a fast muscle movement is necessary to control the foot slapping the ground. They believed that the muscle activity prior to TD might be influential during fatigue and/or performance. Given the dearth of literature regarding muscle activity during BFT/MS running, both preactivity and stance periods require examination, especially if changes in cushioning (i.e. BFT, MS and SH) is investigated, as it will allow the cost of cushioning hypothesis to be investigated.

Although the argument of added mass may indeed be a factor in causing an increase in $\dot{V}O_2$ (Divert, et al., 2008; Franz, et al., 2012; Frederick, 1983), mechanisms behind why adding extra mass to a runner's foot has this metabolic effect have not been forthcoming. It could potentially be due to compensatory changes in muscular activity, such as greater contractions resulting in higher magnitudes or greater muscle fibre recruitment that may induce a higher metabolic cost than BFT without added mass. The muscular changes could be in response to increases in the moment of inertia of the leg/foot due to the added mass. However there is evidence that changes in shoe mass and midsole design have no effect on muscle activity or metabolic demand (Santo, Roper, Dufek, & Mercer, 2012). Consequently, little is known regarding why added mass increases $\dot{V}O_2$. 
4.4.2 Stretch-shortening efficiency changes

It is possible that the attenuated impact forces during BFT may be due to changes in lower extremity muscle pre-activity (Boyer & Nigg, 2004). These would affect the foot strike patterns, whereby greater activity of the plantarflexors (e.g. gastrocnemius) would promote a midfoot/forefoot strike (Giandolini, et al., 2013a). This pre-activity could increase the efficiency of the SSC, as it allows the plantarflexor muscles to store more elastic energy during eccentric contraction, and thus, release more during the propulsive, concentric contraction. Perl et al. (2012) suggested that it is the efficiency of the storage of elastic energy in the Achilles tendon, rather than the plantarflexor muscles, which is increased during BFT due to the smaller excursion of the knee. On the other hand, by increasing the stretch of the Achilles tendon, strain upon the tendon increases. The possibility of Achilles tendon and/or calf injuries could increase due to the greater demand placed upon the MTU. Furthermore, advocates of BFT running argue that trainers make running inefficient by interfering with the natural, spring-like, function of the foot due to stiff soles and arch supports (McMahon, 1987; Perl, et al., 2012).

4.5 Summary

Several investigations have tried to control possible confounding variables by adding extra mass to a BFT condition or instructing runners to run with certain stride lengths. Whilst providing strong internal validity, the practical implications of such examinations to realistic BFT running are questionable. Based on the previous literature it remains unclear whether there is indeed a metabolic cost to cushioning the lower limb when individuals run with no external cushioning layer. Moreover the influence of heightened proprioception separately to cushioning has not been adequately tested. The additional factor of changes in stride length and its effect upon RE is at present unknown. Thus there is a need to control the variation in the level of cushioning and proprioception, in addition to changing stride length, whilst critically investigating self-optimisation strategies through consideration of RE and running mechanics. In order to be able to determine whether modifications to running mechanics under these
conditions are due to each condition, participants need to be familiarised with both BFT and SH treadmill running. Studies have investigated SH treadmill familiarisation (Cavanagh & Williams, 1982; Lavcanska, et al., 2005; Schieb, 1986; White, Gilchrist, & Christina, 2002), but thus far none have analysed BFT treadmill familiarisation. Therefore prior to this experimental manipulation it is necessary to quantify the period of time required for individuals to produce a stable and consistent gait pattern whilst BFT running on a treadmill. Furthermore, such an investigation can highlight the specific gait adjustments produced when becoming familiar with BFT running.

4.6 Aims III

The overall aim of the final part of this thesis was to apply the knowledge obtained in the two previous experimental chapters (6 and 7), to the topical debate of BFT running. Specifically the aims were as follows:

1) To assess the time required to become familiarised to BFT treadmill running.

2) To investigate specific BFT gait adjustments that occur as a result of familiarisation.

3) To investigate the mechanisms behind changes in RE during different stride lengths when varying cushioning and proprioception by comparing SH, MS and BFT RE.

4.7 Hypotheses III

The following hypotheses look to address these aims:

1) Runners will be able to produce a consistent gait pattern within 10 minutes of running BFT on a treadmill.

2) BFT running with a BFT stride length will be the most economical condition.
3) Running with a SH stride length during MS running will be more economical than SH with a BFT stride length, but both will be more economical than SH running.

4) BFT and MS running will have the highest impact accelerations when compared to SH.
Chapter 5 General Methods

5.1 General Experimental Procedures

The four experimental chapters (Chapter 6, 7, 8 and 9) that comprise this thesis involved a total of 52 participants and 73 exercise testing sessions. All sessions were conducted in air-conditioned laboratories at sea level, with an ambient temperature of 18-22°C. Each experimental investigation was approved by the University Ethics Committee prior to the initiation of any testing.

5.1.2 Participants

Participation in all investigations was voluntary and participants were recruited from the student and staff University community, as well as the local community. Participants were free from disease and injury, and only non-smokers were recruited. Recreational runners were used in all but one experimental study, which was chapter 6. To be classed as a recreational runner individuals’ had to have been actively running for at least 6 months prior to testing, in addition to running at least twice a week. In chapter 6 beginner runners were used, this was defined as an individual having had no prior running training, in addition to not being currently involved in any recreational sport. In chapters 8 and 9 participants were only included if they had limited (less than 5 minutes) or no previous experience of running BFT. Prior to testing all subjects were instructed to avoid strenuous exercise (24 hours prior), alcohol (24 hours prior) and caffeine (6 hours prior). When it was necessary to re-test individuals, such as was the case in chapter 6 and 7, testing was conducted at the same time of day (± 2 hours). Treadmill familiarisation was given to all participants in chapters 6, 7, and 9, as well as multiple practice trials when over ground running was performed (Chapter 6).

5.1.2 Informed Consent

Prior to testing, all participants were given an information sheet which described the study’s protocol, what data would be recorded and what would be required from them. Additionally, any potential risks of participating were outlined and
participants were informed that they were free to withdraw at any point, with no disadvantage to themselves. It was also clarified that whilst data may be presented at conferences or in academic journals, their anonymity would be retained at all times. Furthermore, only the researchers involved in the study would have access to an individual’s data, which would be stored safely. Participants were free to ask any questions they may have about the study and only once they were happy they understood the protocol and wanted to proceed did they give their written consent to participate.

5.1.3 Health and Safety

As part of the ethics approval, specific health and safety guidelines had to be outlined and adhered to throughout each experimental protocol. Therefore sets of risk assessments were needed for each experimental study. All respiratory apparatus was cleaned with Virkon disinfectant and sterilised in a Milton (chlorine based) solution. Further to this, all surfaces in the physiology laboratories were cleaned with the Virkon disinfectant, as was the treadmill belt during the BFT running experimental procedures. During fingertip blood collection the investigator wore latex gloves and all sharps/biohazards were disposed of securely. Prior to all experimental testing procedures participants were allowed to perform their own warm-up. Following the testing they were provided with time to perform a cool-down. Participants were instructed to wear a comfortable t-shirt and shorts during biomechanical analysis and were encouraged to bring their own water bottle and drink water *ad libitum* during rest periods. Water was provided if participants came without a water bottle.

5.2 Measurement Procedures

5.2.1 Descriptive Data

Each participant’s stature (SEC-225, Seca, Hamburg, Germany) and mass (SEC-170, Seca, Hamburg, Germany) was recorded prior to experimental testing, to the nearest centimetre and gram respectively. Furthermore a questionnaire relating to the injury history of each participant was completed prior to all investigations and participants were excluded from testing if they had
suffered from a lower limb injury in the past six months. Preliminary screening of participants was performed to assess their cardiovascular risk (see 5.2.10 Preliminary Screening), in addition to participants completing a PAR-Q form. In several cases in chapter 6 participants were required to obtain consent from their doctor prior to participating in the testing protocol.

5.2.2 Treadmill running

All treadmill tests were performed on Woodway treadmills. The same model was used during chapters 6 and 7 (PPS 55 sport slat-belt treadmill; Woodway, Weilam Rhein, Germany). The display section is positioned in front of the treadmill belt. Therefore for chapter 6 it was necessary to conceal the display from the participant’s view, occluding distance, speed and time information. This treadmill has a maximum grade of 25%, which was required for the maximal stages of the graded-exercise test (GXT). Chapter 8 and 9 used a different model that had a smaller frame, (PPS 43med; Woodway, Wielam Rhein, Germany), as this did not occlude kinematic markers. The display is detached from the frame, allowing it to be positioned on the floor away from the participant. Whilst the speed of the treadmill cannot be specifically calibrated, it was recorded and compared to the speed displayed by the monitor. This was necessary during testing to check the standard error of estimate of the speed. During chapter 8 this was important due to the variation in self-selected speeds. Based on the standard error of estimate there was 95% confidence that the speed of the treadmill belt was within 1.7 m-min⁻¹ of the speed displayed on the monitor. The speeds used for each treadmill assessment are given in the respective chapters.

5.2.3 Over ground running

Over ground running was performed during gait analysis in chapter 6. The participants were asked to run along a 12m runway, ensuring one foot made contact with the force plate (960 Hz; Advanced Mechanical Technology, Inc., Watertown, MA) that sat flush with the ground and was situated half-way down the runway. Speed gates were positioned either side of the force plate to monitor running speed. A number of practice trials were given prior to
experimental testing to allow participants to familiarise themselves with correctly hitting the force plate. A trial was deemed successful if the following criteria had been met: 1) the whole foot had made contact with the force plate; 2) the participant executed the run at the experimental speed (2.53 m·s\(^{-1}\) ± 5%) and; 3) the participant did not adjust their running gait in an attempt to make contact with the force plate. Any trials that failed to satisfy all three criteria were discarded. Ten successful trials were recorded for both legs for each participant. Data collection of the left and right legs was performed separately, so step lengths for each leg could be determined.

5.2.4 Gait Analysis

During chapters 6, 8 and 9 a three-dimensional kinematic gait analysis was performed using an eight-camera motion capture system (Vicon Peak, 120 Hz, automatic, optoelectronic system; Peak Performance Technologies, Inc., Englewood, CO). This was positioned in an oval shape around the force plate (Chapter 6) or treadmill (Chapters 8 and 9) to allow the collection of kinematic data. The marker set-up used in each chapter was based on a modified model of Soutas-Little et al. (1987). Reflective spherical markers were positioned on the following anatomical positions (Figure 5.1): the proximal greater trochanter (hip), the medial and lateral condyles (knee), the mid-line of the posterior shank, the musculotendinous junction where the medial and lateral belly of the gastrocnemius meet the Achilles tendon, the mid-tibia below the belly of the TA, the lateral malleolus (ankle) (Chapters 6 and 8 only), the superior and inferior calcaneus, the proximal head of the third metatarsal, and the distal head of the fifth metatarsal (Chapter 6 only). Only the right leg was used for gait analysis in chapter 8 and 9, as reliable data whilst running on the treadmill could not be captured for the left leg. In chapter 6, bilateral kinematic data was recorded separately meaning the final 12\(^{th}\) marker was affixed to the inferior calcaneus of the opposite leg, which was not the primary focus of data collection at the time, to allow step length to be calculated.

The laboratory coordinate system used was such that the x-coordinate represents the medio-lateral axis; the y-coordinate lies perpendicular to the x-coordinate representing the antero-posterior axis. Thus, positive corresponded
to medial displacement (x-axis) and the direction of the run (y-axis). The z-coordinate represented the vertical axis, with positive indicating an upward motion. These coordinate axes were used in each analysis of gait. The kinematic data were synchronised with the kinetic data gathered from the force plate in chapter 6, with the threshold to activate the force plate set at 10 N. This was used as event detection. During chapter 8 and 9 the kinematic data were synchronised with the EMG and accelerometer data using a manual trigger as event detection.

Figure 5.1 Marker set-up for the right leg. a) Frontal plane from the posterior view. b) Sagittal plane from the lateral side. Hip = greater trochanter; Lateral and medial knee = lateral and medial condyles respectively; Achilles 1 = mid-line of posterior tibia; Achilles 2 = musculotendinous junction; Ankle = lateral malleolus; MTP5 = fifth metatarsal; Calc 1 and Calc 2 = superior and inferior calcaneus respectively; Shin = mid-tibia; Toe = third metatarsal.

Stride length, calculated in chapter 8 and 9, was defined as successive foot contacts of the right foot. To determine stride length a camera (Basler, 100 Hz)
was positioned approximately 1.5 m in front of the treadmill. Data were collected at the same time as the kinematic data. Equation 1 (Cavanagh & Williams, 1982) was used to compute the stride length over six gait cycles:

\[
SL = ST \times v_{\text{tread}} \tag{Eq. 1}
\]

SL = stride length (m); ST = stride time (s), measured as the time between successive foot contacts of the right foot and; \( v_{\text{tread}} \) = treadmill velocity (m·s\(^{-1}\)).

Gait analysis was performed in a standardised, neutral Adidas trainer (Dixon & McNally, 2008) in chapter 6 and 9. In chapter 9 this trainer represented the cushioned SH condition. Additionally, gait was analysed whilst wearing a minimalist footwear model (KSO, Vibram FiveFinger) and whilst BFT in chapter 9. During chapters 7 and 8 participants wore their own trainers for the assessment of their running gait.

5.2.5 Graded-exercise Test

To determine \( V\dot{O}_2\text{peak} \) in chapter 6, the Balke-Ware GXT was used (Balke & Ware, 1959). Participants were given at least 6 minutes to become familiarised to treadmill walking prior to the GXT (Matsas, Taylor, & McBurney, 2000). This also served as their warm-up. The GXT protocol was as follows: the grade was set to 0% for the first minute; it was then increased to 2% for the second minute. Following the completion of the first two minutes the treadmill was increased by 1% every minute until the participant reached volitional exhaustion. A constant speed of 90 m·min\(^{-1}\) (5.4 km·h\(^{-1}\)) was used throughout the test. The Balke-Ware GXT was deemed the most appropriate test due to the participant cohort, which were classed as sedentary adults with low fitness levels at the time of initial testing (Pollock et al., 1982).

5.2.6 Heart rate

During all exercise tests conducted on the treadmill heart rate (HR) was measured using a wireless chest strap, short-range telemetry system (Polar Electro T31, Kempele, Finland). Heart rate was manually recorded from a wristwatch strapped to the frame of the treadmill. During the GXT in chapter 6
HR was recorded every minute until completion of the test. In all submaximal, steady-state runs HR was recorded at the 4\textsuperscript{th}, 5\textsuperscript{th} and 6\textsuperscript{th} minute of each run.

5.2.7 Ratings of Perceived Exertion

To assess the participants’ perceived exertion Borg’s 6-20 RPE (Borg, 1998) was used in chapter 6 and 7. In both chapters participants were instructed on how to rate their exertion based on their perception of how hard their body was working. RPE was recorded during the 4\textsuperscript{th}, 5\textsuperscript{th} and 6\textsuperscript{th} minute of the submaximal runs and a mean score calculated and rounded to the nearest whole number. In chapter 6 RPE was also recorded every minute during the GXT.

5.2.8 Calf Muscle Range

In chapter 6 calf muscle flexibility was assessed following the procedures outlined by Bennell et al. (1999). The participants faced a wall in a step-stance position. Whilst holding onto the wall for balance they performed a forward lunge by bending their front knee towards the wall and keeping the posteriorly placed leg fully extended, with their heel flat to the ground. This stretched the calf muscle of their rear leg. An inclinometer was placed on the long axis of the Achilles tendon to measure the angle of the shank to the vertical. The measurement was taken three times, and participants were allowed to stand in a comfortable position in between each measurement. The calf muscle flexibility of both the left and right legs was recorded as the mean of the three measurements.

5.2.9 Sit-and-reach Test

The sit-and reach test was performed in chapter 6 to assess lower back and hamstring flexibility (Heyward, 2006). Participants were seated on the floor with their legs fully extended and flat to the ground. Their feet were flat to the sit-and-reach box. Participants were instructed to lean forwards as far as possible, pushing the bar on the sit-and-reach box. The test was performed three times and a mean score calculated.
5.2.10 Preliminary screening

Prior to experimental testing a cardiovascular risk assessment was undertaken. This involved measuring the brachial artery blood pressure, blood glucose level, cholesterol level and calculating the participant’s body mass index (BMI). Prior to the blood pressure reading, participants rested in a seated position for a minimum of five minutes. They remained seated whilst blood pressure was recorded using an automated sphygmomanometer (Dinamap Pro, GE Medical Systems, Tampa, USA). The mean of two measurements was computed as their resting blood pressure. The participant’s cholesterol and non-fasting blood glucose levels were determined from a fingertip blood sample. An alcohol swab was used to clean the fingertip before the skin was punctured with a disposable safety lancet (Safety-Lanzette, Sarstedt). To calculate BMI, the participant’s height (m) and body mass (kg) were used (mass/height$^2$).

5.2.11 Electromyography

Surface EMG of the right lower limb was recorded during submaximal running tests in chapter 7 and 9. By assessing gross neuromuscular activity, the muscle activity of each muscle during running can be inferred. A total of five muscles were examined in both chapters, these were the: RF, VL, BF, TA and GL. A standardised protocol was followed for the preparation of the skin prior to electrode placement. All participants were instructed to come to the laboratory with shaved legs, meaning a hairless surface was available for electrode placement. The area of each muscle belly was first cleaned using an abrasive gel and then an alcohol swab. The electrodes were placed longitudinally with respect to the muscle fibre direction following standardised criteria recommended by SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles project) (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). Each electrode was affixed with double-sided tape and the positions of the electrodes were outlined using a permanent pen. This allowed them to be precisely repositioned, particularly necessary in chapter 7 when testing was repeated four days later. The electrodes were secured to the lower limb with extra tape and self-adhesive bandages to minimise any unwanted movement. The EMG signal was recorded at 4000 Hz and 2000 Hz in chapter 7 and 9.
respectively, using a Trigno wireless system (Delsys, Boston, MA, USA; parallel bar configuration, contact material 99.9% Ag, interelectrode spacing 10 mm, electrode size 37 x 26 x 15 mm).

5.2.12 Accelerometer

Triaxial accelerometer data (Trigno Wireless EMG, Delsys, Boston, MA, USA) were recorded in all submaximal treadmill runs, except those performed during chapter 6. A surface electrode with an integrated triaxial accelerometer was placed on the heel of the right foot and the vertical accelerations ($A_z$) were used to establish TD and TO. This method was validated by simultaneously collecting GRF data and accelerometer data in a pilot study. From this and previous literature (Chapman et al., 2012; Sinclair, Hobbs, Protheroe, Edmundson, & Greenhalgh, 2013b) the portion of the $A_z$ graph that correlated with the stance period of running was identified (Figure 3.2). The accelerometer data were collected at 296 Hz (Chapter 7) and 148 Hz (Chapters 8 and 9). The $A_z$ of the heel marker was also used to determine impact acceleration in chapter 9, with this corresponding to the peak $A_z$ recorded during stance.

5.2.13 Pulmonary Gas Exchange

Pulmonary gas exchange and ventilation were measured breath-by-breath during all treadmill tests, except those performed in chapter 8. The same system was used to conduct all pulmonary gas exchange analysis (Cortex Metalyzer II; Cortex Biophysik, Leipzig, Germany). This metabolic cart comprised of a bidirectional “TripleV” digital volume transducer and differential paramagnetic (oxygen) and infrared absorption (carbon dioxide) analysers. Prior to each experimental testing session the gas analysers were calibrated with gases of known concentration and the volume sensor was calibrated using a 3-liter syringe (Hans Rudolph, Kansas City, MO). Each participant wore respiratory apparatus comprising of an oxygen mask, head cap, volume sensor and a capillary line that continuously sampled the gas. This determined $\dot{V}O_2$, carbon dioxide output ($\dot{V}CO_2$) and $\dot{V}E$. This was then displayed every 10 s after correction for the volume and concentration levels. After completion of each test...
the raw data (breath-by-breath gas exchange and ventilation) was exported to excel for further analysis.

![Figure 5.2 a) Sample impact acceleration data and b) vertical GRF indicating TD and TO portions of the data (adapted from Sinclair et al., 2013a).]

5.3 Testing procedures

5.3.1 Determination of $\dot{V}O_{2\text{peak}}$ and fractional $\dot{V}O_2$

The breath-by-breath $\dot{V}O_2$ data during the GXT in chapter 6 was averaged across a 30 s time period, with the highest value equating to the $\dot{V}O_{2\text{peak}}$. The fraction of $\dot{V}O_2$ utilised ($\%\dot{V}O_{2\text{peak}}$) during the sub-maximal runs were also calculated, by dividing the relative $\dot{V}O_2$ by the $\dot{V}O_{2\text{peak}}$. 
5.3.2 Defining running economy

The submaximal runs completed in chapter 6, 7 and 9 were performed for six minutes each. The $\dot{V}O_2$ measured in the final two minutes of each run was used to quantify RE, which was defined as the mean $\dot{V}O_2$ during this period. This gave participants four minutes to reach a steady-state (Morgan, Martin, & Krahenbuhl, 1989b; Roy & Stefanyshyn, 2006). The mean of the first six data points was statistically compared to the mean of the last six data points to assess whether they were significantly different. A steady-state was confirmed if the data was not significantly different. This was the case for each data set collected.

5.3.3 Defining Joint Angles

All investigations that included a gait analysis followed a set of basic assumptions. Firstly, the body is modelled as a series of rigid linked segments that have six degrees of freedom. These relate to three translations: medio-lateral, anterior-posterior and vertical; and three rotations in the following planes: sagittal, coronal and transverse. Secondly, marker placement is representative of underlying anatomical structures, from which joint centres can be estimated. The marker set-up (Figure 5.1) then allows the mathematical modelling of the lower limb segments. The thigh was defined as the vector between the hip and the lateral condyle, the shank was the vector between the mid-knee and mid-ankle and finally, the foot was the vector between the inferior calcaneus and the proximal head of the third metatarsal (toe marker). Based on these vectors segment joint angles can be defined. These are relative angles between segments that share a common joint.

The knee angle was calculated between the thigh and shank, with 180° equating to full knee extension and angles <180° indicating knee flexion (Figure 5.3). In chapter 8 and chapter 9, knee extension was classed as 0° and positive angles represent flexion. The ankle angle occurs between the shank and foot segments, with 0° representing the shank and foot at right angles to one another (Figure 5.3). Therefore dorsiflexion was shown by negative angles (-90-0°) and plantarflexion was shown by positive angles (0-90°). Rearfoot
movement was characterised by eversion and inversion. The rearfoot angle was calculated between a vector through the two calcaneus markers and a vector between the two markers on the posterior side of the shank (Figure 5.3). Negative angles represent eversion and positive angles represent inversion. Absolute angles were defined as a segment relative to a line in space. The hip angle was calculated between the thigh segment and a vertical line through the greater trochanter (hip marker) (Figure 5.3). Positive angles denoted the thigh in flexion (anterior to the vertical) and negative angles denoted the thigh in extension (posterior to the vertical). The final absolute angle was the foot angle defined between the foot vector and a horizontal line in the sagittal plane through the inferior calcaneus (Figure 5.3). A positive angle indicated that the toe marker was higher than the inferior calcaneus, meaning the foot was angled up, away from the ground. A negative angle indicated that the toe marker was lower than the inferior calcaneus, meaning the foot was angled down, towards the ground.

Figure 5.3 Graphical representation of joint angles employed. Sagittal plane view (left). Posterior frontal plane view (right).
5.3.4 Defining Ground Reaction Forces

The GRF is based on Newton’s third law of motion, that every action has and equal and opposite reaction. Therefore, during running, when an individual has either foot on the ground, the ground applies and equal and opposite force to the individual. Whilst the appearance of an initial impact (‘passive’) force experienced during running is dependent upon foot strike modality, there is always a ‘push-off’ propulsive (‘active’) force (Cavanagh & Lafortune, 1980; Lieberman, et al., 2010). All three orthogonal axes (x, y and z) contribute to the resultant GRF (GRFr) in varying degrees. The greatest proportion of the GRFr is $F_z$, followed by $F_y$ and then the $F_x$ (Cavanagh & Lafortune, 1980). It is commonplace to break the GRFr down into its three components and sometimes only $F_z$ is reported. However chapter 6 details both the GRFr and its separate components, as metabolically it is considered inadequate to only considered the $F_z$ component (Chang, et al., 2000; Chang & Kram, 1999).

Force (‘kinetic’) data produced from the force plate can also generate information on the centre of pressure. This is the theoretical point through which the GRFr is acting during each frame of data collection (Cavanagh & Lafortune, 1980). The force plate used in chapter 6 uses strain gauges to convert the GRF into electrical signals (Kirtley, 2006). The force applied to the plate changes the resistance of the strain gauges, which essentially change their resistance relative to the strain experienced.

5.4 Data Analysis Procedures

All data analysis, except calculation of physiological measures, was conducted using customised MATLAB (Math Works Inc., Cambridge, MA, USA) scripts created by the author, with the exception of the joint moments calculation, which was collaboratively written with a colleague (L.Damm).

5.4.1 Running economy

During chapter 6 three speeds were used to determine RE: 152, 138 and 152 m-min$^{-1}$ (7.5, 8.3 and 9.1 km-h$^{-1}$, respectively), the faster speed relating to the
speed used during the gait analysis. Given the cohort under investigation it was necessary to use slower speeds than those typically used to measure RE (Nummela, et al., 2007; Paavolainen, et al., 1999), but they were similar to those used previously on trained, female runners (Turner, et al., 2003). RE was the average of all three speeds, therefore the relative $\dot{V}O_2$ at each speed, expressed as the unit cost per minute was converted into the unit cost per distance (Jones, 2007).

$$\dot{V}O_2 \ (mL\cdot kg^{-1}\cdot km^{-1}) = \dot{V}O_2 \ (mL\cdot kg^{-1}\cdot min^{-1}) \times \left[ 60 / \text{speed} \ (km\cdot h^{-1}) \right] \quad (Eq. \ 2)$$

In chapter 6 RE was expressed in a similar manner, as three speeds were used: 152, 183, 200 m-min$^{-1}$ (9.1, 11 and 12 km-h$^{-1}$, respectively). However RE was expressed as the unit cost per minute during chapter 9. All absolute $\dot{V}O_2$ values were adjusted to the participant’s mass, with the exception of data gathered during chapter 9. Due to the difference in the mass of the footwear, known to affect RE (Divert, et al., 2008; Frederick, 1984; Jones, Toner, Daniels, & Knapik, 1984), each RE value in the BFT and MS conditions had to undergo additional adjustments to compensate for the decrease in mass compared to the SH condition, employing a similar methodology to Divert et al. (2008). In effect standardising a participant’s $\dot{V}O_2$ mathematically to compensate for the change in mass rather than adding weights to their feet, as previously done (Franz, et al., 2012). Furthermore, separate calculations were necessary to adjust the BFT condition to the difference in mass when compared to the MS conditions. The following adjustments were made:

1) BFT adjusted to body mass minus the mass of the trainer
2) BFT adjusted to body mass minus the mass of the minimalist footwear
3) MS adjusted to body mass minus the difference in mass between the trainer and minimalist footwear
4) MS adjusted to body mass
5) SH adjusted to body mass
5.4.2 Electromyographic Data

Prior to off-line analysis of all EMG data collected, the raw signal was amplified and band-pass filtered (20-500 Hz) within the Delsys hardware and recorded with a gain of 1000. Following this the raw data underwent full-wave rectification. Then several different procedures were performed on the rectified data depending on the variable of interest. In chapter 7 a temporal coactivation was required. This considers the common muscle on-time duration of two muscles, as a percentage of stance (Heise, et al., 2008). A linear envelope of the rectified EMG data was created using the Root Mean Square, with a 50 ms sliding window (EMG\textsubscript{RMS}). This represents the mean power of the signal. Peak EMG\textsubscript{RMS} during the stance phase of 20 consecutive steps was identified for each muscle and then the mean computed. Each muscles' EMG\textsubscript{RMS} was normalised to their respective mean peak EMG\textsubscript{RMS}. To determine muscle on-time the normalised data had to rise above specific thresholds for at least 50 ms. These thresholds were muscle specific. A baseline level was established by identifying the minimum muscle activity during a gait cycle. This was measured during five separate, but consecutive, gait cycles allowing a mean to be calculated. Then thresholds between 3 and 25\% of peak muscle activity (above baseline) were computed and compared to manually derived thresholds (Steele & Brown, 1999). Based on this procedure the following thresholds were chosen: RF, 7\%; VL, 7\%; BF, 20\%; TA, 12\%; GL, 7\%.

Using a relative peak rather than a maximal voluntary contraction is advocated during dynamic movements, such as submaximal running (Albertus-Kajee, Tucker, Derman, Lamberts, & Lambert, 2011), as it has the least amount of intra-individual variability and greatest amount of sensitivity, particularly to changes in speeds (Albertus-Kajee, et al., 2011). Moreover misleading results can be generated through single definitive thresholds (Ozgunen, Celik, & Kurdak, 2010). Therefore muscle-specific thresholds need to be used when identifying muscle on-off times. Furthermore, there is often great intra- and inter-individual variation when basing thresholds on multiples of SD away from the baseline, making it difficult to set a definitive multiplication factor (Konrad, 2005). Thus, the peak amplitude was used as a reference for threshold
detection, as it is believed to be a much more reliable threshold as it is independent from baseline variation and characteristics (Konrad, 2005).

In chapter 9 an amplitude-temporal based coactivation was also calculated due to its potential to be more sensitive to changes in muscular activation. This was necessary as the focus was on changes in cushioning and proprioception and not speed, as was the case in chapter 7. Before coactivation could be quantified the rectified raw EMG data were submitted to a nonlinear Teager-Kaiser energy operator (EMG\textsubscript{TKEO}) (Hortobagyi et al., 2009; Li, Zhou, & Aruin, 2007). EMG\textsubscript{TKEO} increases the signal to noise ratio and therefore helps to detect muscle on-off times by calculating the energy of the signal using both the amplitude and frequency of the EMG signal. From just three discrete, consecutive time points (EMG\textsubscript{n-1}, EMG\textsubscript{n} and EMG\textsubscript{n+1}), the energy of the signal at time point ‘n’ can be calculated using the following formula (Lauer & Prosser, 2009):

\[
\text{Signal Energy} = EMG_n^2 - (EMG_{n+1}EMG_{n-1})
\]  
\[
\text{(Eq. 3)}
\]

The EMG\textsubscript{TKEO} data was passed through a recurrent low-pass filter with a 40 Hz cut-off frequency to avoid phase shift (Hodges & Bui, 1996). Muscle on-times were derived in a similar fashion to these previously used during the temporal coactivation calculation detailed above, whereby computed thresholds set at a percentage of peak EMG\textsubscript{TKEO} was compared to manually derived thresholds. The muscle on-times were then applied to the raw EMG data after filtering and normalisation. The raw EMG was low-pass filtered with a zero-phase shift and a cut-off frequency of 6 Hz. Each muscle’s filtered data were normalised to its mean muscle activity during five gait cycles in the control condition. The control condition represented when participants were SH and running with a SH stride length. Then each muscle was integrated using the Trapz function in MATLAB and each specific muscle pairing (outlined in Chapter 9) was entered into equation 4. This coactivation calculation has been used in previous gait studies (Franz & Kram, 2012; Peterson & Martin, 2010) and was only used on muscle data during the stance phase of running.

\[
\text{Coactivation} = 2 \times \left[ \frac{\min(EMG_1, EMG_2)}{\int EMG_1 + \int EMG_2} \right] \times 100
\]  
\[
\text{(Eq. 4)}
\]
Min is the minimum of the two EMG signals ($EMG_1$, $EMG_2$). The coactivation value generated is represented as a percentage, and referred to by others as a coactivation index (Franz & Kram, 2012; Peterson & Martin, 2010).

The integrated EMG (iEMG) of each muscle was also calculated in chapter 9, both during stance and preactivity. The muscle on-times derived from the $EMG_{TKEO}$ data underwent the same low-pass filtering as mentioned above, but was integrated using the Trapz function prior to normalisation. Each muscle was then normalised to the average iEMG data of each respective muscle over five gait cycles from the control condition. Pre-activity was defined as 100 ms prior to TD (Albertus-Kajee, et al., 2011), and stance was defined as TD to TO. The resulting iEMG stated in arbitrary units in chapter 9 are technically % s.

5.4.3 Kinematic Calculations

All raw coordinate kinematic data in the experimental investigations were smoothed using a fifth-order quintic spline function that was incorporated within the Vicon system. High order functions are preferred over lower order ones, such as cubics (Soudan & Dierckx, 1979), providing an optimal smoothing technique (Woltring, 1985) that gives accurate first and second derivatives of coordinate data (Soudan & Dierckx, 1979; Vaughan, 1982). Joint angles were then computed from the smoothed coordinate data.

A joint coordinate system was employed to compute knee and ankle joint angles throughout this thesis following the recommendations from Grood and Suntay (1983) and Wu and colleagues (2002). A typical joint coordinate system implemented is shown in Figure 5.4, with the axes representing specific movements as outlined by Cole, Nigg, Ronsky and Yeadon (1993). The steps involved in the computation of the joint angles were as follows:

1) Medio-lateral axis defined from the proximal segment ($i_{prox}$) representing flexion-extension
2) Vertical axis defined from the distal segment ($k_{dist}$) representing axial rotation of the segment
3) Anterio-posterior, floating axis defined as the cross product between \(i_{\text{prox}}\) and \(k_{\text{dist}}\) (FA) representing ad-abduction

To establish flexion/extension the dot product between the \(k_{\text{prox}}\) and FA was calculated. To establish inversion/eversion (at the ankle) the dot product between the \(k_{\text{dist}}\) and the \(i_{\text{prox}}\) was calculated. All dynamic angles were normalised to standing trials to provide anatomically meaningful angles. Angular velocities and accelerations were calculated by differentiation of the joint angles.

During chapter 6 it was necessary to calculate the leg axis angle. The leg axis was defined as the vector between the hip and lateral malleolus markers. The leg axis angle was defined as an absolute angle between the leg axis and the vertical. If the leg axis was perpendicular to the ground the angle equated to \(0^\circ\). If the lateral malleolus marker was in front of the hip marker, the leg axis vector was directed behind the vertical making the angle negative. If the hip marker was in front of the lateral malleolus marker, the vector was directed in front of the vertical meaning the angle was positive.
Vertical oscillation was measured in chapters 6 and 9 by identifying the displacement between the peak vertical hip position and the minimum vertical hip position during the gait cycle. This gave the maximum vertical range of displacement (‘oscillation’) of the hip marker and was used as an estimate of the centre of gravity.

Heel off is usually determined using a pressure plate, however this was not possible during the treadmill running performed in chapter 9. Given suggestions that BFT and MS running may increase the risk of bone stress injuries at specific sites (Giuliani, Masini, Alitz, & Owens, 2011; Ridge, et al., 2013) trying to determine how long the forefoot is loaded during push-off may provide additional insight into BFT, MS and SH running. The second derivative of the positional data of the inferior calcaneus marker was used to identify the time at which the heel lifted off the ground. The maximum acceleration of the inferior calcaneus marker appeared just after a plateau in the vertical position data (Figure 5.5). This plateau represents when the foot was flat to the ground. Therefore the timing of the maximum acceleration was defined as the instant of ‘heel off’.

![Figure 5.5 Typical vertical displacement and vertical acceleration of the inferior calcaneus marker during stance. The red arrow denotes the time corresponding to ‘heel off’.

5.4.4 Kinetic Calculations

Kinetic calculations derived from the force data were conducted in chapter 6. Prior to obtaining any kinetic values, all three components (F_x, F_y, and F_z) were
adjusted for any ‘noise’ fluctuations in the force plate data. This effectively ‘zeroed’ the baseline force prior to the participants’ initial ground contact. Peak medial (negative $F_x$) and lateral (positive $F_x$) forces were obtained, in addition to the medial, lateral and net $F_x$ impulse. Linear impulse being the definite integral of force over time (Zatsiorsky, 2002), thus is described as the area under the curve.

\[ \text{Impulse} = \int_{T_0}^{T_D} F \, dt \quad \text{(Eq. 5)} \]

TD is the beginning of force application; TO is the end of force application and F is the force component of interest. Peak braking (negative $F_y$) and propulsive (positive $F_y$) force were also obtained. In regards to the $F_z$ data, $F_{z\text{impact}}$, loading rate and active force were calculated. $F_{z\text{impact}}$ is the first initial peak in $F_z$, usually within the first 20% of stance (Munro, Miller, & Fuglevand, 1987). Loading rate is the time derivative of the raw $F_z$ data from TD to $F_{z\text{impact}}$. The middle 60% of this time period was used to calculate the average loading rate and peak instantaneous loading rate. A distinguishable $F_{z\text{impact}}$ was detected in 12 out of the 14 participants (Figure 5.6a), suggestive of a rearfoot strike pattern. The other 2 participants were identified as midfoot strikers based on a slight, but not distinct $F_{z\text{impact}}$ (Figure 5.6b) (Cavanagh & Lafortune, 1980). Therefore their $F_{z\text{impact}}$, as well as loading rate data was excluded from this analysis. Active force was defined as the peak $F_z$ that occurs at push-off, which typically occurs between 30 and 50% of stance (Munro, et al., 1987). The timings of each peak force described above were also determined and presented as a percentage of stance. Defining timing relative to the total ground contact time allowed direct comparison between pre and post data.

The angle of the GRF$_r$ vector was also calculated in chapter 6. This was computed relative to the vertical, meaning that a GRF$_r$ perpendicular to the ground i.e. directed straight up, had an angle of $0^\circ$. Anything directed behind the vertical, posteriorly is presented as a negative angle and anything directed in front of the vertical, anteriorly is presented as a positive angle.
Other kinetic variables determined in chapter 6 were frontal and sagittal joint moments, which were calculated using inverse dynamics. Using the ankle joint and foot segment as an example a brief outline of the equations used will be presented (see Appendix B for detailed MATLAB script). The first necessary step to take involves transforming all the parameters required for the calculation of joint moments (GRF, centre of pressure, force of segment resulting from gravity, segment centre of mass accelerations, proximal and distal moment arms, and proximal and distal joint centre locations) from the global coordinate system to the local coordinate system (LCS) of each segment under investigation (Hamill & Selbie, 2004; Hof, 1992). Using the ankle joint as an example, this means transforming all the coordinates into the LCS of the foot through a rotation matrix.

Figure 5.6 a) Typical Fz graph of a rearfoot strike pattern. b) Typical Fz graph of a midfoot strike pattern. The areas relating to impact peaks and loading rate are highlighted on the graphs.
Foot<sub>RF</sub> $= \begin{bmatrix} x_{foot}(i,x) & x_{foot}(i,y) & x_{foot}(i,z) \\ y_{foot}(i,x) & y_{foot}(i,y) & y_{foot}(i,z) \\ z_{foot}(i,x) & z_{foot}(i,y) & z_{foot}(i,z) \end{bmatrix}$ (Eq. 6)

Foot<sub>RF</sub> is the foot reference frame; $x_{foot}$ is the medio-lateral vector of the foot segment; $y_{foot}$ is the anterior-posterior vector of the foot segment; $z_{foot}$ is the vertical vector of the foot segment; $i$ is the frame number; $x$, $y$ and $z$ denote the separate components of the vector. Once all the parameters have been transformed into the LCS the net ankle moment, which comprises of both translations and rotational dynamics (Hamill & Selbie, 2004; Hof, 1992), can be calculated. To determine segment parameters used in the calculation (i.e. mass, length, centre of mass position, inertia components) specialised regression equations that take into account each participant’s individual body mass and height were implemented (Shan & Bohn, 2003). As such these parameters are much more specific to the individual than that using data derived from cadavers, however density values were taken from the work of Dempster (1955). Then by using inverse dynamics, equations are developed to estimate the moment acting on the ankle. Firstly, moment arms were calculated.

$$d_{CoM} = CoM_{foot} - JC_{ank} \quad \text{(Eq. 7)}$$

$$d_{CoP} = CoP_{foot} - JC_{ank} \quad \text{(Eq. 8)}$$

$JC_{ank}$ is the ankle joint centre; $CoM_{foot}$ is the centre of mass of the foot; $CoP_{foot}$ is the point of force application upon the foot. The vectors $d_{CoM}$ and $d_{CoP}$ denote specific moment arms (Figure 5.7); these were then used in the calculation of the separate components that make up the net ankle moment.

$$M_{GRF} = d_{CoP} \times GRF_{foot} \quad \text{(Eq. 9)}$$

$$M_{WEIGHT} = d_{CoM} \times m_{foot}g \quad \text{(Eq. 10)}$$

$$M_{EFF} = d_{CoM} \times m_{foot}a_{foot} \quad \text{(Eq. 11)}$$

$$M_{ank} = - T_q - M_{GRF} - M_{WEIGHT} + M_{EFF} + I\alpha_{foot} \quad \text{(Eq. 12)}$$
$M_{\text{GRF}}$ is the moment applied by the GRF; $\text{GRF}_{\text{foot}}$ is the GRF exerted upon the foot; $M_{\text{WEIGHT}}$ is the moment applied by the weight of the segments, in this example the weight of the foot; $M_{\text{EFF}}$ is the moment applied by the acceleration forces; $T_q = \text{GRF}$ torque vector; $I$ is the moment of inertia matrix; $\alpha_{\text{foot}}$ is the foot angular acceleration matrix. Therefore $I\alpha_{\text{foot}}$ represents the moments due to rotational acceleration and all other components represent moments due to translational acceleration. $M_{\text{ank}}$ is the ankle moment. This equation was derived from the work of Hof (1992) and Zatsiorsky (2002). The inverse dynamics procedure goes from distal to proximal, therefore once the ankle joint moments had been calculated the information was used in the calculation of the knee joint moments.

Figure 5.7 A free body diagram of the foot segment, illustrating the forces, moments of forces and geometry of the foot. All parameters are represented in the LCS of the foot (adapted from Hamill and Selbie, 2004). $R_{\text{ank}}$ is the intersegmental joint reaction force. All other variables are described previously.
Joint (‘apparent’) stiffness, defined as the ratio of change in joint moment to joint angular deflection (Zatsiorsky, 2002), was also determined in chapter 6. It is acknowledged that although many refer to this as joint stiffness (Arampatzis, Brüggemann, & Metzler, 1999; Brughelli & Cronin, 2008; Stefanyshyn & Nigg, 1998), it does not refer to stiffness in the mechanical sense whereby only passive structures are under investigation. Rather the active elements of the musculoskeletal system (e.g. muscles) play a role in determining segmental joint angles (Zatsiorsky, 2002). Therefore no external forces are needed to directly change the stiffness of a joint. However in light of the previous literature, and thus to aid comparisons, it will be referred to as joint stiffness from here on in.

\[ k_{\text{joint}} = \frac{\Delta \text{joint moment}}{\Delta \text{joint angle}} \]  

(Eq. 13)

### 5.5 Statistical Methods

The statistical analyses in all investigations were conducted using the Statistical Package for Social Sciences (version 18, PASW). The specific statistical tests utilised in each investigation are detailed in the experimental chapters. Prior to the statistical tests being carried out on the data, normality was assessed using the Kolmogorov-Smirnov test. If this was violated appropriate non-parametric tests were performed. Statistical significance was set at \( p \leq 0.05 \) and unless otherwise stated data are presented as means ± SD.
Chapter 6 Mechanisms for improved running economy in beginner runners


6.1 Introduction

Running economy (RE), the rate of oxygen an individual consumes at a given speed, is reported to be a good predictor of running performance (Conley & Krahenbuhl, 1980). Saunders et al. (2004a) and Jones and Carter (2000) identified a consensus in the literature that trained runners exhibit a better RE than untrained runners. Running training can lead to improvements in RE (Franch, et al., 1998; Jones, Carter, & Doust, 1999), although the evidence regarding the relationship between running training and RE improvements is equivocal (Daniels, Yarbrough, & Foster, 1978). A contributory factor to these inconsistencies is the initial training status of the participants, with enhancements to RE more likely to occur in untrained individuals than in trained runners (Saunders, et al., 2004a).

Evidence shows that trained runners can utilise a lower percentage of their maximal oxygen consumption (% $\dot{V}O_{2\text{max}}$) at a given submaximal running speed than untrained runners. The better RE in trained runners is associated with a lower percentage of maximal heart rate (%HR$_{\text{max}}$) and with lower minute ventilation ($\dot{V}E$) (Bransford & Howley, 1977; Pate, Macera, Bailey, Bartoli, & Powell, 1992). It has been reported that decreases in $\dot{V}E$ can account for 70% of the improvements in RE (Franch, et al., 1998). However, determinants of RE are not just limited to physiological factors; anthropometric, environmental and biomechanical factors may also be important (Saunders, et al., 2004a).

The biomechanical factors potentially influencing RE encompass kinematics, kinetics, flexibility, and elastic energy storage in the stretch-shortening cycle (SSC) (Saunders, et al., 2004a). Running mechanics, specifically the kinematic variables of shank angle at touchdown (TD) and plantarflexion at toe-off (TO) in
addition to net positive power, have been reported to explain 54% of the variance in RE (Williams & Cavanagh, 1987). These mechanics are believed to have developed through a process of self-optimisation (Williams & Cavanagh, 1987), as individuals adopt a running gait which is most economical for them (Cavanagh & Williams, 1982). Cross-sectional studies have identified various other kinematic, kinetic and flexibility variables to be associated with better RE (Heise & Martin, 2001; Kyrolainen, et al., 2001; Trehearn & Buresh, 2009; Williams & Cavanagh, 1987; Williams, et al., 1987), such as knee extension at TO, vertical impulse, and lower back and hamstring flexibility. Yet there are appreciable inter-individual differences (Conley & Krahenbuhl, 1980) and evidence is often inconsistent between studies. For example Kyröläinen et al. (2001) argue that running mechanics are poor predictors of RE, contradicting Williams and Cavanagh (1987). Furthermore, Kyröläinen et al. (2001) identified the braking kinetic force as the main factor explaining RE and not kinematic variables.

Currently little is actually known about the development of an economical running gait, primarily because research has focused on trained runners. A limited number of studies have examined how individuals develop their gait (Lake & Cavanagh, 1996; Nelson & Gregor, 1976) with only a shortening of stride length being observed (Nelson & Gregor, 1976). Gait manipulation research suggests that self-selected traits are near optimal for oxygen consumption and manipulations to stride length and vertical oscillation away from these self-selected parameters can decrease economy (Cavanagh & Williams, 1982; Egbuonu, et al., 1990; Tseh, et al., 2008). Although informative, these studies only demonstrate the outcome of adjustments to running mechanics and do not consider the underlying changes in kinematics and/or kinetics, which may cause them. Additionally, by studying runners who already exhibit their optimal running gait to examine RE associations, it is difficult to discern whether the biomechanical traits are inherent in those runners or a feature of gait development with training.

The purpose of this study was, therefore, to explore the effect of a 10-week running programme on the running mechanics and RE of beginner runners. The aim of this study was to identify if mechanical or physiological variables
changed over 10 weeks of running in beginners and whether these changes could account for any change in RE.

6.2 Methods

6.2.1 Participants

Fourteen female beginner runners (age 34.1 ± 8.8 yrs, height 1.64 ± .09 m, body mass 69.1 ± 10.8 kg) volunteered for the study through a 10-week beginner's running programme (10wkRP). Fourteen was calculated as an appropriate sample size to provide 80% power to detect changes in kinematics based on magnitudes found in previous gait training studies (Messier & Cirillo, 1989; Tseh, et al., 2008). A beginner runner was defined as an individual having had no prior running training and not being involved in regular sporting activities. All participants were free from injury prior to data collection and did not sustain any injury to the lower extremities during the 10wkRP. They were also free from cardiac abnormalities and provided written informed consent and a medical and athletic history, which covered previous injuries and sports involvement. Ethical approval was given by the University of Exeter Sport and Health Sciences Ethics Committee.

6.2.2 Procedure

Data collection occurred during four laboratory visits: session one occurred prior to initiating the 10wkRP, session two occurred 3 weeks after beginning the 10wkRP, and session three and four took place at least two days apart, after completion of the 10wkRP. Session one and three consisted of a gait analysis, flexibility assessment, a graded exercise test (GXT) and body mass (SECA, Hamburg, Germany) and stature measurements; RE was assessed during session two and four. For both the GXT and RE measurements the same motorised treadmill (Woodway, PPS 55 sport slat-belt treadmill, Germany) was used. Heart rate (HR) was measured via a wireless chest strap telemetry system (Polar Electro T31, Kempele, Finland) and respiratory gas exchange was measured every 10 s using an automated gas analysis system (Cortex Metalyser II, Biophysik, Leipzig, Germany).
6.2.3 Gait analysis

Gait analysis sessions involved the simultaneous collection of kinematic and force plate data during running. A three-dimensional bilateral kinematic analysis was performed using an eight camera motion capture system (Vicon Peak, 120 Hz, automatic, opto-electronic system; Peak Performance Technologies, Inc., Englewood, CO), with cameras positioned in an oval shape around a single floor mounted force platform (960 Hz, AMTI, Advanced Mechanical Technology, Inc., Massachusetts) located half-way down a 12m run-way. Synchronisation of the force and kinematic data occurred within the Vicon software using the initial foot contact as an automatic event detection (vertical force > 10N). A fifth order quintic spline filter was applied to the raw kinematic data within the Vicon system. Kinetic calculations were conducted on the raw force plate data.

Participants were issued with a standardised Adidas neutral cushioning shoe (Dixon & McNally, 2008) of appropriate size and then performed a five minute warm-up to become accustomed to the footwear and the data collection environment. Eleven reflective markers were attached to the following anatomical positions to denote the anatomical position of the thigh, shank and foot using a modified Soutas-Little et al. model (1987): proximal greater trochanter (hip), medial and lateral condyles (knee), the musculotendinous junction where the medial and lateral belly of the gastrocnemius meet the Achilles tendon, the mid tibia below the belly of the tibialis anterior, lateral malleolus (ankle), superior and inferior calcaneus, the third proximal head of the third metatarsal and the distal head of the fifth metatarsal joint. A twelfth reflective marker was placed on the inferior calcaneus of the opposite foot to allow for the calculation of step length. The twelve markers were affixed for the data capture of one leg and then removed and attached to the opposite leg to record the next set of data, as data for each leg were collected separately using a block randomised order to reduce potential familiarisation effects.

Angles were normalised to standing by the collection of a single standing trial on the force plate, in the anatomic position. The resulting standing joint angles were subtracted from angles gathered during the dynamic movement, this adjustment providing anatomically meaningful values. Familiarisation trials
were performed until participants were deemed to be comfortable at the
required running velocity of 152 m min\(^{-1}\) (± 5%). The velocity was monitored by
two sets of timing gates; each set positioned either side of the force plate. Ten
successful trials were recorded per leg and a total of thirty-six biomechanical
variables were collected for each leg (Appendix C). The time of occurrence for
each peak value was reported as a percentage of stance time. The force
conventions used were medial-lateral (F\(_{ML}\)), anterior-posterior (F\(_{AP}\)) and vertical (F\(_V\)).

After the data collection of the first leg, the participants’ flexibility was assessed.
This consisted of the sit-and-reach test (SRT) and calf flexibility stretches for
both legs. To perform the SRT participants were instructed to keep their legs
extended and flat to the floor, with their feet flat against the sit-and-reach box,
whilst they reached as far forward as possible. Calf flexibility was assessed
using the methodology described by Bennell et al. (1998) which uses an
inclinometer to measure the angle between the shank and the vertical during
maximal knee extension in a standing, lunge position. Each test was performed
three times to allow a mean to be calculated. The gait analysis was then
repeated for the second leg.

6.2.4 Graded exercise test (GXT) to \(\dot{V}O_{2\text{max}}\)

After completion of the gait analysis participants performed the Balke-Ware
(Balke & Ware, 1959) walking GXT to volitional exhaustion. Prior to the GXT
participants were given time to become accustomed to walking on the treadmill.
Initially, the gradient was set to 0% for the first minute of the GXT, after which
the gradient was increased to 2% for the following minute. Every minute
thereafter the gradient was increased by 1% until the participant reached
volitional exhaustion. Throughout the test a constant speed of 90 m min\(^{-1}\) (5.4
km h\(^{-1}\)) was used. \(\dot{V}O_{2\text{max}}\) was calculated by taking the average of the three
highest consecutive 10 s measurements during the test. HR was recorded
every minute as were the participants’ ratings of perceived exertion (RPE: Borg,
1998). Information regarding test duration, distance travelled and running speed
was obscured from view for the duration of the GXT.
6.2.5 Running Economy (RE) test

A familiarisation run on the treadmill was performed before the RE assessment. A minimum of 6 minutes were used during the familiarisation to enable a natural running style to be achieved as is required when comparing treadmill and over ground running (Lavcanska, et al., 2005). This period served as the participants’ warm-up. RE was measured on a level treadmill over three test speeds in the following order: 125, 138 and 152 m·min⁻¹. These speeds were chosen following the recommendations that test speeds should be representative of training speeds for RE assessment (Daniels & Daniels, 1992; Jones & Carter, 2000). Although not a randomised protocol, fatigue effects were minimised by progressing from the slowest to the fastest speed. Each speed was sustained for 6 minutes, with 9 minute rest periods between consecutive running bouts. Oxygen consumption (\(\dot{V}O_2\)) was measured during the final 2 minutes of each bout of running and the mean \(\dot{V}O_2\) was calculated. All three \(\dot{V}O_2\) values were used to calculate RE. Additionally, HR was determined by averaging the final two minutes of each test and then the mean HR was calculated from the combination of all three velocities.

6.2.6 10 week beginners’ running programme

The 10wkRP used a combination of walking and running to gradually build-up an individual’s constant running time (Appendix D). Sessions were for women only and were performed within a group setting once a week led by qualified leaders. The leaders set weekly ‘homeworks’ which were to be performed in the individual’s own time. Women were encouraged to run at their own pace throughout the 10wkRP, aiming to be able to run for 30 minutes continuously by the end of the programme.

6.2.7 Statistical analysis

Means, standard deviations and 95% confidence intervals were calculated for all test variables, both for pre- and post- measurements. The Kolmogorov-Smirnov test was performed on all measured variables to determine their distribution. All the variables were normally distributed and consequently paired
samples t-tests were performed (pre- v post- 10wkRP). Stepwise multiple regression was performed on those variables found to significantly change over time, in order to identify which variables significantly contributed to any change in RE. Data analysis was conducted using PASW (Predictive Analytics Software) statistics version 18. Statistical significance was defined as p ≤ 0.05.

6.3 Results

Based on the log book records, all weekly sessions and ‘homework’ sessions were completed by everyone except two participants. These two individuals were unable to attend one running session each. A total of four participants withdrew from the 10wkRP, due to being unable to commit to the weekly sessions, therefore post data could not be collected for these individuals.

6.3.1 Physiological measurements

Running economy was found to significantly improve between pre- and post-measurements (Table 6.2). This was true even if data from only the final speed, which equates to the velocity used during the biomechanical assessment, was used to calculate RE. Time-to-exhaustion during the GXT significantly increased from pre- (16.4 ± 3.2 mins) to post- (17.3 ± 2.8 mins), whereas \( \dot{V}O_{2\max} \) was similar between the two measurements (34.7 ± 5.1 mL kg\(^{-1}\) min\(^{-1}\) and 34.2 ± 5.6 mL kg\(^{-1}\) min\(^{-1}\), pre- and post- respectively). Mass remained unchanged from pre- (69.1 ± 10.8 kg) to post- (70.3 ± 10.6 kg) measurements. The % \( \dot{V}O_{2\max} \), \( \dot{V}E \), HR and RER during submaximal running were also not significantly different between pre- and post- measurements (Table 6.1).

6.3.2 Flexibility

The SRT scores were similar between the two sets of measurements (13.2 ± 9.9 cm and 13.6 ± 9.8 cm, pre- and post- respectively). The left leg calf flexibility significantly decreased from pre- (27.3 ± 6.3\(^{\circ}\)) to post- (23.9 ± 5.6\(^{\circ}\)). In the right leg, a similar trend was observed, but this was not statistically significant (28.6 ± 5.2\(^{\circ}\) and 24.6 ± 7.6\(^{\circ}\), pre- and post- respectively).
6.3.3 Biomechanical variables

Seven biomechanical variables were found to significantly change from pre- to post- (Table 6.2). Kinematic analysis revealed that the knee was significantly less extended and the ankle was significantly less plantarflexed at TO after the 10wkRP compared to baseline in the left leg. Peak dorsiflexion of the right leg occurred later in stance post- running compared to baseline. At TD both the ankle plantarflexion velocity and ankle eversion velocity of the right leg became significantly slower after the 10wkRP compared to baseline. Peak eversion velocity became significantly lower for the right leg with more running experience.

Table 6.2 Means ± (SDs) and 95% confidence intervals (CI) for the biomechanical variables that significantly changed over time

<table>
<thead>
<tr>
<th>Variable</th>
<th>Right</th>
<th>Pre (n=14)</th>
<th>Post (n=10)</th>
<th>95% CI</th>
<th>Left</th>
<th>Pre (n=14)</th>
<th>Post (n=10)</th>
<th>95% CI</th>
</tr>
</thead>
<tbody>
<tr>
<td>TO knee extension (º)</td>
<td></td>
<td>164.2 ± 4.6</td>
<td>159.4 ± 8.6*</td>
<td>-2.74 - 12.5</td>
<td>165.9 ± 4.3</td>
<td>157.0 ± 3.6*</td>
<td>4.62 - 13.20</td>
<td></td>
</tr>
<tr>
<td>TO plantarflexion (º)</td>
<td></td>
<td>-21.3 ± 11.2</td>
<td>-19.8 ± 5.5*</td>
<td>-7.60 - 4.67</td>
<td>-25.0 ± 8.6</td>
<td>-18.7 ± 7.5*</td>
<td>-11.74 - 0.72</td>
<td></td>
</tr>
<tr>
<td>TD plantarflexion velocity (*s⁻¹)</td>
<td></td>
<td>-194.5 ± 81.8</td>
<td>-194.3 ± 26.1*</td>
<td>-151.4 - 49.1</td>
<td>-146.7 ± 109.2</td>
<td>-91.1 ± 37.8*</td>
<td>-150.39 - 39.27</td>
<td></td>
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<tr>
<td>TD eversion velocity (*s⁻¹)</td>
<td></td>
<td>100.6 ± 35.0</td>
<td>-59.2 ± 30.7*</td>
<td>-65.5 - 17.4</td>
<td>-91.9 ± 41.5</td>
<td>-69.5 ± 28.2*</td>
<td>-56.51 - 11.59</td>
<td></td>
</tr>
<tr>
<td>Peak eversion velocity (*s⁻¹)</td>
<td></td>
<td>-110.1 ± 47.1</td>
<td>-77.1 ± 27.1*</td>
<td>-61.8 - 41.8</td>
<td>-97.2 ± 39.4</td>
<td>-83.9 ± 29.1*</td>
<td>-47.08 - 20.55</td>
<td></td>
</tr>
<tr>
<td>Timing of peak dorsiflexion (%)</td>
<td></td>
<td>49.6 ± 6.9</td>
<td>56.2 ± 2.5*</td>
<td>-12.11 - 1.06</td>
<td>53.3 ± 3.5</td>
<td>55.9 ± 3.3*</td>
<td>-6.65 - 1.50</td>
<td></td>
</tr>
<tr>
<td>Peak propulsive force (BW)</td>
<td></td>
<td>0.193 ± 0.040</td>
<td>0.225 ± 0.036*</td>
<td>-0.054 - 0.011</td>
<td>0.198 ± 0.042</td>
<td>0.222 ± 0.031</td>
<td>-0.048 - 0.001</td>
<td></td>
</tr>
</tbody>
</table>

*Significantly different between pre and post (p < .05).
§ Denotes which variables show similar trends to the significant result observed in the opposite leg.
Two individuals were identified as being mid-foot strikers due to an indistinguishable initial $F_V$ peak and were excluded from the analysis of this variable. Peak propulsive force significantly increased from baseline to post-10wkRP in the right leg and was the only kinetic variable to significantly change over time.

6.3.4 Regression Analysis

Regression analysis was performed on all the variables that were found to significantly change. Previous research has considered trends within-groups when investigating RE and running mechanics due to the large variance found (Williams & Cavanagh, 1987). Therefore from the lower extremity variables that had significantly changed, if both legs exhibited a similar trend they were both entered into the regression analysis. The results revealed that a significantly less extended knee at TO, peak dorsiflexion occurring significantly later in stance and a slower eversion velocity at TD explained 94.3% of the variance in the change in RE (Table 6.3).

| Table 6.3 Predictive model for changes in running economy in beginner runners |
|-----------------------------|----------------|----------------|----------------|----------------|
| Variable                    | $Unstandardised$ coefficients | Beta | $R^2$ (adjusted $R^2$) | $p$-value |
| Timing of peak dorsiflexion (%) | -3.054 | .388 | -.738 | .001 |
| TO knee extension (º)       | -1.209 | .207 | -.545 | .004 |
| TD eversion velocity (º s$^{-1}$) | -.189 | .064 | -.282 | .042 |
| Constant                    | 5.731 | .968 (.943) | .176 |

$^a$ Left leg variable. $^b$ Right leg variable.

6.4 Discussion

The aim of this study was to identify if mechanical or physiological variables changed over 10 weeks of running in beginners and whether these changes could account for any change in RE. The results revealed seven biomechanical variables, calf flexibility and time-to-exhaustion significantly changed with an increase in running experience. Of these, eversion velocity at TD, timing of peak dorsiflexion and knee extension at TO contributed significantly to the change in RE, collectively accounting for 94.3% of the variance.
6.4.1 Changes in running mechanics

Research suggests that visual and verbal feedback of running gait can help an individual alter how they run (Crowell & Davis, 2011; Messier & Cirillo, 1989) but that completing a running programme does not necessitate a change in running mechanics (Lake & Cavanagh, 1996). To our knowledge the current study is the first to utilise a detailed kinematic and kinetic analysis to observe individuals using a self-optimising process to develop their running gait with increased running. It appears that during 10 weeks of running, individuals begin to adapt their running style, producing a gait which is more economical than their initial gait. Previously this was a theoretical concept (Williams & Cavanagh, 1987) lacking empirical evidence.

The values obtained at baseline for peak eversion velocity are comparable to those reported previously for female recreational runners (Bischof, Abbey, Chuckpawong, Nunley, & Queen, 2010), although others have found much higher values ($223\,\text{º}\cdot\text{s}^{-1}$) (1998). The slower peak eversion velocities observed for beginner runners after 10 weeks of running may have developed as a protective mechanism to reduce the strain on the musculoskeletal system which could otherwise lead to overuse injuries (Willems et al., 2006). Such low eversion velocities found in the post-10wkRP data are, however, unsupported by previous research. It is likely that the slow running velocity ($152\,\text{m}\cdot\text{min}^{-1}$) and low mean peak eversion angles (-3.38 and -3.35º, right and left respectively) contributed to this finding. This contrasts with previous running literature which reports peak eversion angles between -9 and -16º for test velocities ranging from 186 to 240 m\cdot min^{-1} (Diss, 2001; Fukuchi & Duarte, 2008; McClay & Manal, 1998).

Kinetically, it appears that with increased running experience, beginner runners can generate greater propulsive forces without compromising upward force or affecting sideways adjustments as both the peak $F_V$ and $F_{ML}$ forces remained unchanged. A mechanism that may have accounted for some of the change in peak propulsive force may be the ankle angle differences at TO. The ankle was more flexed at TO after the 10wkRP, possibly as a result of peak dorsiflexion occurring later during stance. It is suggested that at TO more force could be
generated in the direction of the run. A similar difference in plantarflexion at TO has been observed in runners with better RE (Williams & Cavanagh, 1987). Taken together, this suggests that the positioning of the foot leaving the ground can be modified, affecting other mechanical variables and influencing $\dot{V}O_2$. However, a greater understanding of the interactions between mechanical variables and their resultant effect upon RE is needed before firm conclusions can be reached.

6.4.2 Relationship between running mechanics and running economy

The results revealed that biomechanical variables can explain 94.3% of the variance in RE when both legs are considered. This supports the findings of Williams and Cavanagh (1987) who reported that 54% of the variance in RE can be accounted for by the shank angle with the vertical at TD, maximal plantarflexion angle (which occurred at TO) and net positive power. Contrary to these results, Lake and Cavanagh (1996) found that after six weeks of running training there were no gait adaptations and no relationship between RE and running mechanics. Together, these results suggest that adaptations may occur between six and ten weeks. However, Lake and Cavanagh (1996) only used six biomechanical variables based on previous evidence obtained using trained runners, and therefore they may have missed any changes specific to a novice gait.

The significant contribution of knee extension at TO to the variance in RE change suggests that reduced knee extension at TO is a feature of economical female running gait. Although there have been discrepancies reported regarding variations in knee kinematics with gait manipulations and/or altered RE (Messier & Cirillo, 1989; Williams & Cavanagh, 1987; Williams, et al., 1987), this finding is consistent with observations of elite female runners (Williams, et al., 1987). Thus, reduced knee extension at TO is a quality found both in elite female runners, with an established gait pattern, and beginner female runners developing their gait.
The observed change in both the knee and ankle extension at TO mean a less extended leg is generated (Figure 6.1), but the mechanism through which this translates to better RE has yet to be explained. Given that extension of the lower extremities helps to propel the body vertically upwards, facilitating the support leg’s clearance of the ground during its swing phase, some extension is necessary. However it is possible that the leg is in a better position for the swing phase, when less extended, meaning less energy is expended in flexing the leg during swing.

![Figure 6.1 Differences in knee angle and ankle angle at TO between pre and post measurements.](image)

By increasing the length of time spent dorsiflexing, beginner runners spent longer in the eccentric phase of the SSC after training, facilitating elastic energy storage during the absorption phase of ground contact (Mann & Hagy, 1980). The results suggest that prolonging dorsiflexion, towards the higher end of the expected occurrence time of 50-60% of stance (Rodgers, 1988), is more economical than a shorter dorsiflexion time. This is because runners will be able to enhance the performance of the propulsive, concentric phase of the SSC due to an improved eccentric phase (Komi, 2000). The calf muscles became 3.4 and 4° (left and right legs respectively) less flexible after training, suggesting increased calf muscle stiffness, which could also have implications for the SSC. Increasing the stiffness of calf muscle-tendon units contributes to improving RE (Kyrolainen, et al., 2003) potentially via reducing muscle activation. Further research incorporating joint and muscle moment data, in addition to the
kinematics and SSC is necessary to fully understand the biomechanical relationship with the SSC.

The change in TD eversion velocity coupled with a change in TD plantarflexion velocity suggests that approach kinematics, and possibly muscular activity, were altered with an increase in running experience. This observation is consistent with the suggestions of Williams and Cavanagh (1987) that changes to approach kinematics can contribute to oxygen consumption differences. Bonacci et al. (2010) found that seven out of eight triathletes who showed a change in muscular activity also altered their running mechanics after a cycle-run transition compared to a control run. Their results suggested that 73% of the variance in RE can be explained by changes to sagittal plane knee and ankle TD angles. However swing kinematics and electromyography (EMG) data were not analysed in the current study so changes prior to TD can only be speculated upon.

The beginner runners became 8.4% more economical. Physiologically, only time-to-exhaustion improved and thus, only biomechanical factors could account for the variance in RE change. This contrasts with previous research which suggests that physiological differences predominantly explain changes in RE with training (Franch, et al., 1998; Lake & Cavanagh, 1996). Some studies of recreational athletes or runners have failed to find alterations to running mechanics (Franch, et al., 1998; Lake & Cavanagh, 1996). This highlights the importance of using beginner runners with limited prior running experience to improve understanding of RE development. Furthermore, as symmetry was not assumed in the current study, the right and left leg were comprehensively analysed meaning trends between both legs could be observed.

It is important to note that whilst biomechanical rather than physiological changes were clearly responsible for the improved RE in beginner runners following short-term training, physiological changes (perhaps in addition to further biomechanical changes) might contribute to continued improvements in RE in elite athletes or following long-term training (Jones, 2006). These changes might include a lower oxygen cost of cardiac or respiratory work,
changes in muscle stiffness, or transformation of fibre types from type II to type I (Saunders, et al., 2004a).

It is unclear why there were no improvements in $\dot{V}O_{2\text{max}}$ given the initially low fitness levels and improvement in treadmill time-to-exhaustion. One possibility is the motivational aspect of completing a GXT to volitional exhaustion. Beginner runners may lack the desire to push themselves to volitional exhaustion. On the other hand, the improvement found in RE would have enabled participants to perform for longer before reaching the same $\dot{V}O_{2\text{max}}$, meaning that although the participants increased their time-to-exhaustion they may have still terminated the test at a similar, maximal effort level. Due to the protocol used for the GXT, fitter individuals may terminate exercise because of discomfort to their lower back and calf regions rather than volitional exhaustion. However, it was deemed the most appropriate procedure as it can elicit true $\dot{V}O_{2\text{max}}$ values for individuals with low fitness levels (McArdle, Katch, & Pechar, 1973). Additionally, it must be noted that the first RE measurement occurred three weeks after the initial gait analysis due to these low fitness levels and lack of prior running training experience. The delay was necessary for participants to be able to fulfil the requirement of sustaining six minutes of running at three different speeds.

Caution must be taken in generalising these results as within-group differences were often large in many of the biomechanical variables. Additionally, as Williams and Cavanagh (1987) have suggested, the combined effect of the change in running mechanics should perhaps be used to understand why the runners became more economical rather than the single set of variables forming the regression equation. This notion may explain why two of the variables found to significantly explain the variance in this change in RE were only included in the analysis as a result of exhibiting a similar trend to the opposite leg rather than a significant change. Therefore if only one side of the body had been analysed, trends would have been missed and they would not have been considered. Their inclusion demonstrates that gait adaptations occur in both sides of the lower extremities but to differing degrees and that, perhaps, it is not the magnitude of change but the effect that the change has on other mechanical variables that is important for developing an economical gait.
Exploring ways of understanding the interaction of biomechanical variables and their effect on economy is encouraged as opposed to studying variables in isolation.

Even though the test speeds used for the RE assessment were relatively slow in comparison with previous studies, the $\% \dot{V}O_{2max}$ elicited during the RE assessments were high, yet below the 90% $\dot{V}O_{2max}$ outlined by Daniels and Daniels (1992). Biomechanical comparisons with the literature were, however, limited by using a slow velocity since kinematics and kinetics change as a result of velocity. There were also two different procedures used for the RE and running gait assessments: treadmill and over ground, respectively. Whilst both modes of running have been used in previous studies examining running mechanics and RE (Heise & Martin, 2001; Williams & Cavanagh, 1987) and can produce similar values for submaximal oxygen consumption (Bassett et al., 1985), kinematic differences have been observed (Wank, Frick, & Schmidtbleicher, 1998), especially in the knee (Riley et al., 2008). Generally, however, it is considered that treadmill running provides a good representation of over ground running (Riley, et al., 2008). Additionally adequate familiarisation to treadmill running was given to each participant, an important prerequisite when using both modes of running (Lavcanska, et al., 2005). A pilot study from our laboratory also demonstrated that running with respiratory apparatus does not alter running kinematics suggesting that the instrumentation required to measure gas exchange during treadmill running had no bearing on the results.

6.5 Conclusion

This study has demonstrated that beginner runners use a self-optimisation process to develop their running gait with training. This natural modification to running gait explained 94.3% of the variance in the change in RE that was observed.
6.6 Further Data Analysis from Study 1 (not included in the above paper)

6.6.1 Joint moments and stiffness

A customised MATLAB code was written to calculate three dimensional joint moments, using a standard inverse dynamics approach (Hof, 1992). Regression equations, based on data derived from Caucasian females and the mass and height of each participant, were used to determine segmental inertial parameters (Shan & Bohn, 2003). The centre of pressure and free torque values were calculated from the GRF, which was taken from the force plate data. Joint moments were calculated from the ground up and expressed in a nonorthogonal reference frame or joint coordinate system. Mid-knee was determined as the mid-point between the lateral and medial knee markers. The midpoint of the ankle and forefoot were calculated using callipers, to obtain a measured width, and the lateral malleolus and inferior calcaneus markers, to obtain relative positions. The greatest differences in kinematics were seen in the sagittal and frontal planes, therefore joint moments with respect to these planes were calculated at the ankle and knee. Due to the omission of knee angle at TD data, only joint stiffness at the ankle could be calculated using equation 13.

6.6.2 Resultant GRF and leg axis

The magnitude and angle of the GRFr, in addition to the leg axis angle, were determined (see Chapter 5) at specific time points during ground contact, similar to the work of Chang and colleagues (2000). These reflected the time of peak GRFr, peak braking force and peak propulsive force.

6.7 Further Results from Study 1

Ankle stiffness remained similar between pre- and post- in both legs. There were also no changes (p > 0.05) in peak ankle and knee moments in the sagittal and frontal plane. However, some moments demonstrated a trend in both legs towards smaller peak moments after the 10wkRP than before it (Table 6.4).
The left leg exhibited a smaller difference between the leg axis angle and GRFr post- the 10wkRP compared to pre- during peak propulsive force. This was due to a 64.6% (7.1°) increase in the GRFr angle during propulsion. There were no significant changes in the right leg, although there was a slight increase in GRFr angle during propulsion (15.6%, 1.6°), similar to the left leg (Table 6.5).

Table 6.4 Ankle stiffness and magnitude of peak joint moments of the ankle and knee for the right and left legs, both pre and post 10wkRP

<table>
<thead>
<tr>
<th>Variable</th>
<th>Right</th>
<th>Left</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Before</td>
<td>After</td>
</tr>
<tr>
<td>Ankle stiffness (N·m·deg⁻¹)</td>
<td>15.0 ± 4.7</td>
<td>14.1 ± 3.1</td>
</tr>
<tr>
<td>Ankle plantarflexor (N·m)</td>
<td>186.0 ± 57.4</td>
<td>163.5 ± 38.1^</td>
</tr>
<tr>
<td>Ankle inversion (N·m)</td>
<td>19.9 ± 16.4</td>
<td>9.1 ± 6.6^</td>
</tr>
<tr>
<td>Knee extensor (N·m)</td>
<td>151.7 ± 52.3</td>
<td>149.4 ± 56.8</td>
</tr>
<tr>
<td>Knee abduction (N·m)</td>
<td>64.5 ± 30.2</td>
<td>67.1 ± 21.4</td>
</tr>
</tbody>
</table>

* significantly different between pre- and post- (P < 0.05). ^ non-significant differences between pre- and post- (p < 0.10).

6.8 Discussion of Further Results from Study 1

The magnitude of the peak plantarflexor moment before the 10wkRP is similar to previous research using velocities between 4 and 4.5 m·s⁻¹ (Buczek & Cavanagh, 1990; Kyrolainen, et al., 2001; Stefanyshyn & Nigg, 1998). However, whilst there was no significant decrease in plantarflexor moments after the 10wkRP, the magnitude at this point was closer to those reported at slower, jogging speeds (Winter, 1983) due to a slight non-significant reduction (12.1 and 14.7%, for the right and left leg respectively). This implies that, after training, the beginner runners made minor gait alterations to modify the magnitude of joint loading, exerting similar loads to those exhibited by others at a similar speed. It is hypothesised that this decrease is due to reducing the risk of overuse injuries associated with joint loading (Franz & Kram, 2012; McClay, 2000). The knee extensor moments followed a similar trend as the ankle after increased running experience (1.5 and 10.4% decrease, right and left leg respectively), but are lower than previous reports (Buczek & Cavanagh, 1990;
Kyrolainen, et al., 2001; Scholz, et al., 2008). Runners may have been optimising the joint load in both the knee and ankle towards a more tolerable level. Such alterations could only be made after accumulating numerous steps, by increasing running experience. However the lack of significant difference (at p < 0.05) suggests that the initial level of joint loading was not excessive or certainly not excessive enough to warrant significant gait changes to reduce joint loading.

Table 6.5 GRFr and leg axis angles relative to the vertical during peak, braking and propulsive force (*), for the right and left legs, both pre and post 10wkRP

<table>
<thead>
<tr>
<th>Variable</th>
<th>Right</th>
<th>Left</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Before</td>
<td>After</td>
</tr>
<tr>
<td>GRFr at peak force</td>
<td>-3.6 ± 1.2</td>
<td>-3.8 ± 1.2</td>
</tr>
<tr>
<td>GRFr at peak braking force</td>
<td>-9.7 ± 1.9</td>
<td>-10.4 ± 1.2</td>
</tr>
<tr>
<td>GRFr at peak propulsive force</td>
<td>11.6 ± 1.6</td>
<td>13.2 ± 3.2§</td>
</tr>
<tr>
<td>Leg axis at peak force</td>
<td>-11.9 ± 2.6</td>
<td>-11.2 ± 2.1</td>
</tr>
<tr>
<td>Leg axis at peak braking force</td>
<td>-10.7 ± 2.3</td>
<td>-11.2 ± 1.0</td>
</tr>
<tr>
<td>Leg axis at peak propulsive force</td>
<td>20.9 ± 1.9</td>
<td>20.1 ± 2.4</td>
</tr>
<tr>
<td>Difference in angles at peak force</td>
<td>8.2 ± 3.3</td>
<td>7.3 ± 2.9</td>
</tr>
<tr>
<td>Difference in angles at peak braking force</td>
<td>1.0 ± 2.6</td>
<td>0.9 ± 1.8</td>
</tr>
<tr>
<td>Difference in angles at peak propulsive force</td>
<td>9.4 ± 2.1</td>
<td>6.9 ± 4.2§</td>
</tr>
</tbody>
</table>

Positive values represent when the vector was angled in the direction of the run, in front of the vertical. Negative values represent when the vector was angled behind the vertical. * Significantly different between pre- and post- (P < 0.05). § Denotes which variables show similar trends to the significant result observed in the opposite leg.

During propulsion, when greater force was generated (Table 6.2), the runners positioned their GRFr 64.6% more horizontally after 10wkRP than before (Figure 6.2). This is likely to have resulted from the leg being more flexed at TO, so a greater proportion of the force is directed forwards rather than upwards. Chang et al., (2000) argued that by aligning the GRFr with the leg axis there might be a lower metabolic cost to running. The current results support this
A hypothesis providing empirical evidence that shows runners adjusted the alignment of their GRFr and leg axis as they became more economical runners (Figure 6.2). However Chang and colleagues (2000) measured only the GRF and hypothesised that it was aligned with the leg axis. They found that even though the magnitude of GRFr varied, the GRFr angle remained constant across different inertial and gravitational force conditions. This was postulated to contribute to smaller moment arms. This study has shown that with increased running experience runners altered their gait to reduce the difference between the angle of the GRFr and leg axis. Results also show a tendency for the peak ankle and knee joint moments in the sagittal plane to decrease with improved alignment of the leg axis and GRFr. Based on the kinetic and $\dot{V}O_2$ data it appears that it is the alignment of the leg axis and GRFr during the push-off phase of stance which contributes to lowering the metabolic cost of running and not that during braking or peak GRFr.

Although ankle stiffness appeared to decrease with increasing running experience, this was non-significant (6.0 and 19.4%, right and left ankle respectively). During running, stiffness is believed to help stabilise the lower limb, however high levels of stiffness have been linked to injury (McClay, 2000). The tendency for both joint moments and stiffness to decrease suggest that with experience runners may choose to lower joint loading and use the greater mechanical stiffness of the MTU (through greater calf inflexibility), to stabilise their lower limb during running. The greater calf inflexibility could have resulted from the extra eccentric loading and stretching being performed by the calf.
MTU, which has been associated with an improved RE (Gajdosik & Riggin, 2005). Although the efficiency of the SSC can only be hypothesised from the current study, future work incorporating muscle activity data could provide empirical evidence supporting this hypothesis.

It is also important to note that the ankle stiffness from both pre- and post-measurements were much higher than results from previous running literature using faster speeds (Günther & Blickhan, 2002; Rubenson, Henry, Dimoulas, & Marsh, 2006). The only reported results which show similar ankle stiffness’s are those by Arampatzis, Brüggemann and Metzler (1999), however they used a different formula for their stiffness calculation and so are not directly comparable. Therefore, these current values contradict reports that stiffness is expected to be greater when running at faster speeds (Günther & Blickhan, 2002; Stefanyshyn & Nigg, 1998). The reason for the particularly high stiffness values is not known, but mathematically it appears to be due to the small changes in ankle angle that ranged from between 6.7 to 11°. This suggests that the beginner runners either had high levels of dorsiflexion at TD or low peak dorsiflexion values, or a combination of both, which resulted in a low $\Delta$angle. Theoretically it is conceivable that due to the slow test speed used a small spring-like action, requiring less compression, occurred during stance so the ankle only required a limited range of motion to contribute to the compression of the leg. Further work into joint stiffness and body kinematics is required.

In summary, it appears that after 10 weeks of running the beginner runners have developed their running gait through a process of self-optimisation. This has been achieved through aligning their leg axis with the GRFr during propulsion. There is also a tendency to reduce joint loading and ankle stiffness. These adaptations are likely to have beneficial effects upon their economy of movement and injury prevention.
Chapter 7 Relationship between metabolic cost and muscular coactivation across running speeds


7.1 Introduction

Muscular coactivation, or cocontraction, concerns the simultaneous contraction of a pair of muscles. It has been argued that such muscular coordination can help stabilise a joint during locomotion (Lewek, Ramsey, Snyder-Mackler, & Rudolph, 2005). Stability produced in this way can contribute to increased stiffness in the lower limb during dynamic movements (Hortobagyi & DeVita, 2000).

Biomechanically, coactivation has been proposed as a metabolically efficient muscular coordination during running (2008). It is suggested that coactivation can make a runner’s storage of elastic energy more efficient. For example, Heise et al. (2008) reported lower oxygen consumption to be related to greater coactivation between the rectus femoris and gastrocnemius, during the stance phase of running, for female runners when performing at self-selected speeds. Both these muscles are biarticular, meaning they cross two joints, Heise et al. (2008) found that this coactivation across multiple joints had a stronger relationship with metabolic cost of running (Cr) than did activation of a single muscle. They concluded by suggesting that this activation strategy may decrease Cr. However, this suggestion is only partially supported by their earlier findings examining coactivation, which demonstrated similar relationships that did not attain significance (Heise, et al., 1996).

On the other hand, physiologically it has been argued that coactivation is an inefficient process that actually increases the metabolic cost of dynamic movement (Frost, Bar-Or, Dowling, & Dyson, 2002; Frost, et al., 1997; Mian, et al., 2006). For example, studies utilising standardised speeds of locomotion have reported that greater coactivation in the lower limbs contributes to a higher metabolic cost for elderly individuals whilst walking compared to younger individuals (Mian, et al., 2006). Furthermore, Frost and colleagues (Frost, et al.,
2002; Frost, et al., 1997) reported greater coactivation to be associated with
higher metabolic rates for both walking and running in children. Whilst they
investigated a variety of speeds, the significant differences in coactivation
between age groups occurred at the fastest walking and running speeds (Frost,
et al., 1997). Additionally they found coactivation to be an important predictor of
the metabolic rate of both walking and running (Frost, et al., 2002). Interestingly
they suggested that the younger children, who employed greater coactivation
than the older children, did so for stability purposes despite this making them
less metabolically efficient.

There is discrepancy not just between walking (Mian, et al., 2006; Peterson &
Martin, 2010) and running (Heise, et al., 2008) studies in adults, but also
between running investigations in children (Frost, et al., 2002; Frost, et al.,
1997) and adults (Heise, et al., 2008) regarding whether muscular coactivation
is a metabolically beneficial or a detrimental strategy with the potential to either
enhance or impair running performance. Furthermore whilst walking at
increasing speed results in greater coactivation in adults (Peterson & Martin,
2010), the effect of running at greater speeds on coactivation in adults is not yet
known.

The aim of this study was, therefore, to determine coactivation across different,
standardised running speeds and assess their relationship with Cr. It was
hypothesised that coactivation and Cr would be positively related to one
another, such that greater coactivation is associated with a greater Cr (i.e.
higher oxygen cost), and that greater coactivation would occur at faster speeds.
Additionally, the reliability of each coactivation was analysed by quantifying the
inter-day variability.

7.2 Methods

Eleven female recreational runners (age: 21.8 ± 2.9 yrs; mass: 60.4 ± 6.6 kg;
height: 164.8 ± 4.2 cm) took part in the study. All had a minimum of two years
running experience. Before participation all participants provided informed
consent and declared themselves to be free from injury. Testing took place
during two laboratory visits four days apart. The same protocol was used during
both testing sessions. Participants wore their own running shoes throughout testing to remove possible gait alterations that may occur whilst adjusting to different running shoes. Ethical approval was obtained from the Sport and Health Sciences Ethics Committee, University of Exeter.

A familiarisation run on the treadmill was performed before the Cr and EMG data were collected. This was performed for a minimum of 6 minutes to enable a natural running style to be achieved (Lavcanska, et al., 2005). Additionally, it served as the participants' warm-up during the first visit. A similar run was performed during the second visit as a warm-up. The measurements were made while participants ran on a level treadmill at three test speeds in the following order: 152 m-min⁻¹ (speed 1), 183 m-min⁻¹ (speed 2) and 200 m-min⁻¹ (speed 3). Participants were instructed to run at each speed for six minutes, with 10 minute rest periods between consecutive running bouts. EMG data were collected for 20 seconds towards the end of the 5th minute of each speed. Twenty consecutive strides during this 20 second period were used in the analysis. The oxygen consumption data were recorded during the final two minutes of running at each speed.

Surface EMG (Trigno Wireless EMG, Delsys, Boston, MA, USA; parallel bar configuration, contact material 99.9% Ag, interelectrode spacing 10 mm, electrode size 37 x 26 x 15 mm) was used to analyse the activation and activity of six lower limb muscles: rectus femoris (RF); vastus lateralis (VL); biceps femoris (BF); gastrocnemius lateralis (GL); and tibialis anterior (TA). The electrodes were placed longitudinally with respect to the muscle fibre direction following standardised criteria recommended by SENIAM (Hermens, et al., 2000). The skin surface area was prepared using an abrasive gel and then wiped clean with an alcohol swab. The electrodes were affixed to the lower limb and permanent marker pen used to outline their placement. This outline was kept on the participant’s leg until the next testing session so the electrodes could be positioned in the same location on the second visit. Tight shorts and self-adhesive elastic bandage covered the electrodes to minimise their movement.
The raw EMG signal was amplified and band-pass filtered (20-500 Hz) within the Delsys hardware and recorded at a sampling rate of 4000 Hz and a gain of 1000 times. A personal computer was used for off-line analysis and the storage of data. First the data underwent full-wave rectification and then a linear envelope of the EMG signal was created using the Root Mean Square (RMS). The RMS of the EMG (EMG\textsubscript{RMS}) was calculated using a 50 ms sliding window.

The duration of coactivation was calculated in a similar manner to the previous work of Heise et al. (2008) and has also been used in walking studies (Lamontagne, Richards, & Malouin, 2000; Mian, et al., 2006). Specifically, it was temporally quantified as the common duration during stance of muscle on-time between pairs of muscles. This was then recorded as a percentage of stance. The common duration time for each step was divided by the ground contact time for that respective stance period. In total seven muscle pairs were considered; three flexor-extensor (RFBF, VLBF and GLTA), one extensor-extensor (RFGL) and one flexor-flexor (BFTA) pairings. These latter two pairing groups were chosen to examine the relationship of muscular coordination and Cr. Peak EMG\textsubscript{RMS} during the stance phase of 20 consecutive steps for each of the muscles was identified and the mean peak of each muscle calculated. Data were then normalised to the peak EMG\textsubscript{RMS} of each particular speed and cut-off thresholds applied to the normalised data to identify the onset and offset of muscular activation. These were determined through a customised MATLAB (Math Works Inc., Cambridge, MA, USA) script that identified thresholds for specific muscles. Activation had to exceed and be sustained above a certain amplitude threshold for at least 50 ms. Threshold determination followed a similar procedure to that previously used by Steele and Brown (1999), whereby thresholds from 3 to 15 % of peak muscle activity were computed and compared to manually derived thresholds. Following this, the following thresholds were chosen: RF 7%, VL 7%, BF 20%, TA 12%, and GL 7% of the mean peak EMG\textsubscript{RMS} (Figure 7.1).
Figure 7.1. Sample EMG data from: Rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), gastrocnemius lateralis (GL) and tibialis anterior (TA). The dashed grey line represents TD and TO, signifying stance. The dashed black line represents the magnitude threshold for each muscle.

Stance was determined using the triaxial accelerometer integrated in the surface electrode. One electrode was affixed to the right heel of the participant's running shoe specifically for stance detection. This removed any effect of skin movement that may have resulted from using the electrode positioned on the TA. The vertical component was used to identify touch-down and toe-off. This
approach was validated separately by simultaneously collecting force and accelerometer data whilst running over a force plate and wearing the surface electrodes. This identified the period of accelerometer data that related to stance phase.

To measure oxygen consumption, participants were fitted with respiratory apparatus. Breath-by-breath respiratory gas exchange and ventilation were measured with an automated gas analysis system with the mean values displayed every 10 s (Cortex Metalyser II, Biophysik, Leipzig, Germany). Additionally, heart rate was measured via a wireless chest strap telemetry system (Polar Electro T31, Kempele, Finland) and recorded on the 4th, 5th and 6th minute of each run. Mean oxygen consumption during the final 2 minutes of running was calculated to represent \( \text{Cr} \) and mean heart rate was determined from the three measurements over the final 2 minutes.

Results from day 1 were compared to day 2 to analyse day-to-day reliability across each speed separately, using intra-class correlation coefficients (ICC) \( (2, k) \) (Weir, 2005) and the standard error of measurement (SEM). Reliability was classed as strong (ICC > 0.80) or moderate (ICC = 0.60-0.80) (Sleivert & Wenger, 1994). Precision was expressed using the SEM value, both in absolute and relative (100*(SEM of variable/mean of variable)) terms. Non-parametric tests were used in further analysis of data from day 2. To determine the differences in coactivation (relative and absolute terms), individual muscle on-times (relative and absolute terms) and stance time over the three speeds a Friedman’s ANOVA was used. Post hoc analysis was conducted using Wilcoxon sign-rank tests. Three separate Spearman’s rank correlations were performed to assess the relationship between coactivations and \( \text{Cr} \), one for each speed used. Significance level was set at \( p \leq 0.05 \). Data analysis was conducted using PASW statistics version 18 (SPSS Inc., Chicago, Il).

7.3 Results

The mean \( \text{Cr} \) values at speed 1, 2 and 3 were 200 (26.4), 188 (29.9) and 184 (20.2) mL·kg\(^{-1}\)·km\(^{-1}\) for day 1 and 197 (37.0), 184 (24.1) and 181 (19.8) mL·kg\(^{-1}\)·km\(^{-1}\) for day 2 respectively. \( \text{Cr} \) values were not significantly different
across each speed on either day. The mean heart rates at speed 1, 2 and 3 were 163 (13), 173 (16) and 183 (15) beats·min$^{-1}$ for day 1 and 162 (18), 176 (14) and 184 (13) beats·min$^{-1}$ for day 2.

Each muscle pair had a relative precision < 20% for at least one speed, except BFTA. Speed 3 elicited the least amount of precision, with the relative SEM for each muscle > 20%. RFBF during speed 2 was classed as having strong reliability (ICC > 0.80). Moderate reliability (ICC = 0.60-0.80) was shown during speed 1 for RFBF, GLTA and RFGL, during speed 2 for the RFGL, and during speed 3 for RFGL (Appendix E).

Running speed had a significant effect on the relative coactivation of the GLTA ($\chi^2(2) = 18.18, p < .001$) and BFTA ($\chi^2(2) = 6.19, p = .045$). Post-hoc analysis revealed that a higher percentage of coactivation was found at the slower speed than at the fastest speed for both muscle pairings (Figure 7.2).

Additionally, running speed significantly affected the absolute coactivations of all pairings, except RFGL, and also effected all absolute individual muscle activations (Appendix F). Only the relative activation of the TA was significantly affected by running speed ($\chi^2(2) = 12.18, p = 0.02$), with post-hoc analysis revealing speed 1 to be greater than speed 2 and speed 3 (by 6 and 10%...
respectively). Furthermore the TA was active for 4% longer during speed 2 than speed 3, but this was not significant (p = 0.06) (Appendix G). Stance time was also significantly different across speeds ($\chi^2(2) = 22.0$, $p < .001$). As speed increased, stance time decreased from 302.6 (18.1) ms to 247.6 (8.4) ms and 213.3 (11.1) ms (speed 1, 2 and 3 respectively).

The correlational analysis revealed six significant muscle pairs to be associated with Cr across the three speeds (Table 7.1). All the significant relationships were positive (range of r values: .627 - .691), meaning a higher level of coactivation is associated with greater oxygen consumption (higher Cr).

**Table 7.1 Spearman rank correlations between muscle coactivation pairs and metabolic cost of running across speeds**

<table>
<thead>
<tr>
<th>Speed</th>
<th>Muscle pair</th>
<th>Spearman’s rank</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>RFBF</td>
<td>.627</td>
<td>.039</td>
</tr>
<tr>
<td></td>
<td>VLBF</td>
<td>.636</td>
<td>.035</td>
</tr>
<tr>
<td>2</td>
<td>RFBF</td>
<td>.691</td>
<td>.019</td>
</tr>
<tr>
<td></td>
<td>VLBF</td>
<td>.682</td>
<td>.021</td>
</tr>
<tr>
<td></td>
<td>RFGL</td>
<td>.682</td>
<td>.029</td>
</tr>
<tr>
<td>3</td>
<td>RFGL</td>
<td>.627</td>
<td>.039</td>
</tr>
</tbody>
</table>

RFBF = rectus femoris-biceps femoris; VLBF = vastus lateralis–biceps femoris; RFGL = rectus femoris–gastrocnemius lateralis.

### 7.4 Discussion

The aim of this study was to investigate the metabolic cost and effect of speed on coactivation during running. The findings show that Cr was positively related to coactivation meaning that longer coactivation of the proximal and leg extensor muscles was associated with higher rates of oxygen consumption. Furthermore there was a speed effect upon coactivations. Specifically, as speed increased, coactivation of the distal and leg flexor muscles became shorter in duration.
The precise roles that different muscular coactivations play during stance have not been investigated. However based on previous suggestions, it is likely that the thigh (proximal) coactivations predominantly act during the loading phase of stance, bringing the knee into flexion. Without the simultaneous contraction of the quadriceps and hamstrings, the leg would likely collapse (Montgomery, Pink, & Perry, 1994). From our results it seems that the shank (distal) coactivations help stabilise the lower limb and ankle joint during the loading phase, in addition to controlling the forward rotation of the tibia (Mann, Moran, & Dougherty, 1986). It has been argued that the extensor-extensor (RFGL) coactivation acts to stabilise multiple joints, transferring mechanical energy from the proximal joints to the distal joints (Heise, et al., 2008), a role that is also suggested in the current findings. It is possible that the flexor-flexor (BFTA) coactivation helps create an efficient lower limb impact absorption strategy which may contribute to the overall leg stiffness. Given the differing roles described, the effect of speed and its relationship with Cr, is likely to vary depending on the muscles of interest. Our results highlight those muscular pairs that influence Cr (proximal and leg extensor pairs) and those that change with speed (distal and flexor pairs).

The results of the current study show a decrease in the level of relative coactivation in the distal muscles of the lower limb (GLTA) as running speed increases, together with a decrease in the coactivation of the leg flexor muscles (BFTA). However, the relative coactivations of the extensor (RFGL) and proximal muscles (VLBF and RFBF) were unchanged across each speed. Furthermore the Cr remained unchanged across each running speed, which is consistent with previous empirical evidence (Harris, 2003; Margaria, Cerretelli, Aghemo, & Sassi, 1963). A speed effect upon coactivation of both the proximal and distal muscles of the lower limb has been reported by Peterson and Martin (2010) during walking. Yet they found that as speed increased coactivation also increased, in both the proximal and distal pairs. In contrast, the findings from the current study show that as speed increases the coactivation in the distal muscles decrease during running. Whilst the reason behind these differing strategies can only be speculated upon, it is likely that the period of time when both legs are off the ground in running, which is not evident in walking, places
different requirements on the muscle activations and recruitment. Less coactivation of the distal muscles could facilitate greater propulsion of the body, both upwards (off the ground) and forwards (in the direction of the run), as the gastrocnemius muscles plantarflex the foot (Hamner, Seth, & Delp, 2010).

Each coactivation pair that changed across speeds contained the TA muscle, and therefore it is not surprising that the activation of the TA muscle decreased with increasing speed. In fact it was the only muscle to exhibit such a change when measured as a percentage of stance. The shorter activations of the other lower limb muscles were proportionate with the shorter ground contact time. Consequently the shorter activation of the TA is the mechanism behind shorter coactivation times at the faster speeds. This supports previous work by Mann et al., (1986) who also identified that the activation duration of the TA decreases with increasing speeds. The role of the TA muscle is to control the forefoot during the initial phase of stance, essentially performing eccentric work to slow the foot’s descent (Novacheck, 1998) and together with the gastrocnemius muscles, stabilise the ankle (Mann, et al., 1986). Furthermore running at faster speeds is associated with less dorsiflexion at touchdown (Novacheck, 1998), which would reduce the requirement of the TA to act eccentrically. It possibly then performs concentric work to move the shank forwards over the foot (Dugan & Bhat, 2005; Mann, et al., 1986). By shortening the time the TA is activated for, the eccentric contraction of the biarticular gastrocnemius muscles could have a greater contribution to controlling the forward movement of the tibia, as the GL remained active for a similar period of stance regardless of speed. Therefore the gastrocnemius may be utilising energy transferred from the thigh during knee flexion to meet the mechanical energy demands. In support of this mechanical strategy is the fact that the total time spent coactivating the RF and GL did not change with speed. This suggests that the RFGL did not proportionally decrease with stance time, thus maintaining absolute coactivation time.

With such a decrease in distal coactivation, the stability of the ankle joint may be questioned at faster speeds. However previous research has shown ankle stiffness (referred to as the slope of the joint moment-angular displacement curve) to remain constant across submaximal speeds (Arampatzis, et al., 1999).
Together this suggests that stiffness, and potentially stability, can be maintained even with decreases in coactivation. This is not surprising given that the same resultant joint moment can be produced with varying levels of muscle activation.

The positive relationships found between coactivation pairs of the extensor (RFGL) and proximal (VLBF and RFBF) leg muscles and Cr suggest that greater coactivation levels are associated with higher oxygen consumption (i.e. a poor Cr). This supports the study hypothesis but contradicts the work of Heise and colleagues (Heise, et al., 1996; Heise, et al., 2008) who reported higher coactivation during stance to be associated with lower oxygen consumption (i.e. a better Cr). It is possible that methodological differences are responsible for the opposing findings. Heise et al.(2008) required participants to self-select a speed based on perceived effort level, whereas participants were required to run at standardised speeds during the current study. Similar procedures have been used in other investigations (Kyrolainen, et al., 2001; Moore, Jones, & Dixon, 2012), with preference given to one or the other based on the aim of the investigation. Kuitunen et al. (2002) argued that participants’ effort levels are different at standardised speeds and this affects their neuromuscular requirements (Stirling, et al., 2011). However they did not assess Cr in relation to running mechanics. Furthermore the speeds chosen in this study were representative of the participants’ training speeds, meaning they were familiar with each test speed used. It would, therefore, be interesting to identify whether the relationship between running mechanics and Cr differ based on the procedure selected, whether it be standardising the metabolic cost (i.e. perceived effort/percentage of \( \dot{V}O_{2max} \)) or standardising the mechanical movement pattern (i.e. controlled test speed).

Biomechanically, apart from its potential role in elastic energy efficiency, providing joint stability (Lewek, et al., 2005) through coactivation will incur a metabolic cost. This is because to produce a net movement, such as knee flexion, the agonist, in this case the hamstrings, need to produce a stronger muscle force than the antagonist, the quadriceps. The magnitude of activation required to produce the movement is therefore greater than if the agonist was contracting without antagonist contraction, and subsequently the metabolic cost increases (Frost, et al., 1997; Mian, et al., 2006). The findings from the current
study confirm this hypothesis by showing that longer coactivations in the proximal and leg extensor muscles is associated with a greater metabolic cost. This suggests that the quadriceps, which were involved in each muscle pair associated with Cr, strongly influence a runner’s oxygen consumption. Such an argument has previously been reported in singular muscle contractions (Lewek, et al., 2012). Additional support comes from walking studies investigating coactivation across different age groups and running studies in children, which have found similar results to the current study (Frost, et al., 1997; Hortobagyi, et al., 2011). Therefore whilst coactivation may play a beneficial role in terms of joint stabilisation, possibly minimising injury risk, there appears to be a metabolic cost that could be detrimental to performance (Jones, 2006). Consequently runners may be able to enhance their performance, through decreasing Cr, by improving the way they stabilise and control their movements. Training to improve an individual’s dynamic postural control has been shown to reduce the amount of coactivation generated during functional balance tasks (Nagai et al., 2012). Modifying such training for running purposes could have performance implications by increasing the efficiency of a runner’s stabilisation process.

It is important to acknowledge the limitation of using a temporal rather than an amplitude quantification of coactivation. It is possible that longer time overlap could occur with relatively low muscle activation and minimal metabolic cost, meaning an amplitude based coactivation calculation could change its relationship with speed and Cr. However, results from walking investigations suggest that similar relationships are exhibited regardless of the computation used (Peterson & Martin, 2010), implying that differing methodologies may not change the direction of the relationship. Moreover, a temporal quantification was appropriated for the purposes of the current study as it allowed direct comparison with Heise and colleagues (2008).

7.5 Conclusion

Longer coactivations of the extensor and proximal leg muscles during running may be potentially detrimental to performance due to the associated higher
metabolic cost. It is likely that the reason for this is due to the energy requirements of muscular force generation to produce the net movement. Coactivation in the flexor and distal muscles decreased with faster speeds, as a result of short TA activation but stability may still be maintained.

7.6 Practical Implications

- High coactivation of proximal muscles is associated with a greater metabolic cost of running
- Shorter coactivations at faster speeds may be an efficient movement strategy employed by runners
- Training to improve a runner's stabilisation control may enhance their running efficiency
Chapter 8 Barefoot treadmill familiarisation


8.1 Introduction

Barefoot running is growing in popularity with approximately 75% of American runners reporting that they are interested in running barefoot (or in shoes mimicking barefoot running) (Rothschild, 2012). Currently research into barefoot running concerns the potential for it to enhance performance (Franz, et al., 2012; Perl, et al., 2012; Squadrone & Gallozzi, 2009) and reduce injury (Daoud, et al., 2012; Giuliani, et al., 2011). However barefoot running is also utilised as a test condition by many researchers investigating the effect of footwear, even though for many participants it is likely to be the first time they have ever run barefoot. This raises one of the methodological issues surrounding the study of barefoot running i.e. the familiarity of the participants to running barefoot. A lack of familiarity may limit the reliability of data obtained from a barefoot running condition.

Previous investigations assessing over ground or treadmill running gait fall into three categories regarding their barefoot/treadmill familiarisation procedures: 1) They fail to report whether any time was given for barefoot or treadmill familiarisation (Barnes, Wheat, & Milner, 2010; Franz, et al., 2012; Hanson, et al., 2011; Perl, et al., 2012). 2) They state practice barefoot trials (De Wit, et al., 2000; Stacoff, Nigg, Reinschmidt, van den Bogert, & Lundberg, 2000) / treadmill familiarisation (Divert, et al., 2005a; Divert, Mornieux, Baur, Mayer, & Belli, 2005b; Squadrone & Gallozzi, 2009) was performed without specifying time. 3) They report familiarisation was achieved when the participant believed they were comfortable with the condition (Dixon & McNally, 2008). Given that many studies find biomechanical differences between barefoot and shod conditions whilst running (e.g. (De Wit, et al., 2000)), it is possible that some findings may be influenced by initial adjustments made in response to the removal of footwear if inadequate familiarisation was given.
It has been argued that multiple steps need to be accumulated prior to biomechanical analysis of barefoot running (Divert, et al., 2005b), so any gait modifications precede gait assessment rather than occur during testing procedures. However, the time necessary for runners to become familiar with barefoot running on a treadmill, such that their running kinematics stabilise to an acceptable level during a testing session (Lavcanska, et al., 2005; Schieb, 1986), is unknown. Previous research suggested that 8-9 minutes is required for spatio-temporal adjustments whilst running shod on a treadmill (Cavanagh & Williams, 1982; Schieb, 1986). A more recent study has demonstrated that kinematic alterations can be made within 6 minutes of treadmill running (Lavcanska, et al., 2005) and that just 8 seconds is needed for kinetic familiarity (White, et al., 2002). These studies suggest the time taken for individuals to adjust to one factor, treadmill running, is within 10 minutes. By using individuals who are already familiar with treadmill running, only one factor is changed when assessing barefoot treadmill running. Furthermore barefoot running is often seen as another type of footwear condition by researchers, implying kinematic responses to adjusting to such a test condition may be similar. Therefore it is possible that the length of time required for barefoot familiarisation might be similar to shod running, however it requires specific investigation.

The aim of this study was to assess the amount of time required for habitually shod runners, with previous treadmill running experience, to become familiar with barefoot treadmill running. It was hypothesised that runners would be able to produce a consistent gait pattern within 10 minutes of running barefoot on a treadmill.

8.2 Methods

8.2.1 Participants

Twelve female recreational runners (height: 167.7 ± 6.5 cm, mass: 61.4 ± 5.5 kg, age: 24.6 ± 5.4 years) who regularly ran on treadmills volunteered for the study. All participants were free from injury at the time of testing. Only runners who had limited (less than 5 minutes) or no previous experience of barefoot running were included in the study. Thus all participants were classified as
beginner barefoot runners. Ethical approval was obtained from the University’s Sport and Health Sciences department.

8.2.2 Apparatus

An eight camera Peak Motus motion analysis system, situated in an oval shape around a treadmill was used to capture 3D kinematic data (120Hz). The system was calibrated using a wand length of 0.93 m and a fixed volume covering the treadmill belt.

A motorized treadmill (PPS 43med; Woodway, Weilam Rhein, Germany) was used during the running trials. The speed of the treadmill was checked prior to testing by recording the time taken for the treadmill belt to complete four revolutions. This was captured using a Basler camera (100 Hz), which was positioned directly in front of the treadmill, approximately 1.5 m away from the treadmill. The treadmill belt length (3.60 m) was used to calculate the speed of the treadmill belt during four revolutions. This speed was then compared to the digital display on the treadmill monitor. This was completed for each of the different speeds, ranging from 125 to 185 m·min⁻¹ (mean: 155 ± 20 m·min⁻¹). Based on the standard error of estimate there was 95% confidence that the speed of the treadmill belt was within 1.7 m·min⁻¹ of the speed displayed on the monitor.

8.2.3 Marker placement

Ten spherical reflective markers were affixed to the right lower limb of the participant using double-sided adhesive tape. A modified Soutas-Little (1987) model was used, with markers placed on the following anatomical landmarks: the proximal greater trochanter (hip); the medial and lateral condyles (knee); midline of the posterior shank; the musculotendinous junction where the medial and lateral belly of the gastrocnemius meet the Achilles tendon; the mid-tibia below the belly of the tibialis anterior; the lateral malleolus (ankle); the superior and inferior calcaneus; and the proximal head of the third metatarsal.
To determine stance a triaxial accelerometer (Trigno Wireless EMG, Delsys, Boston, MA, USA), sampling at 148 Hz, was affixed to the right heel of the participant's running shoe. The vertical component of the accelerometer data was used to detect touchdown (TD) and toe-off (TO).

8.2.4 Procedures

Each participant was instructed to self-select a speed which they felt they could comfortably run at for 30 minutes and was representative of their training speed. They performed a warm-up on the treadmill for 5 minutes at this speed whilst wearing trainers. Then they ran barefoot at this speed for 3 x 10 minutes, with 5 minute rest periods in between each bout. This amount of time was chosen based on previous treadmill familiarisation studies (Cavanagh & Williams, 1982; Lavcanska, et al., 2005; Schieb, 1986). As barefoot running could potentially cause discomfort during initial runs the protocol included rest periods to decrease the continuous time performing an unfamiliar task.

Data was captured in the first and last minute of each bout of 10 minutes, with the data being recorded during the first minute approximately 10 s after the treadmill had reached the required speed. This resulted in six time points: 1st minute (T1), 10th minute (T2), 11th minute (T3), 20th minute (T4), 21st minute (T5) and 30th minute (T6). Six complete running cycles were collected during each recording with only data during the stance period used for further analysis due to loss of data during the swing phase.

8.2.5 Data reduction

The coordinate data were smoothed within the Peak Motus system using a quintic spline smoothing technique. Further analysis occurred through a customized MATLAB script. The accelerometer data, which was simultaneously recorded alongside the kinematics, was resampled to match the kinematic data collection frequency. Sagittal plane kinematics have the greatest reliability compared to the transverse and frontal planes (McGinley, Baker, Wolfe, & Morris, 2009; Queen, Gross, & Liu, 2006). Therefore only sagittal plane movements were analysed. The hip angle was defined as the angle between
the thigh segment and the vertical line through the hip marker. The knee angle was defined between the thigh and shank segments and the ankle angle defined between the thigh and foot segments. The foot angle was defined as the angle between the ground and the vector created between the inferior calcaneus and the proximal head of the third metatarsal. In addition to the experimental data, a standing trial was recorded. This was performed in the anatomic position and the standing trial was subtracted from the experimental data to normalise each angle.

Positive values represent hip extension, knee flexion and plantarflexion at the ankle. The angles at touchdown (TD) and toe-off (TO) were calculated for the hip, knee and ankle, and foot angle at TD was used to detect foot strike patterns (Altman & Davis, 2012). Additionally, the hip angle at midstance and the peak flexion during stance for both the knee and ankle were determined. Stride length was also calculated using the following formula:

\[ SL = V \times ST \]

SL = stride length. V = speed of treadmill. ST = stride time (the time taken between successive contacts of the right foot) (Cavanagh & Williams, 1982).

8.2.6 Statistical analysis

Using means calculated for each individual’s six gait cycles at each time point, within-subject reliability for all the dependent variables was computed. First, intra-class correlation coefficients (ICC) between consecutive time points (T1-T2, T2-T3, T3-T4, T4-T5 and T5-T6) were established using the means calculated. Secondly, using the same means the standard error of means (SEM) was computed, both in absolute and relative terms. Finally, a one-way repeated measures ANOVA test was used to determine if there were any within-subject significant differences in each dependent variable across the time points. Statistical significance was set at \( p \leq 0.05 \) and all statistical tests were performed using SPSS version 19 (SPSS Inc., Chicago, IL).
8.3 Results

8.3.1 Reliability

The ICCs indicated that the highest reliability was found towards the last 10 minutes of the barefoot running time. All variables except knee flexion at TD showed strong reliability (ICC > 0.8) after 20 minutes of running. Moderate reliability (ICC: 0.6 - 0.8) was shown for all variables after 10 minutes of running barefoot. The most consistent kinematics (ICC > 0.8) throughout the whole run were: foot at TD; dorsiflexion at TD; hip at TD; hip at midstance; hip at TO and peak knee flexion. Additionally stride length was found to have the highest ICC at each time period during the 30 minutes.

8.3.2 Standard error of mean

There was a general trend for the smallest SEM, both in relative and absolute terms, to be found after 20 minutes of running. The only exceptions to this were the peak knee flexion and the hip at TD (Table 8.1), whereby the smallest SEMs were recorded during the first 10 minutes. However the relative SEMs were always below 10% for both variables, suggesting that these were the most reliable kinematics throughout the whole run.

8.3.3 Changes over time

There were four kinematic variables (out of 13) that were significantly different across time periods (Figure 8.1): dorsiflexion at TD; knee flexion at TD; knee flexion at TO; and hip at TO. Post hoc analysis revealed that there were no significant differences after T4, suggesting that the kinematic variables were stable after 20 minutes of running barefoot. No significant differences were observed in the other kinematic variables or the stride length.

8.4 Discussion

This study investigated the time required for habitually shod runners to become familiar with barefoot treadmill running. The results show familiarisation occurred within 20 minutes of running, thus contradicting the study hypothesis
that less than 10 minutes would be required. There were no significant differences in any of the biomechanical variables after 20 minutes (T1 to T4), suggesting that the runners were able to produce a consistent gait pattern following this period of time. Furthermore, all but one of the variables measured were found to have strong reliability, based on ICC values, between 20-21 minutes and 21-30 minutes. Additionally, the smallest SEMs were found during the same time periods.

Table 8.1 Absolute (relative) standard error of means (SEM) of the sagittal plane kinematics and stride length

<table>
<thead>
<tr>
<th>Variable</th>
<th>Time periods</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>T1-T2</td>
<td>T2-T3</td>
</tr>
<tr>
<td>Foot angle TD</td>
<td>1.20</td>
<td>1.82</td>
</tr>
<tr>
<td>Dorsiflexion TD</td>
<td>2.87</td>
<td>2.55</td>
</tr>
<tr>
<td>Dorsiflexion peak</td>
<td>2.33</td>
<td>4.35</td>
</tr>
<tr>
<td></td>
<td>(17.5%)</td>
<td>(32.2%)</td>
</tr>
<tr>
<td>Dorsiflexion TO</td>
<td>7.17</td>
<td>7.15</td>
</tr>
<tr>
<td>Knee flexion TD</td>
<td>3.21</td>
<td>2.00</td>
</tr>
<tr>
<td></td>
<td>(30.6%)</td>
<td>(19.5%)</td>
</tr>
<tr>
<td>Knee flexion peak</td>
<td>1.48</td>
<td>2.81</td>
</tr>
<tr>
<td></td>
<td>(4.0%)</td>
<td>(7.7%)</td>
</tr>
<tr>
<td>Knee flexion TO</td>
<td>2.34</td>
<td>1.52</td>
</tr>
<tr>
<td></td>
<td>(18.2%)</td>
<td>(12.8%)</td>
</tr>
<tr>
<td>Hip TD</td>
<td>0.59</td>
<td>0.77</td>
</tr>
<tr>
<td></td>
<td>(2.8%)</td>
<td>(3.8%)</td>
</tr>
<tr>
<td>Hip midstance</td>
<td>1.63</td>
<td>1.19</td>
</tr>
<tr>
<td></td>
<td>(13.7%)</td>
<td>(10.0%)</td>
</tr>
<tr>
<td>Hip TO</td>
<td>1.89</td>
<td>1.96</td>
</tr>
<tr>
<td></td>
<td>(10.3%)</td>
<td>(10.2%)</td>
</tr>
<tr>
<td>Stride length</td>
<td>0.04</td>
<td>0.04</td>
</tr>
<tr>
<td></td>
<td>(1.7%)</td>
<td>(1.7%)</td>
</tr>
</tbody>
</table>

*Relative standard error of mean was not calculated due to the variation in kinematic values around zero.
T1 = 1st minute. T2 = 10th minute. T3 = 11th minute. T4 = 20th minute. T5 = 21st minute. T6 = 30th minute.
Previous studies have reported that less time is required to become familiar to shod treadmill running, in the region of 6-9 minutes (Cavanagh & Williams, 1982; Lavcanska, et al., 2005; Schieb, 1986). However it is likely that the participants in these studies were habitual shod runners, meaning they only had to adjust only to the movement of the treadmill. The current study suggests that adjusting to the lack of footwear requires more time and is perhaps more complex than adjusting to the movement of a treadmill. The results also highlight that researchers need to give participants appropriate familiarisation time before using barefoot running as a test condition. This is due to the initial adjustments that participants may be making to the lack of footwear, which for most is an unfamiliar feeling.

![Figure 8.1 Kinematic changes over time. a) Ankle at TD. b) Knee at TD. c) Knee at TO. d) Hip at TO. TD = touchdown. TO = toe-off.](image)

Part of this unfamiliar feeling when running barefoot stems from the heightened proprioception that runners feel due to the lack of an external cushioning layer (Lieberman, et al., 2010; Robbins & Hanna, 1987; Robbins, et al., 1988). Such a layer insulates the foot from its own sensory feedback that helps govern the impact during ground contact (Robbins, et al., 1989; Robbins & Hanna, 1987). It is argued that gait adjustments made during barefoot running attenuate mechanical stresses placed upon the feet (Robbins & Hanna, 1987), but the current findings suggest that such modifications to a runner’s gait are not
instantaneous. Based on this, adequate familiarisation of 20 minutes should be given to habitually shod runners prior to barefoot treadmill.

The variation (represented by the SD), particularly at the ankle angle during initial ground contact, could suggest that even though the mean for each kinematic adjustment tended to plateau between 20 and 30 minutes (T4 and T6), there was still large intra-individual variation during this time period. However Figure 8.2 indicates that this is not the case. The variation demonstrated was a result of large inter-individual differences in ankle angle at TD, rather than intra-individual differences. This supports the conclusion that familiarisation occurred within 20 minutes of running BFT.

![Figure 8.2](image)

**Figure 8.2** Individual ankle angles at TD across each time point (grey lines). The mean values for each time point is represented by the black line (± SD).

As well as providing evidence regarding the time taken to adjust to barefoot running, the current study highlights some interesting specific gait adjustments made from the first minute to the 20th minute. Firstly, runners adopted less plantarflexion following the 20th minute familiarisation (2.86 vs. -0.61°, T1 vs. T4 respectively). Initially nine runners had at least 1° or more of plantarflexion at TD compared to after 20 minutes when only three runners exhibited plantarflexion. This suggests that some of the previously reported TD ankle angles (De Wit, et al., 2000; Squadrone & Gallozzi, 2009) could be a result of unfamiliarity to barefoot running. For example De Wit et al. (2000) found runners produced a significantly more plantarflexed foot when barefoot compared to shod. They argued such gait alterations reduced high loads at the heel by increasing the contact area of the heel through a flatter foot at impact.
However the current study has demonstrated that this was a natural response to running barefoot for the first time and is a result of inadequate familiarisation. Furthermore the foot angle did not change during the familiarisation period, therefore in contrast to De Wit and colleagues (2000), this suggests that there was no increase in contact area to disperse the impact load. Other kinematic changes could help explain the cushioning characteristics of barefoot running.

The initial average foot angle during familiarisation suggested that, generally, runners were midfoot striking during both the 1st (4.37°) and 20th minute (5.41°) (Altman & Davis, 2012). Based on the classification of Altman and Davis (2012), (forefoot striking: foot angle < -1.6°; rearfoot striking: foot angle > 8°; midfoot striking: -1.6° < foot angle < 8°) there were 3 forefoot strikers, 5 midfoot strikers and 4 rearfoot strikers. In light of the change in ankle angle and unchanged foot angle, the tibia would need to be rotated further forward after the 20th minute, rather than the foot being placed flatter to the ground. This tibial movement would explain the greater knee flexion recorded at TD with increased running familiarity, consistent with the hip angle at TD being similar across each time point (Figure 8.3). Previous research has reported greater knee flexion at TD when running barefoot compared to running shod (de Koning & Nigg, 1994; De Wit, et al., 2000). However, the current findings suggest adequate familiarisation allows runners to produce greater knee flexion at TD meaning

![Figure 8.3 Leg geometry at touchdown. a) During the first minute. b) During the 20th minute. Angles enclosed by a double lined curve remained constant over time. Angles enclosed with a single curved line changed over time.](image)
previous differences found may be smaller than what could have been achieved with familiarisation. Furthermore De Wit and colleagues calculated that 96% of the variance in foot angle at TD could be determined by the ankle angle and shank angle during barefoot running (De Wit, et al., 2000), showing how intrinsically linked these positional angles are. Therefore, it appears that with increased familiarity runners utilise the knee to a greater degree to help cushion the impact by reducing their effective mass (Derrick, 2004). By adopting a more flexed knee at TD the magnitude of impact force experienced could be reduced (Gerritsen, van den Bogert, & Nigg, 1995), possibly reducing the likelihood of injury (Derrick, 2004). So rather than increasing the amount of contact area to lower the loads experienced, runners tended to change their knee and shank positions to facilitate the reduction of impact force.

Stride length was the most reliable gait characteristic with little variation over time, meaning runners adjusted their stride length almost instantaneously at the beginning of the run. Therefore it is likely that the shorter stride lengths reported during barefoot running (De Wit, et al., 2000; Squadrone & Gallozzi, 2009) may be an anticipatory strategy, such as that used when adjusting leg stiffness in response to changes in surface (Ferris, Liang, & Farley, 1999). This strategy would be controlled by visual cues of the surface, and knowledge of the surface properties from previous experiences (Ferris, et al., 1999) which may heighten proprioception of the surface prior to running on it. Previous results have shown that even a small layer between the foot and the surface that lessens proprioception, such as a minimalist shoe, enables runners to choose a similar stride length to that demonstrated during shod running (Squadrone & Gallozzi, 2009). For such a stride length to be consistently reproducible during shod running on a treadmill may take between 2-4 minutes (Lavcanska, et al., 2005). Conversely by removing the external layers that insulate the foot from impact with the ground, runners are able to adopt comfortable stride lengths almost immediately.

Due to this heightened proprioception when running barefoot the interaction between the surface and the foot will play a greater role in determining the running mechanics of an individual. Elements known to affect a runner’s gait, such as surface stiffness (Dixon, Collop, & Batt, 2000; Ferris, et al., 1999),
could influence the time to familiarisation. The same treadmill was used throughout testing to minimise the effect the surface could have on time to familiarisation, but caution should be exercised when generalising these findings to over ground running with different surface properties. Nevertheless, the results support the argument made by Divert et al (2005b), that multiple steps need to be accumulated prior to assessing the biomechanics of barefoot running. Therefore it is not unreasonable to suggest that numerous practice trials should be given in barefoot over ground running conditions prior to experimental testing. However, further research is needed to assess the time/number of trials required.

It is possible that familiarisation may have occurred sooner than 20 minutes if no rest period was given. However such a protocol was deemed necessary following pilot work, which tested 30 minutes of continuous running and found this caused soreness in the lower limb during and post-exercise. Additionally, familiarisation could have occurred at any point between 11 and 20 minutes. However, due to data being collected at the beginning and end of each bout, the exact time of familiarisation cannot be identified. Further investigations, which record data more frequently, are needed to ascertain the exact minute adequate familiarisation was achieved.

8.5 Conclusion

In conclusion, to familiarise habitually shod, experienced treadmill runners to barefoot treadmill running requires a minimum of 20 minutes of running on a treadmill. Kinematic and spatio-temporal measures were consistent and stable within 20 minutes, suggesting that future studies should include a sufficient period of familiarisation to barefoot running prior to commencing experimentation. After familiarisation, runners adopted less plantarflexion and greater knee flexion during initial ground contact. However stride length changes during barefoot running were adopted immediately.
9.1 Introduction

Running barefoot (BFT) is not a new concept for either runners or scientists. Research in the late 80s first suggested that running BFT could have potential benefits in terms of reducing the risk of injury (Robbins & Hanna, 1987). Moreover, elite runners, such as Abebe Bikila and Zola Budd, have broken world records in the Marathon and 5000m respectively, suggesting there is a potential performance benefit to running BFT. Recently, much attention had been given to the question of whether BFT running is more economical than running in trainers, or shod (SH) (Franz, et al., 2012; Squadrone & Gallozzi, 2009).

There currently is no consensus in the literature regarding the affect that BFT running has on running economy (RE) (Nigg & Enders, 2013), a crucial factor in determining performance in long distance running events (Conley & Krahenbuhl, 1980). The use of differing methodologies is perhaps, in part, the reason for contrasting findings. For example, Divert et al. (2008) reported that oxygen consumption ( $\dot{V}O_2$ ) was significantly lower when running BFT compared to SH. However, this was argued to be a result of the mass effect of the shoe. Based on this, researchers have adjusted for the ‘mass effect’ by attaching equivalent masses to the feet when wearing socks, simulating BFT running (Franz, et al., 2012). This aids the control of the confounding variable, yet detracts from the essence of running BFT, which should essentially be running with nothing on/attached to your feet. Furthermore kinematic adjustments to BFT running, which have been extensively reported (De Wit, et al., 2000; Squadrone & Gallozzi, 2009), are likely to be affected by the extra mass being carried on the foot. Adjusting the absolute $\dot{V}O_2$ to not just a
runner's mass, but also the difference in footwear mass provides an approach for mathematical shoe mass adjustment without potentially affecting running mechanics.

One of the main gait parameters that changes when running BFT is stride length. Previously researchers have found the naturally chosen BFT stride length to be 3% shorter than the naturally chosen SH stride length (Franz, et al., 2012). Based on evidence that the naturally chosen stride length is at or near economically optimal for runners (Cavanagh & Williams, 1982), researchers have argued that stride length plays an important role in determining RE during BFT running (Perl, et al., 2012). However, previous investigations have either not controlled for stride length (Franz, et al., 2012; Squadrone & Gallozzi, 2009) or have instructed runners to run with their SH stride length whilst BFT (Perl, et al., 2012). This latter approach would also affect the BFT running mechanics due to running with an unnaturally longer stride.

Another factor influencing any observed differences between BFT and SH running is the level of proprioception experienced. SH running offers a cushioning protective layer between the foot and the ground that attenuates proprioception. BFT running removes this layer, thus heightening the proprioception experienced (Robbins & Hanna, 1987). Minimalist shoes (MS) also lack a cushioning layer, but do include a thin protective layer. This reduces the potential benefit of increased proprioception provided when BFT, but limits cushioning in a similar manner to BFT. Thus, MS should have a comparable level of proprioception as SH, but a different level of cushioning. Therefore comparison of SH, MS and BFT allows controlled variation of cushioning and proprioception to determine the effect that either and/or both have on RE and running mechanics.

The pursuit of improved running performance through changes in running mechanics is also likely to affect a runner's risk of sustaining an injury. Whilst BFT gait adjustments have been reported to reduce injury likelihood (Lieberman, et al., 2010; Robbins & Hanna, 1987), there is no conclusive evidence to support this notion (Nigg & Enders, 2013). In fact, there is some
evidence of an increase in peak tibial acceleration when running BFT compared with SH (Sinclair, et al., 2013a). On the other hand, the lack of a heel lift and cushioning experienced when BFT, in addition to heightened proprioception, are believed to promote a forefoot strike pattern (Lieberman, et al., 2010). Such a foot strike modality has been associated with reduced impact force (Lieberman, et al., 2010) and injury rate (Daoud, et al., 2012) in comparison to rearfoot striking. Yet RE is similar between the two strike patterns (Perl, et al., 2012). Together these findings suggest changes in running mechanics that potentially reduce injury likelihood are neither beneficial nor detrimental to performance. Consideration of biomechanical changes for both their performance and injury implications is needed to generate greater understanding of the differences between BFT, MS and SH.

It has also been argued that there is a metabolic cost to cushioning the body during BFT running due to muscles having to actively protect the lower limb upon impact and during ground contact. This is known as the ‘cost of cushioning’ hypothesis (Frederick, 1984). Due to the lack of external cushioning present in minimalist footwear, it is likely that this extra metabolic cost will apply to MS running too. Contrary to this expectation, recent evidence has shown that transitioning to running MS can improve RE (Warne & Warrington, 2012), and thus performance. Yet it can also increase bone marrow edema (Ridge, et al., 2013), which is an early sign of bone stress injury. Researchers must therefore consider the affect that any changes in gait may have on both performance and injury mechanisms before being able to justify advocating either BFT, MS or SH running to individuals.

There is a relative scarcity of studies that have assessed muscular activity during BFT/MS running. One notable exception identified that the tibialis anterior (TA) activity was adjusted to suit external conditions, such as changes in cushioning (von Tscharner, et al., 2003). These authors reported greater activity in the TA during ground contact, but less activity just prior to touchdown whilst running BFT compared to SH. Although Lieberman et al. (Lieberman, et al., 2010) did not record muscular activity, they did calculate a low level of ankle stiffness to be present during BFT running (with a forefoot strike pattern). This suggests a low level of coactivation, which can be beneficial to the
metabolic cost of running (Moore et al., under review). To fully investigate the impact of changing cushioning and proprioception during running, muscular activity should also be considered. This would also allow the ‘cost of cushioning’ hypothesis, first proposed by Frederick (1984), to be evaluated.

This study aimed to assess the mechanisms behind changes in RE measured at different stride lengths when varying cushioning and proprioception. This was achieved by comparing SH, MS and BFT RE, biomechanics and EMG. Since the focus was on mechanisms, specific hypotheses concerning kinematics and muscular activity/coactivity were not constructed. Regarding performance benefits and injury concerns it was hypothesised that: 1) BFT running with a BFT stride length will be the most economical condition. 2) MS running will be more economical than SH running, but when MS running with a SH stride length will be more economical than when running with a BFT stride length. And secondly regarding acceleration, it was hypothesized that: 3) BFT and MS running would have greater impact accelerations than SH running.

9.2 Methods

Sample size was determined using a similar methodology to Franz et al., (2012) whereby a 1-2% mean difference in RE across conditions can be established with 10-15 subjects. Therefore, fifteen female (mass: 62.0 ± 6.4 kg, height: 1.66 ± 0.1 m, age: 20.5 ± 1.4 yrs), habitually shod, recreational runners volunteered and were recruited for the study. Each participant provided informed consent, was free from cardiac abnormalities and free from injury for at least three months prior to testing. Ethical approval was granted from the Ethics Committee of the Sport and Health Sciences department at the University of Exeter.

9.2.1 Procedure

Two laboratory visits were completed by each participant. During the first visit, participants underwent familiarisation to running BFT and SH on a motorised treadmill (PPS 43med; Woodway, Weilam Rhein, Germany). The experimental procedure was undertaken during the second visit. Movement analysis and
muscle activity of the right leg were simultaneously recorded, along with oxygen consumption.

9.2.2 Treadmill familiarisation

Based on results from our laboratory and literature evidence, 20 minutes (2 x 10 minutes) was given for BFT running and 6 minutes for SH (Lavcanska, et al., 2005). At the end of each familiarisation period the participant’s natural stride length was recorded using a Basler camera (100 Hz) positioned approximately 1.5 m in front of the treadmill. To determine each stride length the following equation was used: \( SL = ST \times V \), where \( SL \) is the stride length, \( ST \) is the time taken for each stride (right foot contact to right foot contact) and \( V \) represents the treadmill velocity. Six strides were recorded during each familiarisation condition and the average stride length was calculated for both BFT and SH running.

9.2.3 Experimental procedure

Participants ran at each stride length (BFT and SH) during each footwear condition (BFT, MS and SH), performed in a randomised order to reduce fatigue and learning effects. Each run was performed at 167 m·min\(^{-1}\) for six minutes with 10 minute rest periods between each bout. The rest period was used to change over footwear and check the attachment of kinematic markers. A metronome was used to control stride length, with participants instructed to strike the treadmill in time with each beat.

The treadmill used for the experimental procedure was the same as that used for the familiarisation. Heart rate was recorded during the final two minutes of each bout via a wireless chest strap telemetry system (Polar Electro T31; Kempele, Finland). Respiratory gas exchange and ventilation was measured using an automated gas analysis system (Cortex Metalyzer II; Cortex Biophysik, Leipzig, Germany), with mean values displayed every 10 s.

Three-dimensional kinematic data were collected using an eight camera optical system (Vicon Peak, 120 Hz, automatic optoelectronic system; Peak Performance Technologies, Inc., Englewood, CO) positioned around the treadmill. Synchronisation of the kinematic and EMG data occurred via a manual
trigger pressed during the final two minutes of each run. Spherical reflective markers were affixed to the right lower limb of each participant on the following anatomical landmarks: the proximal greater trochanter (hip); the medial and lateral condyles (knee); midline of the posterior shank; the musculotendinous junction where the medial and lateral belly of the gastrocnemius meet the Achilles tendon; the mid tibia below the belly of the tibialis anterior; the lateral malleolus (ankle); the superior and inferior calcaneus and the proximal head of the third metatarsal. Additionally an accelerometer (Trigno Wireless EMG, Delsys, Boston, MA, USA, 148 Hz) was attached to the heel, in between the calcaneus markers.

Surface EMG (Trigno Wireless EMG, Delsys, Boston, MA, USA; parallel bar configuration, contact material 99.9% Ag, interelectrode spacing 10 mm, electrode size 37 x 26 x 15 mm) was used to analyse the activity of five lower limb muscles on the right leg: rectus femoris (RF); vastus lateralis (VL); biceps femoris (BF); gastrocnemius lateralis (GL); and TA. The electrodes were placed longitudinally with respect to the muscle fibre direction following standardised criteria recommended by SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles project). The skin surface area was prepared using an abrasive gel and then wiped clean with an alcohol swab. The electrodes were affixed with double-sided tape to the lower limb, with those on the shank covered with elasticated tubular bandage and those on the thigh with self-adhesive elastic bandage to minimize their movement. The kinematic markers were then affixed on top of the bandages.

9.2.4 Data analysis

RE was defined as the average \( \dot{V}O_2 \) during the final two minutes of each run. Carbon dioxide production (\( \dot{V}CO_2 \)) respiratory exchange ratio (RER) and minute ventilation (\( \dot{V}E \)) were also determined over the same time period. To account for the mass of the shoe, absolute \( \dot{V}O_2 \), \( \dot{V}CO_2 \) and \( \dot{V}E \) values were normalised to each individual’s barefoot body mass, plus the shoe mass difference. The mean mass of the minimal trainer (Vibram FiveFingers™) was 138 g (range: 122 – 149 g), and the mean mass of the traditional, neutral running shoe (Adidas) was
223 g (range: 198 – 249 g). Two separate sets of physiological values were calculated for BFT and MS conditions. One set accounted for the trainer mass to allow comparison to the SH condition and the other accounted for the minimalist trainer so the BFT condition could be compared to the MS conditions. This method, although similar to Divert et al. (2008), has not previously been utilised by researchers assessing BFT, MS and SH running mechanics. However it provides more realistic BFT and MS running conditions than adding weights to the foot. Along with increasing the leg moment of inertia, this approach may interfere with running mechanics.

Stance was defined as initial foot touchdown (TD) to toe-off (TO) and was determined using the vertical heel acceleration data. Six consecutive gait cycles were collected for each condition during the final two minutes of data collection for that condition. Participants were not aware of when data collection was taking place. Kinematic data were filtered using a fifth-order quintic spline filter within the Peak Motus system. Dynamic angles were normalised to standing trials recorded after each run to provide anatomically meaningful angles. As the focus of the study was changes to foot-surface interaction via varying degrees of cushioning and proprioception only angles recorded during ground contact were used. The hip, knee, ankle and rearfoot angles at touchdown (TD), peak (excluding hip) and toe-off (TO) were analysed. Additionally, to identify foot strike modality the foot angle at TD was calculated. The instant of heel off was determined from the kinematic data. This provided information about the time spent loading the forefoot, and thus the metatarsals, during propulsion. The final kinematic variable measured was vertical oscillation, defined by the maximum vertical displacement of the hip marker during one gait cycle.

Data from the BF electrode could not be used for two participants due to electrode movement. Within the Delsys hardware the EMG signal (sampled at 2000 Hz) was amplified and bandpass filtered (20-500 Hz). Offline analysis was performed using a customized MATLAB (Math Works Inc., Cambridge, MA, USA) script. EMG data were baseline adjusted and underwent full-wave rectification. To determine muscle on-off times the rectified data were submitted to a nonlinear Teager-Kaiser energy operator (TKEO) (Li, et al., 2007). Individual muscle on-off thresholds were determined by trialling a range of thresholds.
between 3 and 15% of the peak muscle activity during the six steps. The TKEO data had to rise and stay above each threshold for at least 50 ms. These muscle on-off times was then compared to manually derived thresholds and specific cut-offs where chosen for each muscle.

The rectified muscle activity data for each muscle was integrated (iEMG) during preactivity (100 ms prior to TD) and stance. The iEMG was then normalized to the iEMG of each muscle across five gait cycles during the SH with SH stride length. Normalisation for the coactivation calculation was similar to the above, but average EMG amplitude rather than iEMG was used. The normalised EMG data were then entered into the coactivation calculation, previously used in gait studies (Franz & Kram, 2012). A total of three agonist-antagonist pairs were analysed, two from the thigh (RFBF and VLBF) and one from the shank (GLTA), using the following equation [1]:

\[
\text{Coactivation} = 2 \times \left[ \frac{\min(EMG_1, EMG_2)}{\int EMG_{1+} + \int EMG_{2}} \right] \times 100
\]  

[1]

Where EMG\textsubscript{1} and EMG\textsubscript{2} denote the two sets of EMG data used to calculate muscular coactivation and min refers to the minimum of the two sets of EMG.

### 9.2.5 Statistical analysis

Means (± SDs) of the six strides were calculated for each variable across each condition. Paired T-tests were used to verify a steady-state \(\dot{V}O_2\) by comparing \(\dot{V}O_2\) during minutes 4-5 to minutes 5-6. All p-values were greater than 0.05 signifying no time-dependent change in \(\dot{V}O_2\). To make direct comparisons of RE across different footwear conditions and in each stride length condition, paired T-tests were used. This allowed each condition to be compared to SH with SH stride length, the habitual running condition, and enabled footwear mass and stride length to be independently tested for significance. Additionally, paired T-tests were used to compare the other physiological parameters, EMG measurements and kinematics, with statistical significance set at \(p \leq 0.05\). Cohen’s \(d\) effect sizes (ES) were also calculated for all RE comparisons.
### 9.3 Results

The mean BFT stride length was significantly shorter than the SH stride length (2.58%), indicating a significantly higher stride frequency (165 ± 7.67 v 161 ± 6.79 steps·min⁻¹, BFT stride frequency and SH stride frequency respectively). SH with a SH stride length had a similar \( \dot{V}O_2 \) to MS with a BFT stride length (ES = 0.19, \( p = 0.064 \)), but was significantly higher than MS with a SH stride length (3.1%, ES = 0.32, \( p = 0.003 \)), BFT with a SH stride length (4.9%, ES = 0.50, \( p = 0.001 \)) and BFT with a BFT stride length (6.5%, ES = 0.64, \( p = 0.002 \)). SH with a BFT stride length was similar to both MS conditions (BFT stride length: ES = 0.19, \( p = 0.075 \), SH stride length: ES = 0.25, \( p = 0.095 \)), but higher than both BFT conditions (BFT stride length: 6.4%, ES = 0.61, \( p = 0.001 \); SH stride length: 5.1%, ES = 0.48, \( p = 0.001 \)). Additionally, from the BFT conditions only \( \dot{V}O_2 \) during BFT with a BFT stride length was significantly lower than both MS conditions (BFT stride length: 4.6%, ES = 0.43, \( p = 0.004 \); SH stride length: 3.5%, ES = 0.35, \( p = 0.017 \)) (Table 9.1). However no differences were found in \( \dot{V}O_2 \) between BFT and SH stride lengths within footwear conditions. Similar findings were recorded in the \( \dot{V}CO_2 \) and \( \dot{V}E \) data, however the highest HR was observed during the SH with BFT stride length condition (Table 9.1).

There was no change in iEMG preactivation of any of the lower limb muscles. During stance, iEMG of two muscles, BF and TA, exhibited significant differences across conditions (Table 9.2). Furthermore GLTA coactivation was ~12% lower during MS with BFT stride length than BFT with BFT stride length. There were several kinematic variables that were found to change across conditions, with the lowest vertical oscillation and lowest peak dorsiflexion (Figure 9.1) and eversion (Table 9.2) recorded during BFT with BFT stride length running. However, impact acceleration was highest during both BFT conditions, with impact acceleration remaining consistent across the others (Table 9.2). There was also a trend for lower angular velocities to be reported during the BFT running conditions, especially at peak ankle and rearfoot velocity (Table 9.3).

Measurement of initial foot angle provided data on the foot strike modality employed for each condition. Based on the classification of Altman and Davis (forefoot striking: foot angle < -1.6°, rearfoot striking: foot angle > 8°, and
midfoot striking: \(-1.6^\circ \text{ < foot angle < } 8^\circ\) the greatest number of forefoot strikers was observed during BFT with BFT stride length running (47%) and the least amount of rearfoot strikers (20%). SH with SH stride length demonstrated the opposite trend, with the least number of forefoot strikers (20%) and the greatest amount of rearfoot strikers (67%) (Table 9.4). The latest occurrence of heel off was observed during SH with SH stride length running, with BFT and MS running demonstrating similar timings that were up to 22ms (~12.7%) earlier than SH with SH stride length (Table 9.2).

Table 9.1 Means (± SDs) of physiological measures, adjusted for SH and MS mass, for each condition

<table>
<thead>
<tr>
<th>Variable</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
</tr>
</thead>
<tbody>
<tr>
<td>HR (beats-min(^{-1}))</td>
<td>166 ± 15(^{E})</td>
<td>166 ± 15(^{E})</td>
<td>167 ± 13(^{E})</td>
<td>167 ± 16</td>
<td>170 ± 14(^{A,B,C})</td>
<td>168 ± 15</td>
</tr>
<tr>
<td>RE (mL·kg(^{-1})·min(^{-1}))</td>
<td>35.25 ± 3.65(^{E,F})</td>
<td>36.75 ± 3.99(^{E,F})</td>
<td>36.96 ± 4.16(^{E})</td>
<td>36.53 ± 3.64(^{F})</td>
<td>37.69 ± 3.98(^{A,B,C})</td>
<td>37.71 ± 3.74(^{A,B,D})</td>
</tr>
<tr>
<td>(\dot{V}CO_2) (mL·kg(^{-1})·min(^{-1}))</td>
<td>32.14 ± 3.88(^{E,F})</td>
<td>32.43 ± 4.24(^{E,F})</td>
<td>33.53 ± 4.57(^{E})</td>
<td>33.94 ± 4.27</td>
<td>35.03 ± 4.23(^{A,B,C})</td>
<td>34.68 ± 4.21(^{A,B})</td>
</tr>
<tr>
<td>RER</td>
<td>0.91 ± 0.04</td>
<td>0.91 ± 0.04</td>
<td>0.91 ± 0.03(^{E})</td>
<td>0.92 ± 0.04</td>
<td>0.93 ± 0.04(^{C})</td>
<td>0.92 ± 0.04</td>
</tr>
<tr>
<td>(\dot{V}E) (mL·kg(^{-1})·min(^{-1}))</td>
<td>1049.54 ± 32.52(^{D})</td>
<td>1058.90 ± 32.81(^{D})</td>
<td>1061.60 ± 33.64(^{D})</td>
<td>1073.04 ± 34.19(^{D})</td>
<td>1104.10 ± 154.60(^{A,C})</td>
<td>1102.28 ± 157.40(^{A,B,D})</td>
</tr>
</tbody>
</table>

- denotes significantly different to BFT with BFT stride length. \(^{A}\) denotes significantly different to BFT with SH stride length. \(^{B}\) denotes significantly different to MS with BFT stride length. \(^{C}\) denotes significantly different to MS with SH stride length. \(^{D}\) denotes significantly different to SH with BFT stride length. \(^{E}\) denotes significantly different to SH with SH stride length. HR = heart rate. RE = running economy. RER = respiratory exchange ratio. \(\dot{V}CO_2\) = carbon dioxide production. \(\dot{V}E\) = minute ventilation.
9.4 Discussion

In support of our first hypothesis, the most economical conditions were when running BFT; however, there was no difference between each stride length condition. The MS running was only more economical than SH with SH stride length when running with a SH stride length, which partially supports our second hypothesis. However, there was no significant difference in RE between the two MS conditions, thus partially contradicting our second hypothesis. These results are particularly interesting given the argument that runners feel more comfortable running when the energy demand is low (Nigg & Enders, 2013). Runners exhibit similar stride lengths when SH and MS (Squadrone & Gallozzi, 2009) and adopting such a running style (MS with SH stride length) was found to favour RE improvements in the current study compared to running with shorter stride lengths (MS with BFT stride length). It is conceivable therefore that runners naturally choose to keep their SH stride length when MS as the metabolic cost of running is reduced and thus feels more comfortable for the runner.

Perl et al., (2012) found a similar metabolic advantage of MS over SH running when instructing runners to maintain a SH stride length, but suggested a greater advantage may have been recorded with higher stride frequencies (shorter stride lengths). Contrastingly, our results suggest that if shorter stride lengths (i.e. a BFT stride length) were taken, Perl et al. (2012) may not have found any economical advantage. On the other hand, several studies have failed to find a significant difference in RE between BFT, MS and SH conditions (Burkett, et al., 1985; Franz, et al., 2012; Squadrone & Gallozzi, 2009). Suggested reasons behind these disparities could be: a lack of familiarisation with BFT running (Burkett, et al., 1985); added shoe mass affecting running mechanics (Franz, et al., 2012); or failure to account for differences in footwear mass (or lack of) when calculating relative $\dot{V}O_2$ (Squadrone & Gallozzi, 2009).
Table 9.2 Means (± SDs) of lower limb running mechanics, impact accelerations and muscular activity variables that were significantly different between conditions

<table>
<thead>
<tr>
<th>Variable</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Kinematics and spatiotemporal variables</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Foot angle at TD (°)</td>
<td>-3.79 ±</td>
<td>-0.25 ±</td>
<td>-1.10 ±</td>
<td>4.41 ±</td>
<td>1.69 ±</td>
<td>4.97 ±</td>
</tr>
<tr>
<td></td>
<td>16.12&lt;sup&gt;D,E,F&lt;/sup&gt;</td>
<td>20.85&lt;sup&gt;F&lt;/sup&gt;</td>
<td>17.75&lt;sup&gt;F&lt;/sup&gt;</td>
<td>13.89&lt;sup&gt;A&lt;/sup&gt;</td>
<td>19.23&lt;sup&gt;A&lt;/sup&gt;</td>
<td>17.64&lt;sup&gt;A,B,C&lt;/sup&gt;</td>
</tr>
<tr>
<td>Peak dorsiflexion (°)</td>
<td>-9.61 ±</td>
<td>-11.28 ±</td>
<td>-16.68 ±</td>
<td>-16.94 ±</td>
<td>-22.24 ±</td>
<td>-20.28 ±</td>
</tr>
<tr>
<td></td>
<td>4.14&lt;sup&gt;C,D,E,F&lt;/sup&gt;</td>
<td>4.57&lt;sup&gt;F&lt;/sup&gt;</td>
<td>5.50&lt;sup&gt;A,E&lt;/sup&gt;</td>
<td>6.00&lt;sup&gt;A&lt;/sup&gt;</td>
<td>3.50&lt;sup&gt;A,C&lt;/sup&gt;</td>
<td>5.80&lt;sup&gt;A,B&lt;/sup&gt;</td>
</tr>
<tr>
<td>Plantarflexion at TO (°)</td>
<td>11.31 ±</td>
<td>13.00 ±</td>
<td>15.29 ±</td>
<td>17.21 ±</td>
<td>16.50 ±</td>
<td>18.59 ±</td>
</tr>
<tr>
<td></td>
<td>7.17&lt;sup&gt;C,F&lt;/sup&gt;</td>
<td>7.79&lt;sup&gt;F&lt;/sup&gt;</td>
<td>9.07&lt;sup&gt;A&lt;/sup&gt;</td>
<td>6.46 ±</td>
<td>7.34 ±</td>
<td>8.49&lt;sup&gt;A,B&lt;/sup&gt;</td>
</tr>
<tr>
<td>Knee flexion at TD (°)</td>
<td>12.35 ±</td>
<td>10.35 ±</td>
<td>9.90 ±</td>
<td>8.26 ±</td>
<td>10.77 ±</td>
<td>11.35 ±</td>
</tr>
<tr>
<td></td>
<td>4.14&lt;sup&gt;XC,D&lt;/sup&gt;</td>
<td>4.52&lt;sup&gt;D&lt;/sup&gt;</td>
<td>3.46&lt;sup&gt;XA,D&lt;/sup&gt;</td>
<td>2.96&lt;sup&gt;A,B,C,F&lt;/sup&gt;</td>
<td>3.54 ±</td>
<td>2.65&lt;sup&gt;D&lt;/sup&gt;</td>
</tr>
<tr>
<td>Peak eversion (°)</td>
<td>-3.13 ±</td>
<td>-4.29 ±</td>
<td>-5.18 ±</td>
<td>-6.04 ±</td>
<td>-7.17 ±</td>
<td>-5.40 ±</td>
</tr>
<tr>
<td></td>
<td>2.82&lt;sup&gt;C,D,E,F&lt;/sup&gt;</td>
<td>3.38 ±</td>
<td>3.43&lt;sup&gt;A&lt;/sup&gt;</td>
<td>4.00&lt;sup&gt;A&lt;/sup&gt;</td>
<td>3.39&lt;sup&gt;A&lt;/sup&gt;</td>
<td>3.39&lt;sup&gt;A&lt;/sup&gt;</td>
</tr>
<tr>
<td>Heel off (ms)</td>
<td>151 ±</td>
<td>157 ±</td>
<td>155 ±</td>
<td>160 ±</td>
<td>162 ±</td>
<td>173 ±</td>
</tr>
<tr>
<td>Vertical oscillation (cm)</td>
<td>8.96 ±</td>
<td>9.22 ±</td>
<td>9.26 ±</td>
<td>9.33 ±</td>
<td>9.67 ±</td>
<td>9.61 ±</td>
</tr>
<tr>
<td>EMG and accelerometer variables</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Impact acceleration (g)</td>
<td>7.00 ±</td>
<td>6.74 ±</td>
<td>5.60 ±</td>
<td>5.70 ±</td>
<td>5.63 ±</td>
<td>5.50 ±</td>
</tr>
<tr>
<td></td>
<td>0.80&lt;sup&gt;C,D,E,F&lt;/sup&gt;</td>
<td>0.73&lt;sup&gt;C,D,E,F&lt;/sup&gt;</td>
<td>1.01&lt;sup&gt;A,B&lt;/sup&gt;</td>
<td>0.85&lt;sup&gt;A,B&lt;/sup&gt;</td>
<td>0.60&lt;sup&gt;A,B&lt;/sup&gt;</td>
<td>0.60&lt;sup&gt;A,B&lt;/sup&gt;</td>
</tr>
<tr>
<td>TA activity (a.u)</td>
<td>72.40 ±</td>
<td>69.45 ±</td>
<td>58.85 ±</td>
<td>72.69 ±</td>
<td>57.81 ±</td>
<td>56.22 ±</td>
</tr>
<tr>
<td></td>
<td>50.47&lt;sup&gt;XC,E,F&lt;/sup&gt;</td>
<td>64.16&lt;sup&gt;XE&lt;/sup&gt;</td>
<td>32.91&lt;sup&gt;XA&lt;/sup&gt;</td>
<td>52.27&lt;sup&gt;E,F&lt;/sup&gt;</td>
<td>54.99&lt;sup&gt;XA,B,D&lt;/sup&gt;</td>
<td>31.99&lt;sup&gt;A,D&lt;/sup&gt;</td>
</tr>
<tr>
<td>BF activity (a.u)</td>
<td>29.03 ±</td>
<td>40.51 ±</td>
<td>38.59 ±</td>
<td>34.89 ±</td>
<td>25.35 ±</td>
<td>28.05 ±</td>
</tr>
<tr>
<td></td>
<td>23.10&lt;sup&gt;B,C&lt;/sup&gt;</td>
<td>37.94&lt;sup&gt;A,XE&lt;/sup&gt;</td>
<td>29.61&lt;sup&gt;A,XE&lt;/sup&gt;</td>
<td>39.14 ±</td>
<td>16.47&lt;sup&gt;XB,XC&lt;/sup&gt;</td>
<td>16.97 ±</td>
</tr>
<tr>
<td>GLTA (%)</td>
<td>61.63 ±</td>
<td>54.33 ±</td>
<td>49.35 ±</td>
<td>58.22 ±</td>
<td>57.98 ±</td>
<td>55.96 ±</td>
</tr>
<tr>
<td></td>
<td>13.89&lt;sup&gt;C&lt;/sup&gt;</td>
<td>16.48 ±</td>
<td>18.23&lt;sup&gt;A&lt;/sup&gt;</td>
<td>17.42 ±</td>
<td>14.76 ±</td>
<td>13.15 ±</td>
</tr>
</tbody>
</table>

<sup>*</sup> denotes significantly different to BFT with BFT stride length. <sup>+</sup> denotes significantly different to BFT with SH stride length. <sup>C</sup> denotes significantly different to MS with BFT stride length. <sup>D</sup> denotes significantly different to MS with SH stride length. <sup>E</sup> denotes significantly different to SH with BFT stride length. <sup>F</sup> denotes significantly different to SH with SH stride length. <sup>X</sup> denotes nearing significance (p < 0.06). TD = touchdown. TO = toe-off. TA = tibialis anterior. BF = biceps femoris. GLTA = gastrocnemius lateralis – tibialis anterior.
9.4.1 Preactivity

When removing the cushioning layer (SH to BFT/MS) there were no changes to muscular preactivity in any of the muscles. It has been argued that to adjust to impact changes, such as the removal of cushioning, runners use an anticipatory strategy termed ‘muscle tuning’ to gear the leg for impact (Boyer & Nigg, 2004).

Table 9.3 Means (± SDs) of lower limb angular velocities (°.s⁻¹) during each condition

<table>
<thead>
<tr>
<th>Variable</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle velocity at TD</td>
<td>-34.6 ± C,D,E,F</td>
<td>6.4 ± 4.5 C,D,E,F</td>
<td>79.1 ± 158.5 A</td>
<td>40.2 ± 89.6 A</td>
<td>79.7 ± 88.0 A,B</td>
<td>87.9 ± 122.6 A,B</td>
</tr>
<tr>
<td>Peak dorsiflexion velocity</td>
<td>-172.6 ± C,D,E,F</td>
<td>-234.0 ± 96.6 A,F</td>
<td>-276.1 ± 81.4 A,F</td>
<td>-312.3 ± 106.1 A</td>
<td>-341.3 ± 79.7 A,B</td>
<td>-341.9 ± 87.9 A,B</td>
</tr>
<tr>
<td>Rearfoot velocity at TD</td>
<td>-93.3 ± D</td>
<td>-85.5 ± 37.3 D</td>
<td>-124.5 ± 66.9</td>
<td>-153.5 ± 87.7 B</td>
<td>-140.1 ± 62.6</td>
<td>-162.9 ± 79.6</td>
</tr>
<tr>
<td>Peak eversion velocity</td>
<td>-168.7 ± D,E,F</td>
<td>-171.3 ± 62.5 C,D,E,F</td>
<td>-208.9 ± 68.5 B</td>
<td>-232.1 ± 97.8 A,B</td>
<td>-179.5 ± 76.4 A,B</td>
<td>-223.0 ± 98.2 A,B</td>
</tr>
<tr>
<td>Knee velocity at TD</td>
<td>319.6 ± 76.2 B</td>
<td>271.2 ± 85.3 A</td>
<td>312.2 ± 81.4</td>
<td>308.4 ± 74.9</td>
<td>285.7 ± 82.2</td>
<td>278.9 ± 66.6</td>
</tr>
<tr>
<td>Peak knee flexion velocity</td>
<td>-256.8 ± 62.0 B</td>
<td>-233.6 ± 59.0 A,D</td>
<td>-240.6 ± 64.1</td>
<td>-253.1 ± 64.1 B</td>
<td>-243.3 ± 57.3</td>
<td>-261.3 ± 65.0</td>
</tr>
</tbody>
</table>

* denotes significantly different to BFT with BFT stride length. † denotes significantly different to BFT with SH stride length. ‡ denotes significantly different to MS with BFT stride length. § denotes significantly different to MS with SH stride length. ¶ denotes significantly different to SH with BFT stride length. †† denotes significantly different to SH with SH stride length. TO = toe-off

Even though no statistical difference in preactivity was found, there were changes in initial TD angles and angular velocities at the knee and foot. This suggests that different leg geometries can be achieved with similar muscular activations and that both kinematic and EMG data are needed to understand how runners gear the leg for impact. It must be noted that there was no change in TD ankle angle, which is contrary to previous reports (De Wit, et al., 2000; Squadrone & Gallozzi, 2009). This is possibly a result of adequate barefoot treadmill familiarisation being given to runners prior to experimental testing in...
the current study, since results from our laboratory have shown TD ankle to change over a familiarisation period.

Table 9.4 Number of participants exhibiting each foot strike pattern (as classified by Altman and Davis, 2012) across each running condition

<table>
<thead>
<tr>
<th>Foot strike</th>
<th>BFT BFT stride length</th>
<th>SH stride length</th>
<th>MS BFT stride length</th>
<th>SH stride length</th>
<th>SH BFT stride length</th>
<th>SH stride length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rearfoot</td>
<td>3</td>
<td>7</td>
<td>5</td>
<td>6</td>
<td>9</td>
<td>10</td>
</tr>
<tr>
<td>Midfoot</td>
<td>5</td>
<td>3</td>
<td>6</td>
<td>5</td>
<td>2</td>
<td>2</td>
</tr>
<tr>
<td>Forefoot</td>
<td>7</td>
<td>5</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>3</td>
</tr>
</tbody>
</table>

9.4.2 Foot angle (foot strike)

The findings support suggestions that the initial foot angle (or foot strike) does not directly affect RE (Perl, et al., 2012). Furthermore BFT and MS conditions have no external heel lift, meaning the results imply that foot angle is not just a consequence of having different heel heights (Gruber, et al., 2013). Rather the observed change in foot angle for BFT to one that contributes more to plantarflexion results from shorter stride lengths. By adopting a shorter stride length it is easier to position the foot closer to the line of the centre of gravity and initially strike the ground more anteriorly on the foot. Interestingly, the shorter stride lengths appear more influential than heightened proprioception at altering foot angle, as BFT with BFT stride length and MS with BFT stride length do not differ significantly. Recently, it has been reported that surface characteristics contribute to foot strike patterns with rearfoot strike patterns being maintained on soft surfaces (Gruber, et al., 2013). Together with our results this suggests that it is a combination of both stride length and surface characteristics that affect foot strike modality. With cushioned shoes, a foot angle contributing more to dorsiflexion is demonstrated to a greater degree with longer strides (SH with SH stride length). With the removal of cushioning (SH to MS), the same is also true. By heightening proprioception (MS to BFT) longer strides produce a similar foot angle to shorter strides with less proprioception (MS with BFT stride length), therefore indicating it is these two factors combined that contribute to strike pattern.
9.4.3 Heel off

By lifting the heel off earlier in the contact phase, as observed during BFT and MS conditions, longer time is spent loading the metatarsals. This, in addition to the forefoot being more angled towards the ground (BFT conditions and MS with BFT stride length) suggesting a forefoot strike pattern, is a potential concern for injuries such as metatarsal stress fractures (Ridge, et al., 2013). The forefoot would be loaded twice during each stance period (at initial foot strike and in midstance) and for a significantly longer time period. It may be suggested that without a transition period that gradually increases exposure to BFT/MS running, this extra loading could have injury implications. Runners who have previously experienced forefoot injuries should be cautious about running BFT/MS, regardless of the potential performance improvements.

9.4.4 Pronation, stiffness and muscle activity

The small amount of pronation (low peak eversion and dorsiflexion) during BFT, particularly with shorter strides, agrees with previous findings (Paquette, et al., 2013) and suggests the subtalar joint is more stable when BFT, but may be a concern for mechanical shock absorption. The high TA activity exhibited during MS with SH stride length and BFT running, consistent with previous research (von Tscharner, et al., 2003), does not appear to be directly detrimental to the metabolic cost of running, contradicting the metabolic cost of cushioning hypothesis (Frederick, 1984). This is because BFT running conditions were the most economical conditions and only when running with a SH stride length was MS more economical than SH running.

A possible explanation as to why there is not a detrimental effect of the TA activity on RE may be found in the GLTA coactivation results. It is likely that together these levels of muscle activity helped stabilise (Marques, et al., 2013) and increase ankle stiffness. Therefore rather than having a compliant ankle, with high amounts of pronation absorbing the impact energy, the muscles in the lower limb acted as shock absorbers. MS with BFT stride length displays low coactivation and TA activity, so optimal ankle stiffness may not have been achieved, hence RE was similar to SH and worse than BFT. This may also
explain why there was greater BF activity when running MS with a BFT stride length, to aid lower limb stability (Marques, et al., 2013). Thus shorter strides without cushioning (SH) or heightened proprioception (BFT) result in less stability and may not be beneficial to performance. Furthermore the BF has a high proportion of Type II muscle fibres (Garrett, Califf, & Bassett, 1984) whereas the TA has a high proportion of Type I muscle fibres (Jakobsson, Borg, & Edstrom, 1990). Therefore an increase in TA activity is not as detrimental to RE as an increase in BF activity due to a greater efficiency of energy production. Previous research has revealed the BF to have a positive association with $\dot{V}O_2$, meaning greater levels of activity during stance were related to higher metabolic costs (Kyrolainen, et al., 2001).

However the implications of greater TA activity on bone stress and muscle strain should not be overlooked. Greater muscle activity of the TA would lead to greater compressive stress on the anterior side of the tibia and greater tensile stress on the posterior side. These muscle imbalances are potentially harmful as they reduce the protection capabilities of the muscles (Mizrahi, Verbitsky, & Isakov, 2000a), which are needed as bone is weaker in tension than in compression (Reilly & Burstein, 1975). It may also explain the high proportion (30.8%) of shin complaints reported during a three-month training program which aimed to change a runner’s rearfoot strike pattern to a midfoot strike (Giandolini, Horvais, Farges, Samozino, & Morin, 2013b). So by trying to maintain joint stiffness to aid performance, the removal of a cushioning layer without shorter strides (MS with SH stride length) could lead to muscular imbalances across the tibia and greater tensile stress. Similarly, the removal of cushioning with heightened proprioception (both BFT conditions) could produce muscle imbalances and tensile stress.

9.4.5 Impact acceleration and effective mass

Our final hypothesis was also only partially supported, as only BFT running exhibited high impact accelerations and MS running demonstrated similar values to SH running. The high impact accelerations observed during BFT running may have been a result of a potential reduction in effective mass. Two kinematic conditions at TD were satisfied only by the BFT conditions; these
were that the knee was more flexed and foot angled more towards the ground (toes lower than heel). Both these factors, in isolation, have been shown to reduce the effective mass (Derrick, 2004; Lieberman, et al., 2010). Along with the removal of a footwear mass, these changes would imply that the BFT condition had the lowest effective mass. If effective mass denominates in the relationship between impact force and acceleration, as explained by Derrick (2004), then increased impact acceleration and potentially decreased impact force would be expected, as observed previously (Lieberman, et al., 2010; Squadrone & Gallozzi, 2009). High impact accelerations have been shown to lead to increased energy absorption at the hip, knee and ankle (Derrick, Hamill, & Caldwell, 1998). Greater energy absorbed via eccentric contractions can lead to greater energy released during concentric contractions, enhancing the efficiency of the stretch-shortening cycle (Komi, 2000). This implies that when increasing proprioception (MS to BFT) and removing external cushioning (SH to BFT) enhanced RE is prioritised over reducing high impact acceleration and possible injury risk.

9.4.6 Kinematics

It has previously been shown that MS running does not mimic BFT running (Bonacci, et al., 2013), and the current findings support this. The shank is rotated further forwards during SH and MS due to similar peak knee flexion and greater peak dorsiflexion than the BFT conditions. This suggests a lower CoG during stance during SH and MS running. At TO the ankle is more plantarflexed in SH and MS compared to BFT, which implies that runners were trying to ‘push-up’ to a greater degree (Moore, et al., 2012). Previous research has shown that less plantarflexion at TO is a factor in economical running (Moore, et al., 2012), possibly due to a greater proportion of propulsive force being directed forward, a feature of BFT running (Paquette, et al., 2013). These differences in ankle angle during stance likely contributed to the observed greater vertical oscillation in SH running compared to BFT with a BFT stride length. Consequently, this explanation also addresses why BFT with SH stride length generated a similar vertical oscillation to the MS and SH conditions, as differences in dorsiflexion angles were smaller than those when running in the
BFT with BFT stride length condition. Whilst for many a lower vertical oscillation is a logical characteristic of an economical running form (Anderson, 1996), few have reported supporting empirical evidence (Dallam, et al., 2005). Given the small change of 0.65cm (6.8%) it is unlikely that it was the determining factor in RE differences, particularly as MS with SH stride length had a similar vertical oscillation to SH with SH stride length yet was more economical.

Figure 9.2 Representative sample ankle angle data during each condition.

The lower angular velocities reported during BFT running could have also contributed to the improved RE, in line with previous findings (Moore, et al., 2012). It has been suggested that slower rates of pronation reduce the strain on the musculoskeletal system (Willems, et al., 2006), a trait which was observed when BFT. The change in muscular activity could have resulted in the change in angular velocities. As the TA dorsiflexes and inverts the foot, perhaps the greater TA activity opposed the eversion movement, consequently slowing down this motion. However, it is harder to explain how it contributed to the lower ankle angular velocity. Conceivably it is because the TA was used to absorb the
impact shock rather than dorsiflexing the foot. This would mean that the higher activation produced less movement at a slower rate, as it acted to move the ankle in a controlled fashion and attenuate the impact. Future investigations considering lower limb kinematics and their relationship with lower limb muscular activation are warranted to understand how the two factors interact during running.

9.4.7 Limitations

Whilst the methods employed mathematically adjusted for shoe mass, the effect of the change in leg moment of inertia during both BFT and MS running was not directly accounted for in the adjusted RE. Results show that the effect on the energetic cost of walking is slightly higher when changing leg mass than leg moment of inertia (4 and 3.4% increase respectively) (Royer & Martin, 2005). Therefore it is possible that the potential effect of decreasing leg moment of inertia on RE was encompassed in the calculation to adjust for the decrease in mass. On the other hand, the adjusted RE could have slightly underestimated the overall effect of running BFT or MS on the metabolic cost of running. However, given the current level of understanding, the method utilised in the current study was considered to be the most suitable approach for accounting for differences in shoe mass.

9.5 Conclusion

BFT running offers a small economical advantage over both MS and SH running. Therefore heightening proprioception and removing an external cushioning layer (SH/MS to BFT) may have the potential to produce slight performance benefits. To a lesser degree the same is true when just removing cushioning (SH to MS), but only when a SH stride length is maintained. It appears that the economical BFT advantage is due to a lower effective mass at ground strike, greater dependence on efficient (Type I) muscles during stance, less plantarflexion at TO, lower angular velocities and lower vertical oscillation. Whilst there is no additional metabolic cost of actively cushioning the lower limb, the greater impact accelerations and TA activity, in addition to low pronation
angles and earlier heel off, suggest that BFT running does not come without a potential increase in injury risk. Consequently, runners interested in transitioning to BFT running should not do so without an adequate, incremental transition period that gradually exposes them to the stresses of BFT running.
Chapter 10 General Discussion

Running mechanics have intuitively been connected to RE for a number of years, with research predominantly using inter-individual comparisons to discern which running mechanics are economical gait characteristics. This has highlighted a large number of variables that may be connected to an economical running form. However, through such comparisons it cannot be concluded whether an economical gait has developed over time or whether their running gait is unconnected to their RE, which may have improved through training. Accordingly, a short-term running intervention may be able to identify gait modifications associated with RE changes.

Typically, research in this area has tended to be very selective either considering kinematics or kinetics or muscular activity variables. Given the dynamic movement involved in running, encompassing angular deflection, force generation and muscular contractions whilst exploring mechanisms behind an economical running form is needed. Such a broad, inter-disciplinary approach may provide findings, which previously, have been missed or misinterpreted.

At present, many methodological issues within previous literature have meant results regarding economical running mechanics are equivocal, especially concerning muscular coactivation (Chapter 3) and BFT running (Chapter 4). Addressing these issues is needed prior to coming to conclusions and recommendations on what constitutes an economical running gait. Furthermore, there is currently a lack of research utilising a broad, inter-disciplinary approach to assess whether changing cushioning and proprioception effects RE, and what changes in running mechanics may influence such effects.

To address all the issues identified above, this thesis set out to use intra-individual comparisons of both acute and short-term self-optimisation. In particular, the thesis aimed to establish which features of running mechanics contribute to an economical running gait. The main findings, addressing each research question, will be summarised. This will be followed by sections highlighting potential injury risks due to footwear choice, how the interaction between running mechanics and muscular activity influence RE, detail on self-optimisation strategies that were evident throughout the thesis and the potential implications these have on performance and injury. Finally, the limitations of the
thesis shall be addressed, as well as the applications of the thesis findings, future research suggestions, recommendations and conclusions.

10.1 Summary of the Main Findings

10.1.1 Does a 10wkRP improve the RE of beginner runners?

Chapter 6 demonstrated that by taking beginner runners with very low fitness levels and merely increasing the volume of running they complete every week, for 10 weeks, improves their RE by 8.4%. The 8.4% improvement in RE fits in with the upper end of the range of improvements previously recorded and could have a big effect upon long distance running performance. For example, runners who improve their RE by 1% are estimated to be able to increase their speed per unit cost of $\dot{V}O_2$ by 0.049 m·s$^{-1}$ (Hanson, et al., 2011; Perl, et al., 2012). Therefore an increase of 8.4% would translate to an increase in maximal aerobic speed by 0.412 m·s$^{-1}$. Such an improvement would have significant implications on the time taken to complete long distance running events. For example completing a marathon (42.195 km) at 2.94 m·s$^{-1}$ (176 m·min$^{-1}$), rather than 2.53 m·s$^{-1}$ (2.53 + 0.41 = 2.94) (152 m·min$^{-1}$) would dramatically decrease the time to finish by 38 minutes. Furthermore, the 8.4% improvement was also experienced by the participants who, based on qualitative feedback, were able to run continuously for 30 minutes at the end of the 10 weeks, when this was not the case prior to the 10wkRP. Interestingly RPE was similar both before and after 10wkRP (14 and 13, respectively), suggesting that this decrease in $\dot{V}O_2$ did not translate to tangible changes in perceived effort.

10.1.2 Are the short-term improvements in RE a result of changes in running mechanics and/or running physiology?

The observation that the three variables of eversion velocity at TD, time to peak dorsiflexion and knee extension at TO significantly contributed to the change in RE indicates that biomechanical, rather than physiological adaptations are responsible for decreasing the metabolic cost of running. Further to these biomechanical changes there were several other variables that also changed over time and may have indirectly influenced RE. For example the lower
angular velocities, possibly a result of differences in muscular activity, may have placed less strain on the body and be produced with less mechanical energy. Thus a greater mechanical efficiency could contribute to a greater metabolic economy. The later occurrence of peak dorsiflexion, the lower plantarflexion and knee extension produced at TO, the more aligned leg axis angle and GRFr during propulsion and the greater propulsive force produced are likely to be inter-linked. These results suggest the body is able to translate a greater proportion of the GRFr forwards after 10wkRP. Collectively, these adjustments could improve RE by enhancing the SSC and facilitating a longer eccentric phase and thus, greater storage of elastic energy during the absorption phase, which is subsequently released during the propulsive phase. The fact that the calf muscle became less flexible after 10wkRP could have also contributed to enhancing the SSC as the MTU becomes less compliant, yielding greater amounts of stored elastic energy during the eccentric phase i.e. as the calf stretches during impact (Nelson, et al., 2001). Recent observations have strongly associated increasing musculotendinous stiffness with improved RE, again suggesting that this is indicative of greater energy storage and release (Albracht & Arampatzis, 2013; Spurrs, et al., 2003). The fact that the SRT scores were unchanged over the 10 week period, suggests that running may elicit greater MTU stiffness in the distal lower limb muscles, rather than the proximal muscles such as the hip extensors and knee flexors.

It is evident that when considering the running mechanics that influence RE, a broad approach is required encompassing as many elements as possible, such as kinematics, kinetics and flexibility. Moreover rather than trying to understand how each variable contributes to an improved RE in isolation, a deeper understanding can be generated by combining all those that significantly changed over time. Interestingly, there were more changes in kinematics than kinetics. There could be two contrasting explanations for this, either optimal kinetics are adopted instantaneously or only optimal kinematics can be acquired in the short-term, with optimal kinetics being a long-term adaptation requiring at least several months. However from the current findings it is unclear which strategy regarding optimal kinetics is applicable.

Contrary to previous research, there appeared to be no physiological explanation for the improved RE (Brooks, Hittelman, Faulkner, & Beyer, 1971a,
Conceivably by using highly inexperienced runners, alterations to running mechanics were quicker to develop than physiological changes, even though physiologically such a participant cohort has the greatest margins for physiological improvement. Any further, long-term changes to RE may be a result of physiological adaptations. The goal of the 10wkRP was to be running continuously for 30 minutes, meaning speed or distance covered was not important. Perhaps if participants were encouraged to run as far as possible in 30 minutes they would have trained harder during the 10 weeks and more physiological changes may have been observed.

Whilst kinematics explained a high proportion of the variance in the change in RE, the underlying muscular activity strategies adopted were not explored and may have contributed to adjusting certain kinematics, such as slower rates of movements and altering angular deflection. Such muscular strategies required attention, specifically the effect of muscular coactivation and duration of muscular activation.

10.1.3 Is greater muscular coactivation associated with a higher metabolic cost of running?

It was evident from Chapter 7 that greater muscular coactivation was related to greater $\dot{V}O_2$. This relationship was apparent across each speed performed. Physiologically this relationship is logical, due to the agonist having to produce a greater force, requiring more energy to overcome the force of the antagonist, which also requires energy. The findings are also specific to the extensor (RFGL) and proximal (VLBF and RFBF) pairs measured. A common factor in all pairings was the muscle group the quadriceps, one of the biggest muscle groups in the leg. Activation of this large muscle group has been associated with the metabolic cost of walking (Lewek, et al., 2012) and running (Modica & Kram, 2005). Furthermore higher thigh coactivation during walking has been linked to worse walking economy (Christiansen, Davidson, Schenkman, & Kohrt, 2011; Peterson & Martin, 2010). Therefore, when assessing the muscular coactivation-metabolic cost of running relationship muscle pairs involving the quadriceps are of particular importance.
To allow direct comparison with the work by Heise et al. (2008) a similar muscular coactivation calculation was used. Therefore the contradictory findings in this thesis of greater coactivation being associated with a greater metabolic cost of running cannot purely be down to computation methodology. The main differences were the length of time participants ran for and the speed at which they ran. Both of these may be of particular importance with regard to fatigue, which is known to affect muscular coactivation (Kellis, et al., 2011). Participants were instructed to self-select a speed eliciting a hard RPE (RPE of 6 on a 0-10 scale) and instructed to run for 30 minutes by Heise et al. (2008). RE was defined as the average $\dot{V}O_2$ over the final two minutes of the 30 minute run. There appeared to be no calculation to determine whether the $\dot{V}O_2$ recorded was indeed a steady-state, it is conceivable that participants may have been eliciting the slow-component of $\dot{V}O_2$ during their 30 minute run. If this were the case a true steady-state would not have been observed, with possible ramifications upon muscular coactivation. Although not conclusively demonstrated, the slow component of $\dot{V}O_2$ may reflect reduced muscle efficiency via progressive recruitment of fast twitch muscle fibres (Borrani et al., 2001; Borrani et al., 2003; Gaesser & Poole, 1996), which may have affected coactivation.

Assessing RE on two separate occasions four days apart during chapter 7 allowed the typical error for RE from our laboratory to be calculated. For the three different speeds the typical errors were: 2.52, 1.89 and 2.70% for 152, 183 and 200 m-min\(^{-1}\) respectively. Thus the average typical error was 2.37%. Therefore differences greater than this (i.e. changes in RE $> 2.37\%$) are deemed worthwhile and differences less than this cannot be conclusively reported as resulting from the given intervention. All typical errors calculated were similar to those previously reported for RE (Briswalter & Legros, 1994; Morgan et al., 1994a; Morgan, Martin, Krahenbuhl, & Baldini, 1991; Saunders, Pyne, Telford, & Hawley, 2004b).

The findings from chapter 7 demonstrate that the relationship between muscular coactivation and the metabolic cost of running is muscle specific, but what is not known is how speed affects muscular coactivation. This is particularly important given the contradictory previous research (i.e. Heise et al. (2008)), which used a range of speeds. It is likely that muscular coactivation is speed dependent,
given that activation of each individual muscle is also speed dependent (see Review of Literature II).

10.1.4 Does running at faster speeds elicit higher muscular coactivation in the lower extremities?

At the faster speeds the time spent coactivating was actually found to decrease in the distal lower limb muscles (GLTA) and the leg flexor muscles (BFTA). The TA, involved in each pairing, was shown to decrease in activation duration with increasing running speed, supporting previous evidence (Mann, et al., 1986). Consequently, it can be concluded that the reduced TA activation time is the mechanism behind shorter coactivation times. The decrease in coactivation could facilitate propulsion, as the gastrocnemius acts to plantarflex the foot (Hamner, et al., 2010). Additionally, it could indicate that the biarticular gastrocnemius muscles are influencing tibial rotation to a greater degree through eccentric contraction and effectively utilising energy transferred from the RF and BF activation to produce and control knee flexion. The results from the RFGL coactivation support this mechanical strategy, as it was the only muscular pairing where the absolute time spent coactivating remained similar across speeds. This suggests that even at faster speeds, such a muscular coordination may be of paramount importance to an individual’s running gait. Further work into the contribution of muscular coactivation changes on running mechanics would increase the knowledge of the mechanisms regarding why muscular coactivation may decrease with increasing speed.

Combining the knowledge of economical muscular strategies and kinematic adjustments to an experimental manipulation within the topical area of BFT running has previously not been undertaken. Therefore, the final sections of the thesis were dedicated to exploring self-optimisation, in terms of BFT familiarisation and acute gait manipulation. It was necessary to investigate BFT treadmill running familiarisation prior to experimental manipulation of such running, as currently it is not known how long it takes for habitually SH runners to produce a stable BFT running gait and, therefore, exactly how runners acutely adapt to BFT running.
10.1.5 Can runners produce a consistent gait pattern within 10 minutes of running BFT on a treadmill?

10.1.6 What specific gait adjustments occur as a result of familiarisation?

It was evident that BFT treadmill familiarisation takes longer than SH treadmill familiarisation (20 and 6-9 minutes, respectively), contradicting our hypothesis, in addition to highlighting that the complexity of adjusting running mechanics to the lack of footwear is greater than modifying running mechanics to the movement of a treadmill belt. The interplay of a lack of external cushioning and heightened proprioception mean that runners have greater awareness of the mechanical stress they are placing their feet and lower limbs under when running. What is hard to determine from the findings in chapter 8 is whether initial alterations, in the first few minutes, are a result of trying to attenuate impact and if alterations that took slightly longer to occur are a result of economical self-optimisation. Simultaneously collecting $\dot{V}O_2$, kinematic and kinetic data would enable researchers to assess not just $F_{z\text{impact}}$ and loading rates, but also mechanical efficiency, which may help address why these alterations take place.

After familiarisation runners adopted a different strategy than that previously reported to potentially reduce $F_{z\text{impact}}$ (De Wit, et al., 2000). Runners appeared to lower their $m_{\text{eff}}$ striking the ground. By modifying the leg geometry prior to TD, the tibia was rotated further forwards placing the knee and ankle into a greater degree of flexion. Both strategies of attenuating mechanical stress may not necessarily be mutually exclusive, but rather it could be argued the findings from chapter 8 are acute gait alterations to prolonged running exposure and perhaps, the flatter foot gait adaptation is more akin to acute gait alterations with minimal running exposure i.e. running a few meters over a force plate.

The final experimental chapter of the thesis addressed whether BFT running was more economical than MS and SH running, and in particular investigated the biomechanical (running mechanics and muscular activity) mechanisms behind any changes to RE. This allowed acute self-optimisation to be examined.
10.1.7 Will running BFT with a BFT stride length be the most economical way to run when compared to BFT with a SH stride length, MS with a BFT stride length and with a SH stride length, and SH with a BFT stride length and with a SH stride length?

10.1.8 Does running with a SH stride length during MS running produce a better RE than running with a BFT stride length and are both more economical than SH running?

Our hypotheses were confirmed with regards to the fact that BFT with a BFT stride length was the most economical condition and MS with a SH stride length was more economical than SH with SH stride length. However there was no difference between each stride length condition whilst wearing minimalist footwear.

The novel aspect of mathematically adjusting absolute $\dot{V}O_2$ to account for the difference in shoe mass between the three footwear conditions in chapter 9, allowed the effect that varying footwear cushioning and proprioception had on RE to be analysed, without compromising running mechanics. Generally, the lack of external cushioning (BFT and MS conditions) equated to more economical running. Further to this, increasing a runner's proprioception (from MS to BFT) resulted in an additional improvement in RE (Table 10.1). It is possible to infer from the results presented in Table 10.1 that RE appears to be enhanced to a slightly greater degree when transitioning to heightened proprioception (MS to BFT) compared to transitioning to a lack of cushioning (SH to MS). This is because on average (considering both stride length conditions together) heightened proprioception improves RE by 3.03% and taking away external cushioning improves RE by 2.55%. Consequently the jump from SH to BFT has the greatest average improvement in RE (5.58%). Given that the average typical error from our laboratory for RE is 2.37% the change in RE reported between MS and SH running is just on this threshold. Interestingly, MS with BFT stride length is non-significantly lower than SH with SH stride length and the percentage change that this represents falls within the typical error range determined from our laboratory. Therefore based on both the T-test and typical error results the percentage change in RE when MS with a BFT stride length is likely to be a result of biological variation, equipment or testing.
procedures (Saunders et al., 2004b) rather than the change in footwear and/or stride length.

**Table 10.1 Differences in RE (%) between footwear conditions**

<table>
<thead>
<tr>
<th>Variable</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
<th>BFT stride length</th>
<th>SH stride length</th>
</tr>
</thead>
<tbody>
<tr>
<td>RE (SH adjusted)</td>
<td>6.50</td>
<td>4.91</td>
<td>1.99</td>
<td>3.11</td>
<td>0.47</td>
<td>Baseline</td>
</tr>
<tr>
<td>RE (MS adjusted)</td>
<td>Baseline</td>
<td>-1.43</td>
<td>-4.83</td>
<td>-3.62</td>
<td>NA</td>
<td>NA</td>
</tr>
</tbody>
</table>

Positive difference denotes that the condition had a better RE than baseline (represented as either SH with SH stride length or BFT with BFT stride length). Negative difference denotes that the condition had a worse RE than baseline.

**10.1.9 What mechanisms are behind changes in RE during different stride lengths when varying cushioning and proprioception?**

Economical running conditions (i.e. BFT and MS with SH stride length) exhibited high TA activity during stance. This goes against the cost of cushioning hypothesis (Frederick et al., 1983) because the increase in muscular activity did not appear to be directly detrimental to the metabolic cost of running. However given the high proportion of Type I muscle fibres (Jakobsson, et al., 1990), there is a high efficiency of energy production from the TA. Thus, it could be argued that running BFT (heightened proprioception, no cushioning) and MS (no cushioning) with a SH stride length evokes an efficient use of muscular activation by relying on slow twitch muscle fibres. However it is important to note that other factors, such as conduction velocity, will also influence a muscle's energy efficiency but cannot be determined from the current findings.

In agreement with the increase in TA activity during stance, von Tscharner et al. (2003) found higher TA activity during the impact phase when BFT compared to when SH, yet a lower TA pre-activity. However, we did not find any difference in pre-activity across the muscles measured. Based on recent results from Giandolini and colleagues (2013a), the lower TA pre-activity is likely to be the mechanism responsible for changing foot strike modality, from rearfoot to mid/forefoot. This may explain why no difference was found in chapter 9, as we did not specify which foot strike modality individuals should use, as evident by
the variety of foot strikes observed (Table 9.3). Therefore greater variations were seen in muscular pre-activity patterns, which contributed to significant differences not being found. Interestingly Giandolini et al. (2013a) also reported the TA activity to be significantly lower over the entire gait cycle when midfoot striking and when running in racing flats (i.e. MS) with a midfoot strike pattern and a 10% increase in stride frequency (i.e. decrease in stride length). Again, it appears that the instruction to midfoot strike may be the distinguishing characteristic resulting in a lower TA activity. It is important to note that RE was not investigated in the study by Giandolini et al. (2013a) therefore conclusions on economical running cannot be made. Whilst there was a higher proportion of midfoot/forefoot strikers in the economical running conditions, the combination of utilising the TA and not the BF to help stabilise the lower limb (along with other kinematic adjustments), seems to be a key characteristic.

Similar to chapter 6, lower joint angular velocities were reported in the most economical condition in chapter 9. These were both at TD and peak, yet in chapter 9 there were no alterations in pre-activity or consistent changes in iEMG that may have accounted for the lower velocities. This implies that for the same level of muscular activity in the lower limb, movement can be adjusted, reducing the rate of angular change and in particular the overall rate of pronation. Without considering muscular activity during specific phases, such as loading and propulsion, it is not specifically clear how the combination of angular velocity and muscular activity contributes to RE. Perhaps the more economical conditions utilise muscle activation to control and minimise angular deflection and the rate of movement, whilst uneconomical conditions use it to produce greater movement at faster rates activating a higher proportion of fast twitch fibres. At present, such an explanation is purely speculative. Empirical evidence is needed to analyse the role of muscle activity in terms of movement control.

Vertical oscillation was significantly reduced between BFT with BFT stride length and SH with SH stride length, supporting the previous argument presented by many researchers that a low vertical oscillation is crucial to a good RE (Anderson, 1996; Saunders, et al., 2004a; Tseh, et al., 2008; Williams, 1985). Minimising the vertical motion means that energy is not unnecessarily wasted and instead can be directed to the forward progression of the run. The
reduced dorsiflexion from SH with SH stride length to BFT with BFT stride length coupled with an unchanged knee flexion angle suggests that the hip was positioned higher during stance when BFT running compared to when SH as the tibia was rotated further back. So rather than positioning themselves ‘lower’ to the ground during BFT running, runners appear to have a stiffer, less compliant leg that helps minimise the total displacement of their centre of mass. This finding is supported by previous research showing BFT running to exhibit stiffer $k_{\text{vert}}$ and $k_{\text{leg}}$ than SH running (Divert, et al., 2005a). Reducing vertical oscillation rather than the actual height of the centre of mass, has been advocated as a beneficial strategy for enhanced RE due to the inverse relationship observed between $\dot{V}O_2$ and centre of mass oscillation (Halvorsen, Eriksson, & Gullstrand, 2012). Runners appeared to achieve such a result through increasing their leg stiffness when running BFT with a BFT stride length.

The notion that individuals can alter their $m_{\text{eff}}$ appears to have significance in the findings from chapter 9. Firstly, the fact that BFT and MS running exhibited different impact accelerations even though both lack external cushioning is a novel finding that can be explained by changes in leg geometry. Secondly, MS and SH running had similar impact accelerations, even though the SH condition had an external cushioning layer. Again, this can be explained by differences in leg geometry at TD (see Chapter 9 for explanation). The lowering of $m_{\text{eff}}$ by greater leg flexion, principally at the knee, is believed to be a runner’s way of self-optimising their running gait to reduce injury risk. Such a gait alteration may have the potential to increase the metabolic cost of running (Derrick, 2004). However this argument is not supported by the results presented in this thesis. On the contrary, the conditions believed to have the lowest $m_{\text{eff}}$, actually had the lowest $\dot{V}O_2$. The question regarding the contribution that lowering the $m_{\text{eff}}$ had on RE however cannot be answered using these results. Perhaps such changes in $m_{\text{eff}}$ are detrimental to RE, yet they were countered by other economical gait alterations. Without modelling energy demand and $m_{\text{eff}}$ during running the answer to such a question may remain elusive. The difficulty in determining what factors not only contributed to improved RE, but also why/how such changes improve RE is amplified by the multifactorial nature of what constitutes ‘good’ running mechanics. Furthermore, variables should not just be
considered in isolation and the inclusion of EMG seems particularly necessary to further the understanding in this area.

Whilst the choice of footwear (or lack of) has potential performance implications, researchers also need to consider the impact of footwear on the potential to injure runners. Impact accelerations were recorded to provide information on how BFT, MS and SH running affects the deceleration of the lower limb during initial ground contact.

10.1.10 Do both BFT and MS running elicit higher impact accelerations compared to SH running?

Contrary to our hypothesis, only BFT running produced greater impact accelerations than SH running (Table 9.2). The higher accelerations when running BFT are a potential concern, especially when coupled with the greater loading rates reported by others (De Wit, et al., 2000; Paquette, et al., 2013; Sinclair, et al., 2013a). So although BFT running may eliminate or attenuate the $F_{\text{zimpact}}$, it may increase other variables associated with overuse injury. The lack of cushioning present in BFT running cannot completely explain the higher impact accelerations, as MS running, which also lacks cushioning, exhibited lower impact accelerations. The attenuation of $F_{\text{zimpact}}$ is believed to be brought about by the heightened proprioception experienced when BFT (Robbins, et al., 1989). Thus, by adopting a running gait that reduces force (and $m_{\text{eff}}$), runners may inadvertently increase impact acceleration. Such an increase may be harder for runners to detect than a higher force. It may be advisable therefore for older/veteran runners to not transition to BFT running on the basis that they already exhibit high loading rates (Bus, 2003; Lilley, Dixon, & Stiles, 2011), which BFT running may make worse. Further to the impact accelerations recorded, other injury implications drawn from the available kinematic data were also considered.

10.2 Influence of Footwear on Potential Injury Risks

Based on the lower pronation values (peak eversion and dorsiflexion) obtained when participants ran BFT in chapter 9, it seems plausible to suggest that ‘over-pronators’ could be prescribed BFT running as a training intervention. This is a
potentially exciting finding, especially given the inconclusive evidence regarding the success of orthotic and/or motion-control shoes interventions (Mills, Blanch, Chapman, McPoil, & Vicenzino; Richards, Magin, & Callister, 2009). It is also far cheaper for the runner. On the other hand, the findings could be highlighting a potential concern with regard to mechanical shock absorption. Pronation is a natural foot movement that acts to absorb the impact of the foot colliding with the ground by ‘unlocking’ the transverse tarsal joint, thus increasing the flexibility of the foot (Novacheck, 1998). If the foot is not providing this protective mechanism then another structure, perhaps the MTU, will need to. This potential transfer of shock could result in different injuries occurring rather than reducing the rate of injuries in runners.

The metatarsals were loaded for longer during BFT and MS running compared to SH. This coupled with the lack of cushioning could be potentially dangerous for runners who have previously sustained a metatarsal stress fracture, as well as for runners who have no previous injury history (Ridge, et al., 2013). These findings may explain why Ridge et al. (2013) found higher levels of bone edema in participants who had completed a 10 week MS running training program compared to those who had completed it in traditional trainers. Without external cushioning the metatarsals have larger exposure to the GRF, as less force is attenuated. Additionally the greater time spent loading these small bones could induce micro-damage to the bone tissue, resulting in greater bone edema. This is a potentially worrying result and highlights the importance of implementing an effective transitioning program, as it is likely that the feet of novice BFT/MS runners are unable to cope with the increased exposure to the forefoot loading.

10.3 Biomechanical and Muscular Activity Influence upon Running Economy

Several studies have investigated the role of muscles during a gait cycle (Barrett, Besier, & Lloyd, 2007; Gazendam & Hof, 2007; Guidetti, Rivellini, & Figura, 1996; Hamner, et al., 2010). This was not the specific aim of either chapter 7 or 9. Rather the aims were to understand how the muscular activation strategies contributed to the metabolic cost of running. Furthermore, rather than just considering the function of the muscles i.e. mono- or bi-articular (Heise, et al., 2008), the muscle fibre distribution and the influence this may have on
efficiency, and thus economy, was considered. This approach provided a unique insight into BFT running, as chapter 9 is the only study to comprehensively address muscular activity strategies adopted when running BFT.

Chapter 7 demonstrated that muscular coactivation was associated with an increased metabolic cost of running. However presenting such a simple relationship regarding the interaction of two muscles and RE cannot explain some of the findings in chapter 9. For example and partially contradicting chapter 7, the thigh coactivations were similar across all conditions. Intriguingly, the running condition with the lowest GLTA coactivation had a worse or similar RE to SH or BFT running, both of which had higher coactivations. However, the GLTA coactivation was not related to RE in chapter 7. Therefore it is conceivable that the role muscular coactivation plays in terms of joint stabilisation is of greater importance when considering the distal lower limb coactivations. This notion is synonymous with the muscle strategy adopted by women with a history of falls who exhibit greater shank coactivation to aid stability (Marques, et al., 2013). Such a strategy is especially important with regards to the BFT with a BFT stride length condition, which had very low amounts of mechanical shock absorption via pronation. The findings suggest that either to compensate for or the mechanism behind the small amount of pronation, was greater TA activity and GLTA coactivation. The TA, which acts to invert and dorsiflex the foot, may be taking the brunt of the impact shock by creating a stiffer ankle joint through greater GLTA coactivation. Heise et al. (2008) argued that lower limb coactivation might produce an efficient SSC as greater amounts of elastic energy are stored in the MTU during the impact phase, and hence there is more elastic energy released during propulsion. It is possible to infer from these results that when running BFT the muscular strategies and kinematic adaptations used might in fact be inherently efficient modifications that enhance RE, if we are to believe that we evolved as economical BFT runners (Lieberman, et al., 2010).

The composition of muscle fibres within a muscle is important in determining how efficient it is, with slow twitch (Type I) fibres being more efficient at energy production than fast twitch (Type II) fibres (Crow & Kushmerick, 1982; Kushmerick, Meyer, & Brown, 1992). For example the TA has predominantly
slow twitch fibres (Jakobsson, et al., 1990), which aerobically generate
adenosine triphosphate (ATP) and are more fatigue resistant than fast twitch
fibres. The BF on the other hand has a greater proportion of fast twitch fibres
(Garrett, et al., 1984), producing ATP through anaerobic metabolism. Therefore
the switch to a greater reliance on the TA during the most economical
conditions (BFT with BFT stride length and MS with SH stride length) suggests
that innately efficient strategies are adopted when runners run BFT with a
shortened stride length or when retaining longer strides during MS running.
Previous observations have demonstrated that the slow twitch muscle fibres
have a more efficient SSC through recoil of elastic energy (Bosco & Rusko,
1983; Bosco, Tihanyi, Komi, Fekete, & Apor, 1982). Consequently, it is possible
that the greater reliance on the TA could verify the claim by Divert et al. (2008)
that BFT running is more mechanically efficient than SH, as greater energy may
be produced in the TA even with a lower metabolic cost to running. However,
the higher levels of BF activity in the uneconomical conditions appear to be a
stability muscular strategy (Marques, et al., 2013), adopted to combat low shank
coactivation levels to the detriment of RE. It must be acknowledged that this
presents a simplistic overview of muscle fibre distribution and energy efficiency.
Inter-individual variability of the muscle structure has been shown to explain
differences in RE within a homogeneous group of runners (Kyrolainen, et al.,
2003), and may also explain some of the differences found in this thesis.
However without muscle biopsy data this information cannot be determined.

Given this high level of TA activity, the amount of dorsiflexion produced during
BFT running was, perhaps surprisingly, lower than other conditions. Therefore,
the TA activation was not generating greater amounts of movement in the
sagittal plane, indeed quite the opposite. It could be argued that the TA was
instead performing the two roles mentioned above, opposing eversion and
helping to stiffen the ankle joint during stance. This may have contributed to the
tibia being rotated further backwards and produced a stiffer leg, yet during MS
with SH stride length running a similar level of pronation to the SH with SH
stride length condition was produced. Therefore the high TA activity will have
contributed to the greater dorsiflexion, rather than opposing eversion. Evidence
also shows that TA activity is high during the loading phase of running when the
foot angle contributes to plantarflexion (i.e. BFT with BFT stride length) or to
dorsiflexion (i.e. MS with SH stride length), suggesting it plays a role in
attenuating impact force (Christina, White, & Gilchrist, 2001; Goryachev, Debbi, Haim, & Wolf, 2011). Consequently, purely measuring the TA activity could have led to a different explanation of the role it performed during BFT running.

There is difficulty in explicitly stating what the higher TA iEMG actually represents. In terms of the TA activity, it is possible that a greater number of motor units were recruited during BFT running, producing greater force during contraction (Hanon, Thepaut-Mathieu, Hausswirth, & Le Chevalier, 1998). However increases in iEMG could also reflect higher rates of firing (stimulation) or progressive increases in ATP requirement within muscles already recruited eliciting metabolic changes (Saunders, et al., 2000; Scheuermann, et al., 2001). In the current findings that latter notion of a greater requirement of ATP within the muscle is unlikely to be the case as there was no increase in metabolic cost. Both the increase in recruitment or increase in firing rates could suggest that the TA is actively trying to stabilise the lower limb and dissipate impact. This warrants further attention to improve our understanding of the role lower limb muscles play during impact. In particular, future work should employ frequency-domain wavelet analysis, which examines the intensity of the EMG signal within a certain frequency band. Thus far it has provided interesting findings in footwear studies (von Tscharner, et al., 2003; Wakeling, Pascual, & Nigg, 2002), but has not been used during investigations simultaneously measuring RE. Inferences regarding muscle fibre-type recruitment can be made from high and low frequency changes, as these appear to be consistent with fast and slow twitch motor unit changes (Wakeling, Pascual, Nigg, & von Tscharner, 2001) and hence, relationships between muscle fibre recruitment and RE can be assessed.

The experimental studies conducted as part of this thesis have highlighted the need to consider both running mechanics (kinematics and/or kinetics) and muscular activity when investigating RE. It was hypothesised in chapter 6 that alterations to muscular activity patterns may have been responsible for the lower angular velocities produced at TD. This was then investigated in chapter 9, but the hypothesis was not supported as pre-activity in all muscles analysed remained similar even though angular velocities were altered. Pre-activity levels however were fairly small with a high amount of inter-individual variance, meaning significant differences were extremely hard to find. It is possible that
runners were able to lower their angular velocities with minimal changes in muscular activity patterns. Therefore no additional muscular work was needed to change TD kinematics. Additionally it is conceivable that coactivity may have affected TD kinematics, but this was not assessed during pre-activation as previous evidence had shown only examination of singular muscle activity was needed to investigate muscular tuning (Boyer & Nigg, 2004). Nevertheless the findings from this thesis suggest that to try and understand the complexity of kinematic and muscular alterations during running both need to be simultaneously investigated to appreciate how the interactions affect the outcomes.

10.4 Self-optimisation: Implications for Performance and Injury Risk

Previously researchers have focused upon TD kinematics/foot strike patterns and the absorption phase of stance, believing that this held the key for an economical running gait, e.g. Perl et al. (2012). However the findings from this thesis show that the push-off, propulsive kinematics are very influential. An apparent feature of self-optimisation that was evident both during acute and short-term changes in running mechanics was that participants appeared to decrease the $I_{leg}$ at TO. Both chapter 6 and 9 found less plantarflexion at TO, with the knee also being less extended during chapter 6 (Figure 6.1). This would decrease the distance between the foot’s centre of mass and the origin of rotation, in this case the hip. A positive relationship is seen between metabolic rate and $I_{leg}$, so any decrease in $I_{leg}$ is mirrored by a decrease in metabolic rate (Browning, Modica, Kram, & Goswami, 2007). Therefore not only does greater leg flexion at TO place the lower limb in a better position for the swing phase, it may also be less energetically demanding to move due to a reduction in its resistance to rotation. Furthermore, BFT running benefits from a reduction in distal load i.e. footwear mass, which further decreases the $I_{leg}$ and thus the metabolic rate/metabolic cost (Browning, et al., 2007; Martin, 1985; Myers & Steudel, 1985; Royer & Martin, 2005). It is important to note that the consequence of previous studies strapping extra mass to the foot to account for the added mass of the shoe (i.e. Franz et al., 2012), could have altered the participants’ stride length (Browning, et al., 2007; Martin, 1985; Royer & Martin, 2005), angles at TO (Browning, et al., 2007), muscle activity patterns
(Browning, et al., 2007; Royer & Martin, 2005) and $l_{\text{leg}}$ (Browning, et al., 2007) to those which are not representative of natural BFT running.

Another self-optimisation strategy that was evident after 10 weeks of running was the alignment of the leg axis and GRFr, minimising the angle between the two through flexing the leg more at TO. It is likely that this was either a result of, or contributed to, being able to generate a greater amount of propulsive force during push-off. Interestingly, Paquette and colleagues (2013) reported BFT running to have a higher propulsive force than SH running. This, together with BFT running being the most economical running condition with less plantarflexion at TO in chapter 9 implies that this kinetic adjustment, and kinematic alignment, may have significant implications on RE as predicted by Chang et al. (2000). It is important to acknowledge that leg and GRFr alignment were not measured by Paquette et al. (2013) and kinetics were not assessed in chapter 9.

For a number of years, researchers have believed that the stride length runners naturally chose is near to or is their most economical optimal stride length. This was originally based on findings from Cavanagh and Williams (1982), with recent reports suggesting that this is certainly the case for trained runners who exhibit a stride length closer to their mathematically derived optimal than novices (de Ruiter, Verdijk, Werker, Zuidema, & de Haan, 2013). Cavanagh and Williams (1982) used habitually SH runners, thus the freely chosen stride length in their study is representative of the SH stride length measured in chapter 9. This particular study by Cavanagh and Williams acutely manipulated the participant’s stride length in six conditions: $\pm 6.7\%$, $\pm 13.4\%$ and $\pm 20\%$ of leg length from the freely chosen SH stride length. They reported that the absolute difference between economical optimal and freely chosen stride lengths was 4.2 cm, suggested to be a “reasonably large” deviation (p.34). However manipulation of their raw stride length data for each participant (Table 10.2) shows that this, on average, equates to a 3.2% difference in stride length. The majority (7 out of 10) appear to ‘over stride’. It could therefore be argued based on their data, together with the findings in chapter 9, that there is an optimal range for stride length. This range is the freely chosen stride length $\pm 3\%$. Consequently, the shorter strides that individuals tend to adopt when BFT,
between 2.5 - 3%, are within this optimal range, hence there was no effect of stride length on RE in chapter 9.

Table 10.2 Raw stride length data showing the difference between freely chosen and optimal stride length (adapted from Cavanagh and Williams, 1982)

<table>
<thead>
<tr>
<th>Participant</th>
<th>Freely chosen stride length (cm)</th>
<th>Optimal stride length (cm)</th>
<th>Absolute Difference (% of freely chosen)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>128.1</td>
<td>125.2</td>
<td>2.26</td>
</tr>
<tr>
<td>2</td>
<td>122.5</td>
<td>126.2</td>
<td>3.02</td>
</tr>
<tr>
<td>3</td>
<td>141.2</td>
<td>137.5</td>
<td>2.62</td>
</tr>
<tr>
<td>4</td>
<td>135.2</td>
<td>126.1</td>
<td>6.73</td>
</tr>
<tr>
<td>5</td>
<td>129.0</td>
<td>132.7</td>
<td>2.87</td>
</tr>
<tr>
<td>6</td>
<td>137.3</td>
<td>131.6</td>
<td>4.15</td>
</tr>
<tr>
<td>7</td>
<td>129.7</td>
<td>123.1</td>
<td>5.09</td>
</tr>
<tr>
<td>8</td>
<td>129.6</td>
<td>127.2</td>
<td>1.85</td>
</tr>
<tr>
<td>9</td>
<td>128.1</td>
<td>130.8</td>
<td>2.11</td>
</tr>
<tr>
<td>10</td>
<td>140.5</td>
<td>139.1</td>
<td>1.00</td>
</tr>
<tr>
<td><strong>Mean</strong></td>
<td><strong>132.1</strong></td>
<td><strong>129.9</strong></td>
<td><strong>3.17</strong></td>
</tr>
</tbody>
</table>

The fact that runners were able to naturally adopt a shorter stride length, without apparent conscious effort (based on discussions after testing, many were completely unaware that they had been running in two different stride length conditions), highlights that such a gait adjustment requires little effort, low energy demand and is comfortable for the runner. It is likely that major changes to stride length, greater than ±3%, requires a more conscious effort, greater energy demand and is less comfortable and natural as a result.

The results from Squadrone and Gallozzi (2009) lend some support to the concept of an optimal stride length range of ±3%. The stride length recorded during SH and MS running were statistically similar, but actually differed by 2%, and subsequently RE was significantly improved when MS compared to SH (Table 10.3). BFT running however actually resulted in a 7% decrease in stride length and resulted in a similar RE compared to SH. Thus, there was no benefit to running BFT. The question arises whether if they had run with only a 3% change in stride length when BFT whether RE would have significantly improved during BFT running. Contrary to the findings of chapter 9, MS with a 2% shorter stride exhibited enhanced RE compared to SH. However it must be noted that this was the stride length freely adopted by participants suggesting it
was the least metabolically demanding for them and it was not significantly different than the stride length exhibited during SH running.

### Table 10.3 Raw \( \dot{V}O_2 \) and stride length data, with absolute difference in stride length for different footwear conditions (adapted from Squadrone and Gallozzi, 2009)

<table>
<thead>
<tr>
<th>Footwear condition</th>
<th>BFT</th>
<th>MS</th>
<th>SH</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \dot{V}O_2 ) (mL·kg(^{-1})·min(^{-1}))</td>
<td>45.7</td>
<td>45.0</td>
<td>46.3*</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>2.19</td>
<td>2.29*</td>
<td>2.34*</td>
</tr>
<tr>
<td>Difference in stride length (%)</td>
<td>6.8</td>
<td>2.1</td>
<td>0</td>
</tr>
</tbody>
</table>

*Significantly different to BFT condition. * Significantly different to MS condition.

A successful short-term training intervention to adjust stride length beyond the 3% range has been performed by Morgan et al. (1994b). Prior to the intervention Morgan and colleagues mathematically derived the ‘optimal’ stride length for individuals using curve-fitting procedures and manipulating stride length by the same six conditions used by Williams and Cavanagh (1982). The outcomes of the intervention were that participants not only changed their stride length by 7.3%, meaning it was within 2.5% of their mathematically optimal stride length, but they also improved their RE. The study by Morgan et al. (1994b) demonstrates that stride length can successfully be changed by greater than 3% over a short period of time, improving RE in the process. However other investigations have not been as successful, notably due to unsuitable procedures without accurate mathematical calculations (Messier & Cirillo, 1989). Nevertheless, the range of 3% still holds true with regards to acute variations. Thus the importance of this range of economically optimal stride length translates into acute performance implications. Consequently, based on this theory, slight variations in stride length (±3%) would not be detrimental to RE and subsequent performance during, for example, a marathon.

The potential to calculate whether runners are choosing the most economically optimal stride length by systematically increasing and decreasing their stride is an interesting concept. However, results from chapter 6 imply that although runners can modify running mechanics towards an economical movement pattern, their step lengths and hence stride lengths, remained unchanged after 10 weeks of running. It is therefore conceivable that this gait characteristic is more resistant to natural change in the short-term. As variations in stride length
were not measured in this chapter, it cannot be determined whether a participant’s freely chosen stride length was indeed their most economical. However, it is unlikely that their chosen stride length was the most economically optimal, as novice runners tend to adopt stride frequencies that are 8% less than optimal (de Ruiter, et al., 2013). Consequently, greater improvements in RE may have been recorded if stride length was manipulated.

Assuming runners, both experienced and novice, tend to naturally over stride (de Ruiter, et al., 2013), this could not only have detrimental implications upon performance but also upon injury risk. Reducing stride length by 10% can decrease overuse injury risk and $F_{\text{zimpact}}$ (Edwards, Taylor, Rudolphi, Gillette, & Derrick, 2009). Conversely, however, smaller reductions in stride length, such as those found in BFT running, do not affect $F_{\text{zimpact}}$ (Giandolini, et al., 2013a). Therefore when BFT, runners may alter other running mechanics in an attempt to minimise $F_{\text{zimpact}}$ (Robbins, et al., 1989). Switching to either a midfoot or forefoot strike can reduce or eliminate $F_{\text{zimpact}}$ (Giandolini, et al., 2013a; Hamill, et al., 2011; Lieberman, et al., 2010; Quadrone & Gallozzi, 2009). This is achieved by modifying foot angle at TD and, consequently the $m_{\text{eff}}$ striking the ground. Quadrone and Gallozzi (2009) reported BFT and MS conditions to have a higher strike index (an indication of foot strike) and a flatter foot at TD compared to SH. Collectively, this suggests that foot strike was different to SH running and thus, $m_{\text{eff}}$ was lower (Lieberman, et al., 2010). Further to this, only the BFT condition had a significantly shorter stride length (Table 10.3), yet both BFT and MS had lower $F_{\text{zimpact}}$ than SH. This highlights that foot strike modality may be more influential than stride length on $F_{\text{zimpact}}$, suggesting a shorter BFT stride length is an economical self-optimisation strategy. Unfortunately running kinetics can only be inferred from the experimental data obtained in chapter 9 meaning substantial acute self-optimisation injury implications based on kinetic factors (i.e. loading rate, joint moments) cannot be made. However, results from chapter 6 suggest that beginner runners did not significantly alter the magnitude of many kinetic measures, $F_{\text{zimpact}}$ being one of them, although there was a trend to decrease peak joint moments and thus potentially joint loading, which may reduce the risk of overuse injuries (Franz & Kram, 2012; McClay, 2000). As discussed previously participants may have either not displayed excessively high loading rates/ $F_{\text{zimpact}}$, or that such a gait alteration took place immediately
(during the warm-up and familiarisation trials) or that it occurs as a long-term adaptation.

The use of accelerometer data to consider the impact of the lower limb with the treadmill provided some intriguing results regarding self-optimisation (see Chapter 9). Whilst it is plausible that higher impact accelerations could increase injury risk and stress on the tibia (discussed in Chapter 9), it has been argued that this is not necessarily the case. Derrick et al. (2002) stated that if the increase in acceleration is mirrored by a decrease in $m_{\text{eff}}$, than the rise in impact acceleration is not an injury risk, as a smaller portion of body mass is being accelerated upon ground contact (and there is a lower $l_{\text{leg}}$). However without being able to determine the actual $m_{\text{eff}}$ striking the ground, it is possible that $m_{\text{eff}}$ was not substantially lowered, yet impact acceleration was significantly higher when BFT, thus there could have been a greater injury risk associated with the most economical conditions (BFT running). Even if there was a decrease in $m_{\text{eff}}$, the cumulative strain placed through the muscles by the faster deceleration of the lower limb during repetitive impacts could lead to a greater number of muscular injuries. Quantification and/or modelling of $m_{\text{eff}}$ is necessary to determine the effect that higher accelerations may have on the lower limb muscles.

As discussed in chapter 9, the change in TA activity could have ramifications upon the amount of bone stress sustained. As the TA is responsible for regulating impact loading (Christina, et al., 2001), the possible alteration to both the compressive and tensile stress could reduce the capabilities of the TA muscle to protect the bone (Mizrahi, et al., 2000a). This could cause microdamage to bone tissue, which over time results in overuse injury (Chapman, et al., 2008). This could perhaps be overcome by increasing the muscular strength of the gastrocnemius to reduce the muscle imbalance or by increasing the strength of smaller muscles around the ankle. This latter approach may be of primary importance if BFT or MS (with a SH stride length) running is adopted in the long-term. Based on model calculations the smaller muscles around the ankle appear crucial to both performance and protection of the ankle joint, as they help increase the foot’s reaction time and reduce the forces going through the ankle joint (Nigg, 2005; Nigg & Enders, 2013). Therefore by increasing the strength of the smaller muscles, the TA activity may decrease as a result.
Notwithstanding the evidence presented in this thesis regarding self-optimisation, there is data that suggests humans are not tuned to maximise RE. This argument is based on findings that show there to be no specific speed that all back and lower limb muscles are ‘tuned’ to (Carrier, Anders, & Schilling, 2011). Rather each muscle activity was minimised over a range of speeds. The authors therefore take this to be evidence that humans have not evolved to maximise RE as previously thought (Lieberman, et al., 2010). Yet, in the study by Carrier and colleagues (2011) there was no measure of the metabolic cost of running, no consideration of muscle fibre-type and no description regarding the athletic status of the participants. The authors assumed high EMG relates to high muscle metabolism and thus, a poor RE, in addition to believing all muscles, regardless of fibre-type, to have similar metabolic rates. Furthermore research has shown that muscle characteristics, such as thickness, pennation angle and fascicle length differ between sprinters and distance runners, demonstrating that athletic status is an important factor (Abe, Fukashiro, Harada, & Kawamoto, 2001; Abe, Kumagai, & Brechue, 2000; Lee & Piazza, 2009). These limitations, as well as Carrier and colleagues’ (2011) own conclusions that humans are thus capable of sustaining a range of speeds, highlights the complexity of investigating self-optimisation.

10.5 Limitations

10.5.1 The running mechanics of the upper body

Running mechanics of the lower limb were the focus of this thesis, however, it is important to note that the upper body also affects an individual’s movement pattern and metabolic cost (Arellano & Kram, 2011, 2012; Hamner, et al., 2010; Hinrichs, 1987; Hinrichs, Cavanagh, & Williams, 1987; Pontzer, Holloway, Raichlen, & Lieberman, 2009; White, Scurr, & Smith, 2009; Williams & Cavanagh, 1987). Variables such as trunk inclination, arm swing, wrist excursion and/or arm position and breast kinematics were not measured. The arms and adequate breast support can help reduce rotational torque in the trunk for female runners and whilst it has been argued that arm kinematics may not improve performance, eliminating arm swing is detrimental to the metabolic cost of running (Arellano & Kram, 2011, 2012). This is because in addition to affecting angular momentum, the arms help contribute to the vertical oscillation
of the body during running (Hamner, et al., 2010; Hinrichs, 1987; Hinrichs, et al., 1987). Therefore by measuring this parameter an inference with regards to arm kinematics could be made. Vertical oscillation remained stable over a 10wkRP (Chapter 6) but decreased when running BFT (Chapter 9). Therefore it is unlikely that arm kinematics changed over the short-term (10wkRP), but perhaps variations in footwear, in particular the lack of cushioning and heightened proprioception, affected arm movement.

There is also growing research concerning the effect of breast support on running mechanics (White, et al., 2009). As all experimental testing sessions required continuous exercise (i.e. longer than five minutes of walking/running) participants were advised to wear comfortable sports bras. However participants were not screened for breast size and this may have been a confounding factor. Nevertheless, all statistical tests involved intra-individual data comparisons and therefore, the influence of such a confounding variable would have been reduced.

10.5.2 Does electromyography data represent the underlying muscular activity?

Surface EMG is a useful, non-invasive technique that was used to infer the underlying muscular activity of the lower limb muscles during chapters 7 and 9. The data generated quantifies gross electrical activity of the muscles, but is not without limitations. Firstly, EMG only reflects the electrical rather than mechanical timing of events and thus, introduces an element of electromechanical delay. Therefore interpretations of the EMG signal, particularly in regards to force production should be made with caution (Hof, 1984). Several factors influence the electrical signal received by the surface EMG electrode such as cross-talk, distance of de-polarised motor neurons from the electrode which can be affected by the movement of the muscle relative to the electrodes, amplitude cancellation and muscle fibre conduction velocity (Enoka, 2008; Weir, Beck, Cramer, & Housh, 2006). It is therefore important to acknowledge such limitations when interpreting EMG results. Secondly, several factors can also affect day-to-day repeatability (Chapter 7), such as changes in skin temperature influencing impedance, slight variations in electrode placement and differences in skin preparations (Daanen, Maze, Holewijn, & Van der Velde, 1990). Whilst the latter two factors were standardised, slight
alterations across the days may have reduced the reliability of the measures. However in general, good reliability was found in the majority of the muscles in one of more of the three speeds.

10.5.3 Impact acceleration as representative of tibial acceleration

Tibial acceleration is often used to estimate the loading occurring at the tibia during impact by affixing an accelerometer onto the skin at the distal anterior-medial aspect of the tibia, just above the ankle joint centre (Clansey, Hanlon, Wallace, & Lake, 2012; Laughton, Davis, & Hamill, 2003; Mercer, et al., 2003b). However, chapter 9 used an electrode affixed to the heel in between the kinematic calcaneus markers (see 5.2.12 Accelerometer) due to limitations in the number of available electrodes. During BFT running, the surface electrode with the integrated accelerometer was affixed directly to the skin using double-sided tape. However during MS and SH running the electrode had to be affixed to the footwear heel counter. Accordingly caution regarding the interpretation of impact accelerations should be exercised. It is likely that foot strike patterns adopted influenced the accelerations measured using an electrode on the heel. Additionally, the fact that the electrode was attached to an external layer rather than the skin in the MS and SH conditions means estimations of bone loading can only be inferred from the data. Notwithstanding these limitations, the conclusions drawn from the data such as BFT running exhibiting greater accelerations than SH, was in agreement with previous research even though the magnitudes were slightly lower in the current findings (McNair & Marshall, 1994; Sinclair, et al., 2013a). Additionally, accelerations recorded at the heel provide the best representation of impact forces during running (Nigg, 1986). Therefore the methods used to determine impact acceleration provided not just a good indirect estimate of tibial loading, but also of $F_{\text{impact}}$.

10.5.4 Mathematically adjusting $\hat{\dot{V}}O_2$

Mathematical adjustment to absolute $\hat{\dot{V}}O_2$ undertaken in chapter 9 for footwear mass technically adjusts the whole body mass i.e. at the centre of mass, and thus does not specifically adjust for mass at the foot. The limitation of such an approach is that it does not account for inertial changes and therefore the influence of a decrease in $I_{\text{leg}}$ on RE when going from SH to BFT was
unaccounted for. The mass effect has been well established, however the most prominent research employing the strongest methodology added mass without changing $I_{\text{leg}}$ (Divert, et al., 2008). Therefore the influence of changing $I_{\text{leg}}$ cannot be determined from such a study. It has been reported that $I_{\text{leg}}$ and added mass have separate and independent effects upon energy cost (Royer & Martin, 2005). But it is not known whether, when trying to account for changes in both, there may be a cumulative effect. To the author’s knowledge no previous studies have employed methodologies to adjust $\dot{V}O_2$ for $I_{\text{leg}}$. Therefore the calculation used in chapter 9 was deemed the most appropriate due to the current understanding regarding the effect of footwear mass and $I_{\text{leg}}$ on $\dot{V}O_2$. Furthermore given the moderate effect sizes the effect of BFT running on RE may still be apparent even if changes in $I_{\text{leg}}$ were accounted for.

10.6 Applications

The novel finding that runners naturally self-optimise towards a more economical movement pattern over a short-period of training supports the hypothesis of self-optimisation proposed by Williams and Cavanagh (Cavanagh & Williams, 1982; Williams & Cavanagh, 1987). Allowing runners to, most probably subconsciously, self-select the way they run meant runners adopted specific gait characteristics that helped improve RE. This potentially pulls into question whether or not the generic, universal running form instructed by coaches is in fact the most beneficial for a runner. The best technique coach may in fact be the individual themselves, as they find the ‘easiest’ or ‘laziest’ way of running (Snyder, Snaterse, & Donelan, 2012), adopting a comfortable running form that is tailored to them (Nigg & Enders, 2013). So rather than prescribing a one-size fits all technique, coaches and runners alike should perhaps acknowledge that running form is very idiosyncratic, and therefore each runner will have an individualised running form.

The findings from chapter 7 suggest that runners should aim to lower the amount of lower limb muscular coactivation they produce during running, particularly as a result of quadriceps contraction. Perhaps the anecdotal notion of running ‘relaxed’ could be prescribed to runners who seem to have high muscular coactivation. Thus, rather than trying to produce a stiff lower limb by
coactivating both the agonist and antagonist muscles, runners should aim to avoid actively stiffening the upper leg. It is possible that increasing stability through balance training may provide an internal, passive mechanism through which the lower limb produces adequate stiffness. Therefore runners would still be able to utilise and generate an efficient SSC, but possibly without the detrimental ramifications upon $\dot{V}O_2$. Introducing certain balancing/stability interventions, tailored specifically to running, into a runners training program may be beneficial.

Although the finding that familiarisation to BFT treadmill running occurs within 20 minutes was not directly connected to RE, it is an important finding for future research and demonstrates acute self-optimisation to BFT running. Without conducting the experimental procedure in chapter 8, the biomechanical observations made in chapter 9 would have lacked internal validity. The conclusions drawn based on the biomechanical adjustments could then have been a result of unfamiliarity. This is especially true at the ankle angle where familiarisation resulted in a more neutral, slightly dorsiflexed ankle angle at TD. Previously plantarflexion at TD has often been reported to be a characteristic of BFT running (De Wit, et al., 2000; Lieberman, et al., 2010), however this was not evident in chapter 9. It is therefore advocated that when experimentally testing individuals who are not BFT runners, adequate familiarisation of 20 minutes should be given. It is important to note that familiarisation relates to short-term consistency in running gait and that habituation to BFT running is likely to take much longer, perhaps even a few years as it relates to long-term adaptation.

The findings in chapter 9 provide strong evidence that BFT running can produce performance benefits. In addition to this, given that the runners were not instructed to run in a certain way, i.e. they received no information on specific gait adjustments that are often described as a ‘BFT running form’, the results demonstrate that after a period of familiarisation, runners will self-optimise their running gait during BFT running. Furthermore, the evidence strongly supports the notion that runners prefer a running gait that has the lowest energy demand. Previously it has been shown that runners produce a stride length that is similar during SH and MS running (Squadrone & Gallozzi, 2009). The RE produced when running MS with a SH stride length is better than SH with a SH stride
length. However there is no difference between SH with a SH stride length and MS with a BFT stride length. It appears that a natural gait adjustment (or lack of) is driven by underlying energy demand (Nigg & Enders, 2013). Runners therefore that take to MS running due to its claims of mimicking BFT running and with knowledge of certain BFT running characteristics may, unnaturally, try to reduce their stride length. Encouraging runners to naturally self-optimise when running in minimalist footwear could actually be more beneficial for their performance than encouraging them to adopt a ‘BFT running form’.

There are also injury implications from chapter 9, which have been described earlier (10.2 Influence of Footwear on Potential Injury Risks and 10.4 Self-optimisation: Implications for Performance and Injury Risk). Runners should not ignore these in pursuit of improved performance. How best to transition to BFT running is not yet understood, but the findings suggest that this should be addressed, as an adequate transition period is necessary to allow the body to adjust to the extra muscular activity, longer metatarsal loading and greater accelerations experienced.

10.7 Topics for Further Research

10.7.1 Could an economical running gait be trained?

The findings in the thesis present strong evidence of self-optimisation, however, this does raise an interesting question of whether or not an economical running gait could be trained. Whilst investigations have demonstrated that running gait can be trained in such a way that reduces tibial acceleration, $F_{\text{impact}}$, loading rates, vertical oscillation and step frequency (Crowell & Davis, 2011; Davis, Crowell, Fellin, & Altman, 2009; Eriksson, 2010; Eriksson, et al., 2011), it remains to be seen whether a purely technique based training intervention, with strong methodology, can improve RE. Messier and Cirillo (1989) performed a 5 week running technique intervention, with both verbal and visual feedback, but they failed to improve RE. However several limitations constrain how useful these results are. The primary limitation being that they did not provide a description of what constituted ‘excessive’ motion for each variable considered uneconomical. Therefore, it is possible that participants were not displaying movement patterns that were uneconomical. This may especially be the case
with regards to stride length, as optimal stride length was not mathematically calculated. More recently Warne and Warrington (2012) performed a training intervention incorporating transitioning to MS running. Their findings were positive in that runners became more economical, however, there were no specific technique instructions given and running mechanics were not recorded in this study. Therefore whilst technique changes can be hypothesised to have occurred, empirical evidence is lacking. A technique-based intervention aimed at training runners to become more economical would need to involve trained runners. This is because the intervention would have to be such that running sessions were similar in intensity, frequency and volume as their usual training schedule, thus minimising the likelihood that RE changed as a result of training changes.

10.7.2 Long-term running economy and gait development

This thesis only examined acute and short-term changes in RE, with running mechanics being able to explain a high portion of variance in RE. In the long-term, it is likely that runners will be able to produce a stable running pattern as they become habituated to running. As a result, mechanisms behind any further changes in RE are more likely to be physiological in nature. Several investigations have reported such mechanisms behind RE, yet these generally stem from correlational analysis derived from comparing different runners. Therefore to address long-term developments in both RE and running mechanics, specifically to assess whether physiological or biomechanical adaptations influence RE, a longitudinal investigation is required that follows a group of runners over a substantial period of time i.e. 2 years. This would provide crucial insight into not just physiological mechanisms but also development over time of both RE and running gait.

10.7.3 The potential for stability training to improve running economy

The importance of the stabilisation of the lower limb was highlighted in chapters 6 and 9. The musculoskeletal system can use both active and passive mechanisms to produce lower limb stability. It could be hypothesised therefore that if runners can rely on passive mechanisms, rather than active mechanisms that require energy to stabilise, they could potentially lower their metabolic cost.
of running by having a more efficient stabilisation strategy. The theory that inflexible muscles may contribute to a more efficient SSC and enhance RE was partially supported by the findings in chapter 6, and it is possible that by increasing this passive stiffness within the MTU that runners relied less on muscles to actively stabilise the lower limb. Research has shown that increasing the MTU stiffness is related to an improved RE (Spurrs, et al., 2003) and improving an individual's dynamic postural control through training involving functional balance tasks produces lower levels of coactivation (Nagai, et al., 2012). Translating such balance tasks into running specific tasks could have potential performance implications through increasing the use of passive stabilisation mechanisms.

10.7.4 Modelling the effective mass based on energy demand

Effective mass has received very limited attention from the academic community. Yet with the results presented in chapters 6, 8 and 9, together with the current interest in running mechanics, foot strike modality and BFT running, an argument could be made that knowledge and understanding regarding this principle needs to be increased. There are several factors that affect the m\textsubscript{eff}, they are: 1) leg geometry at TD (encompassing foot strike modality, knee angle, and rearfoot angle); 2) foot velocity at TD; 3) ankle and knee joint stiffness; 4) soft tissue stiffness and 4) shoe and surface characteristics (encompassing BFT/MS conditions and variations in terrain). Accordingly, alterations to any of these factors can bring about changes in timing and magnitude of F\textsubscript{zimpact} and impact acceleration.

The m\textsubscript{eff} has been modelled by several authors (Chi & Schmitt, 2005; Denoth, 1986; Lieberman, et al., 2010), demonstrating how it can be calculated using a combination of kinematic and kinetic data. A few assumptions are made in the calculations presented below: 1) only force and velocity in the vertical direction are considered; 2) m\textsubscript{eff} is constant during the impact phase and 3) acceleration can be estimated using tibial / impact accelerations. Lieberman et al. (2010) derived the following equation to estimate m\textsubscript{eff} using the time integral of F\textsubscript{z} i.e. impulse.

\[ \int_{T_2}^{T_1} F_z(t) = m_{body} (\Delta v_{CoM} + g \Delta T) = m_{eff} ((v_2 - v_1) + g (T_2 - T_1)) \quad (Eq. 14) \]
\( F_z(t) = \) time-varying vertical GRF; \( T_2 = \) time of impact transient; \( T_1 = \) time of TD; \( m_{\text{body}} = \) body’s mass; \( \Delta v_{\text{CoM}} = \) change in vertical speed of the body’s centre of mass; \( g = \) acceleration due to gravity; \( \Delta T = \) the change in time between \( T_1 \) and \( T_2; \) \( v_2 = \) vertical velocity of foot at \( T_2; \) \( v_1 = \) vertical velocity of foot at \( T_1. \) It also follows therefore, based on Newton’s second law of motion, that:

\[
\int_{T_1}^{T_2} F_z(t) = \int_{T_1}^{T_2} m_{\text{eff}}(a_{\text{impact}} + g)(t) \quad (\text{Eq. 15})
\]

\( a_{\text{impact}} = \) the impact acceleration. Rearrangement of equation 14 and 15 gives:

\[
m_{\text{eff}} = \frac{\Delta F_z}{(\Delta v + g \Delta T)} = \frac{\Delta F_z}{\Delta T(a_{\text{impact}} + g)} \quad (\text{Eq. 16})
\]

\( \Delta F_z = \) the change in \( F_z \) between \( T_1 \) and \( T_2; \) \( \Delta v = \) the change in velocity between \( T_1 \) and \( T_2; \) \( \Delta T = \) the change in time between \( T_1 \) and \( T_2. \)

Based on data from chapter 6, whereby \( \Delta T, \Delta F_z \) and foot strike patterns remained constant, but angular velocities were lower and RE improved after the 10wkRP, then \( m_{\text{eff}} \) may have decreased after 10 weeks of running (Eq. 16). It is important to note, however, that the angular velocities are assumed to be representative of the vertical velocity of the foot when making this conclusion. Moreover, whilst kinetic data was not recorded in chapter 9, the change in foot angle and thus foot strike, the knee being slightly more flexed, lack of footwear and higher impact accelerations indicate a decrease in \( m_{\text{eff}} \) as RE improved (BFT compared to SH). In the relationship shown in Eq. 15, Derrick argues that it is the \( m_{\text{eff}} \) that dominates. Consequently, it is conceivable that during the BFT running conditions that \( F_{z\text{impact}}, \) and thus \( \Delta F_z, \) reduced even though \( a_{\text{impact}} \) increased, supporting previous empirical evidence (Hamill, et al., 2011; Squadrone & Gallozzi, 2009).

The link between a lower \( m_{\text{eff}} \) and RE was not specifically addressed in this thesis and others have not previously investigated it. Therefore only a speculative connection can be drawn from the data available. As \( m_{\text{eff}} \) decreases the lower limb muscles have a smaller portion of the whole body’s mass to
control upon impact, yet the rate of deceleration is higher. Mechanical modelling studies have computed the effect of knee angle, strike index and heel pad deformation upon $m_{\text{eff}}$ (Chi & Schmitt, 2005; Denoth, 1986; Lieberman, et al., 2010), but no such modelling has been performed on the energetic requirements of $m_{\text{eff}}$. Such an investigation would be the first step in determining whether lowering $m_{\text{eff}}$ could contribute to an improved RE.

10.7.5 Transitioning to barefoot and minimalist shod running

While some progress is being made with regards to safe transition to BFT running, specifically concerning strengthening foot muscles (Goldmann, et al., 2013), fairly limited knowledge currently exists. Therefore investigations into different transition programs, which monitor biomechanical injury risk factors and bone edema (Ridge, et al., 2013), could lead to the development of an appropriate program. It is likely that a mixture of muscle strengthening and stretching exercises would be needed to complement the increase in BFT running volume, such as repetitive push-off activities (Goldmann, et al., 2013) and calf raises (Warne & Warrington, 2012).

Effective transitioning to either BFT or MS running could be aided by visual feedback. Such training has been shown to reduce impact accelerations and loading rates (Crowell & Davis, 2011), reported to be higher when BFT and MS running (De Wit, et al., 2000; Paquette, et al., 2013; Sinclair, et al., 2013a). Additionally gait re-training performed in this way can be successfully maintained in the long-term (Davis, et al., 2009). However it is not known whether gait training via visual feedback could alter the time spent loading the metatarsals or whether it would have positive or negative effects upon RE. Nevertheless it could be used, after identification of the optimal stride length, to aid stride length modifications (Eriksson, et al., 2011).

The influence that implementing such programs have on RE is an interesting area worth pursuing, as it would be able to identify the optimal transition program both for injury prevention and performance enhancement. To date two studies have considered a transition program to MS running, with Warne and Warrington (2012) primarily investigating physiological responses over 4 weeks and Lieberman et al. (2010) investigating foot strike patterns and kinematics over 6 weeks. Whilst Warne and Warrington (2012) stated that their minimalist
footwear was simulating BFT running, this may actually have not been the case given the recent evidence that MS running does not mimic BFT running (Bonacci, et al., 2013). Therefore future work should examine BFT and MS transitions, examining both the physiological and biomechanical responses, and considering how they influence each other to discover the most efficient program for each footwear condition.

10.8 Recommendations

The main finding from this thesis on self-optimisation is to run ‘relaxed’, exerting minimal effort. This refers to finding a running form that has a low metabolic cost, perhaps through a sub-conscious rather than conscious act. The recommendations below are based on the theoretical findings of the thesis. It is beyond the scope of the thesis to investigate how these could be practically implemented. Nonetheless, coaches and/or athletes could try to consider how this would be done and look to apply these strategies to their athletes’ and/or their own running gait:

- Produce a **less extended leg at TO** through either lower plantarflexion or less knee extension
- **Greater alignment of the leg axis and the resultant GRF** through greater propulsive force
- Generate **lower angular velocities** at the ankle and of the rearfoot
- **Lower thigh muscular coactivation** and rely more on passive internal structures for leg/joint stiffness
- Create a **smaller m_{eff} upon TD** by removing footwear, flexing the knee or having a more anterior foot strike (toes lower than heel)
- Increase leg stiffness during stance to help **lower vertical oscillation**
- Achieve stabilisation through **greater dependence on (predominantly) Type I muscles** rather than angular deflection or (predominantly) Type II muscles
10.9 Conclusions

In summary, the findings from this thesis have demonstrated that runners naturally self-optimise the way they run. This is seen both as an acute (changes in footwear) and short-term (10 weeks) response to changing running gait. Study one identified biomechanical changes to be responsible for the improvement in RE, with runners naturally producing a more economical movement pattern. Specifically, a less extended knee at TO, a later occurrence of peak dorsiflexion and lower eversion angular velocity at TD were identified as contributing to 94.3% of the variance in change in RE. Several other variables were significantly modified after 10 weeks of running, suggesting that less calf flexibility and greater propulsive force may indirectly influence RE. Study two demonstrated that economical runners appear to use different muscular strategies to less economical runners, with greater levels of proximal lower limb muscles (RFBF and VLBF) and extensor-extensor muscles (RFGL) associated with greater metabolic costs of running. Furthermore, study two revealed that the distal lower limb muscles (GLTA) and flexor-flexor muscles (BFTA) coactivate less with increasing speed due to a decrease in the activation duration of the TA muscle.

Study three highlighted the importance of familiarising individuals to BFT treadmill running prior to experimental manipulation of BFT running gait, such as that performed in study four. Although stride length was adopted instantaneously, several kinematics took longer to stabilise. Specifically, the knee and ankle angles at TD became more flexed and dorsiflexed, respectively. Based on these findings, it is suggested that future investigations should give participants 20 minutes of BFT treadmill running to allow running mechanics to adequately stabilise. Study four revealed that BFT running with a BFT stride length is more economical than SH or MS running. Several kinematic and muscular activity measures were shown to change with the removal of cushioning (SH to MS) and the heightening of proprioception (MS to BFT). Specific kinematics that appear to contribute to an enhanced RE when BFT are lower vertical oscillation, lower angular velocities and a lower $m_{eff}$. This study also underscores the importance of using an interdisciplinary approach to understand changes in RE, particularly the integration of both kinematics and muscular activity measures. Both study one and four suggest that runners adopt
specific movement patterns that may promote efficient storage and release of elastic energy, in addition to identifying the importance of the propulsive, push-off phase of stance to economical running. Furthermore it was evident from study four that economical running may be characterised by a greater dependence on efficient, Type I muscle fibres (TA rather than BF), which act to stabilise the lower limb. Consequently, there does not seem to be an increased metabolic cost to cushioning the body when BFT or MS when using this muscular activity strategy. Study four also supports the argument that individuals feel more comfortable running when the metabolic cost is low. Such a requisite ('comfort') is likely to be one of the subconscious driving forces behind any modifications to running gait, as runners naturally self-optimise. It is also important to note that economical biomechanical adjustments do not always favour a reduction in injury risk. In particular mechanical shock absorption (via pronation) appears to decrease in the most economical condition, in addition to runners experiencing higher impact accelerations and loading the forefoot for longer periods during stance. Such findings could have implications for overuse injuries.

Therefore, from the results of this thesis it can be concluded that running mechanics can be self-optimised to enhance RE. Furthermore, economical runners appear to use muscular strategies that contribute to a low \( \dot{V}O_2 \). Moreover, in the pursuit of improved performance through biomechanical adjustments it is likely that performance dominates rather than injury prevention, in terms of acute self-optimisation.
References


Arias, P., Espinosa, N., Robles-Garcia, V., Cao, R., & Cudeiro, J. (2012). Antagonist muscle co-activation during straight walking and its relation to...


Denadai, B. S., Ortiz, M. J., Greco, C. C., & de Mello, M. T. (2006). Interval training at 95% and 100% of the velocity at VO2 max: effects on aerobic physiological indexes and running performance. Applied Physiology, Nutrition, and Metabolism, 31(6), 737-743.


Scheuermann, B. W., Hoelting, B. D., Noble, M. L., & Barstow, T. J. (2001). The slow component of O(2) uptake is not accompanied by changes in
muscle EMG during repeated bouts of heavy exercise in humans. *Journal of Physiology*, 531(Pt 1), 245-256.


## Appendix A Table summary of the relevant literature investigating running mechanics and running economy

<table>
<thead>
<tr>
<th>Study</th>
<th>Participants</th>
<th>Methodology</th>
<th>Spatio-temporal</th>
<th>Flexibility</th>
<th>Measured Variables</th>
<th>Results and Conclusions</th>
</tr>
</thead>
<tbody>
<tr>
<td>Allison, Bailey &amp; Folland</td>
<td>10 male runners</td>
<td>10 min running at 70% VO\textsubscript{max} before and after prolonged static stretching, and no stretching intervention. Knee extensors, hip, knee and plantar flexors.</td>
<td>SL, SRT, knee extensor strength, counter-movement jump.</td>
<td>SL, stride rate, GCT, deceleration time, acceleration time, step rate.</td>
<td>SL, stride rate │ RE unchanged between conditions. Spatio-temporal variables unchanged. After static stretch intervention: SRT increased. Strength increased. Jump height decreased.</td>
<td></td>
</tr>
<tr>
<td>Ardigo, Lafortuna, Minetti, Mognoni &amp; Saibene (Ardigo, et al., 1995)</td>
<td>8 participants</td>
<td>All participants performed in two foot strike conditions: forefoot and rearfoot. Seven speeds were used, ranging from 2.5 to 4.17 m·s(^{-1}).</td>
<td>SL during 12\textsuperscript{th} and 20\textsuperscript{th} min of training.</td>
<td>SL, stride rate, GCT, deceleration time, acceleration time, step rate.</td>
<td>SL, stride rate, GCT, deceleration time, acceleration time, step rate.</td>
<td>RE unchanged between foot strike conditions. Step rate unchanged between foot strike conditions. Mechanical external work greatest for forefoot strike. Therefore greater efficiency during forefoot strike potentially due to greater storage and release of elastic energy. Decelerating time unchanged between foot strike conditions. GCT and accelerating time shorter for forefoot strike.</td>
</tr>
<tr>
<td>Bailey &amp; Messier (1991)</td>
<td>28 male collegiate students</td>
<td>Exp group: freely vary SL during training Con group: maintain a constant SL during training 3x20 min runs for 7wk.</td>
<td>SL during 12\textsuperscript{th} and 20\textsuperscript{th} min of training.</td>
<td>SL, stride rate, GCT, deceleration time, acceleration time, step rate.</td>
<td>SL, stride rate, GCT, deceleration time, acceleration time, step rate.</td>
<td>RE improved, however, absolute VO\textsubscript{2peak} remained unchanged. SL unchanged in Exp group. SL variations no effect on RE.</td>
</tr>
<tr>
<td>Beaudoin &amp; Whatley Blum (2005)</td>
<td>17 female collegiate athletes</td>
<td>Flexibility assessed and correlated with RE.</td>
<td>Trunk and hip rot, SRT, DF and PF range.</td>
<td>Trunk and hip rot, SRT, DF and PF range.</td>
<td>No relationships between RE and flexibility measures. Right hip rotation (r = -0.54) dismissed due to left side not presenting a similar result.</td>
<td></td>
</tr>
<tr>
<td>Bonacci et al. (2010)</td>
<td>17 moderately trained triathletes</td>
<td>Stride frequency, stride duration.</td>
<td>Sagittal plane: hip, knee and ankle</td>
<td>Sagittal plane: hip, knee and ankle</td>
<td>RF, BF, GL, TA</td>
<td>RE, kinematics and EMG unchanged between conditions. Ankle and knee angle at TD explained 73% of the variance in RE. Increasing DF and knee extension at TD associated with greater VO\textsubscript{2peak}. Individual level: 46% altered kinematics and EMG</td>
</tr>
<tr>
<td>Burkett, Kohrt &amp; Buchbinder (1985)</td>
<td>21 male runners</td>
<td>Treadmill running in three conditions: 1) BFT, 2) SH and 3) SH with orthotic.</td>
<td>GCT</td>
<td>Frontal plane patella movements</td>
<td>GCT</td>
<td>RE better in BFT running than SH with orthotic. Less angular displacement of the knee during BFT running than SH conditions. GCT shorter during BFT running than SH conditions.</td>
</tr>
<tr>
<td>Cavanagh, Pollock &amp; Landa (1977)</td>
<td>14 elite runners, 8 good runners</td>
<td>Comparing running mechanics of elite and good runners.</td>
<td>SL (m and %LL), stride rate, GCT, swing and flight time, VOSC</td>
<td>Hip, thigh, knee and ankle angles</td>
<td>Hip muscle torque</td>
<td>RE better in elite runners. Elite runners have: Higher stride rates (not significant) Less PF at TO Smaller VOSC (not significant)</td>
</tr>
<tr>
<td>Study</td>
<td>Participants</td>
<td>Conditions</td>
<td>Measurements</td>
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<tr>
<td>Cavanagh &amp; Williams (1982)</td>
<td>10 male well-trained runners</td>
<td>FCSL, ± 6.7%, ± 13.4% and ± 20% of FCSL</td>
<td>SL (cm and %LL)</td>
<td>Runners adopt a FCSL close to optimal. Mean deviation from optimal = 0.2 mL·kg⁻¹·min⁻¹</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Craib, Mitchell, Fields, Cooper, Hopewell and Morgan (1996)</td>
<td>19 male sub-elite distance runners</td>
<td>Flexibility assessed and correlated with RE.</td>
<td>Trunk rot, side bend, quad stretch, external rot of hip, SRT, hip flexion, straight leg raise, DF and PF range</td>
<td>RE positively correlated with hip rot and DF range. 47% of variation in RE explained by hip rot and DF range.</td>
<td></td>
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<tr>
<td>Dallam, Wilber, Jadellis, Fletcher &amp; Romanov (2005)</td>
<td>16 sub-elite triathletes (could run 10km in ≥ 42min)</td>
<td>Exp group: Pose method training for 12 weeks (1 hour sessions once a week) Con group: No training for 12 weeks</td>
<td>SL, stride rate, VOSC, GCT</td>
<td>Hip to ankle distance at foot strike</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Divert et al. (2008)</td>
<td>12 male runners</td>
<td>Exp conditions: BFT and diving socks loaded with added mass. Con Conditions: shoes with added mass.</td>
<td>VOSC, vel. of CoM, GCT, stride frequency</td>
<td>3D GRF and impulses, kₜₜ, kₐₕₐₜ</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Egbuonu, Cavanagh &amp; Miller (1990)</td>
<td>7 trained female runners</td>
<td>RE assessed at baseline, exaggerated VOSC (+3 SD) and arms behind back</td>
<td>VOSC</td>
<td>Compared to baseline, RE worse in both conditions.</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ferrauti, Bergermann &amp; Fernandez-Fernandez (2010)</td>
<td>22 experienced recreational runners (15 male, 7 female)</td>
<td>Exp group: 8 wk strength and endurance training Con group: 8 wk endurance training</td>
<td>SL (cm), Stride frequency, GCT</td>
<td>Leg and trunk MVC</td>
<td></td>
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<tr>
<td>Fletcher, Bartlett, Romanov &amp; Fotouhi (2008)</td>
<td>16 male recreational runners and triathletes</td>
<td>Exp group (n=8): Pose method training (7 x 1 hour sessions, on 7 consecutive days) Con group (n=8): Traditional running technique drills (7 x 1 hour sessions, on 7 consecutive days)</td>
<td>GCT, VOSC, SpL, step frequency, horizontal displacement of CoM, CoM velocity</td>
<td>Av. knee flexion angular velocity, Av knee extension angular velocity, Time occurrence and magnitude of peak braking, acceleration and vertical force</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Franck, Madsen, Djurhus &amp;</td>
<td>36 male recreational runners</td>
<td>Intensive distance training Long-interval Short-interval</td>
<td>SL, stride frequency</td>
<td>VL (muscle fibre composition)</td>
<td></td>
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</table>

Intensive distance or long-interval running more efficient than short-interval at improving RE and aerobic performance.
<table>
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<td>Pedersen (1998)</td>
<td>3 x 20-30 min per week for 6wk</td>
<td>( V'E ) Decreases correlated with RE improvements (( r = 0.77 )), accounts for 25-70% of decreased RE. Muscle fiber composition and running mechanics unaltered after training</td>
</tr>
<tr>
<td>Franz, Wierzbinski &amp; Kram (2012)</td>
<td>12 male runners, midfoot strikers Barefoot (with weights added) and minimalist shoes SL</td>
<td>3.3% shorter strides when barefoot. Barefoot running with added mass 3-4% greater RE, than shod running of equivalent weight.</td>
</tr>
<tr>
<td>Frost, Dowling, Dyson &amp; Bar-Or (1997)</td>
<td>30 children (10 7-8 yr olds, 10 10-12 yr olds, 10 15-16 yr olds) Four running speeds, varied for each age group and ranged from 1.79 – 2.73 m.s(^{-1})</td>
<td>VL, hamstring group (unspecified muscle), TA, SQ RE worse for youngest children. Thigh and shank coactivation highest in youngest children.</td>
</tr>
<tr>
<td>Gleim, Stachenfeld &amp; Nicholas (1990)</td>
<td>38 females and 62 males Flexibility assessed and correlated with RE.</td>
<td>RE inversely related to total flexibility (( r = -0.43 )), with a more flexible individual having a higher metabolic cost. Data split into three flexibility groups: Inflexible group better RE than flexible group.</td>
</tr>
<tr>
<td>Godges, Macrae, Longdon, Tinberg &amp; Macrae (1989)</td>
<td>7 males RE and hip ROM assessed prior to and after two conditions: 1) a static stretching protocol and 2) soft tissue mobilisation.</td>
<td>Hip ROM RE improved after static stretching (40, 60 and 80% ( V'O_{2\max} )). RE improved after soft tissue mobilisation (60% ( V'O_{2\max} )).</td>
</tr>
<tr>
<td>Godges, Macrae &amp; Engelke (1993)</td>
<td>25 male collegiate athletes 3 week stretching intervention either hip extension or trunk flexion</td>
<td>Hip extension, trunk flexion RE unchanged in either intervention. Flexibility improved in each intervention.</td>
</tr>
<tr>
<td>Hausswirth, Brisswalter, Vallier, Smith &amp; Lepers (2000)</td>
<td>7 male triathletes Isolated run for 45 mins (IR). Triathlon: 30 min swim, 60 min cycling, 45 min run (TR). Prolonged run for 2 hours 15 mins (PR).</td>
<td>VL, MVC of knee extensors RE worse in PR than TR and IR. RE worse in TR than IR. VL activity higher in PR than TR, but was not cause of changes in RE.</td>
</tr>
<tr>
<td>Hayes &amp; Walker (2007)</td>
<td>7 male distance runners 4 conditions completed by each participant prior to RE assessment: 1) Con, 2) static stretch, 3) progressive static stretch and 4) dynamic stretch</td>
<td>SRT RE unchanged in each condition. SRT improved after each experimental condition.</td>
</tr>
<tr>
<td>Heise &amp; Martin (1998)</td>
<td>16 recreational runners Treadmill physiological measures, over ground biomechanical measures. ( k_{sep}, k_{wert} )</td>
<td>RE inversely related to ( k_{sep} ) (( r = -0.48 )). More economical runners are less compliant. RE unrelated to ( k_{sep} ).</td>
</tr>
<tr>
<td>Study</td>
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<td>Heise, Morgan, Hough &amp; Craib (1996)</td>
<td>9 male athletes</td>
<td>EMG data recorded during RE assessment.</td>
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<tr>
<td>Heise &amp; Martin (2001)</td>
<td>16 male well-trained runners (10 km in 38-45 min)</td>
<td>Treadmill RE, over ground kinetics (3.35 m/s)</td>
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<tr>
<td>Heise, Shinohara &amp; Binks (2008)</td>
<td>16 female experienced runners</td>
<td>EMG data recorded during RE assessment.</td>
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<td>Hogberg (1952)</td>
<td>One subject (male), well-trained runner</td>
<td>4 speeds (14,16, 17 and 18km h), SL monitored and metronome used to change SL</td>
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<td>Jones (2002)</td>
<td>34 male international runners</td>
<td>Flexibility assessed and correlated with RE.</td>
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<td>Kelly, Girard &amp; Racinais (2011)</td>
<td>12 male recreational athletes</td>
<td>1 hour run at 10% higher than first ventilatory threshold with and without custom-moulded orthoses.</td>
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<td>Kyrolainen, Bell &amp; Komi (2001)</td>
<td>8 female middle-distance runners</td>
<td>Submaximal runs: 3.25, 4.00, 4.50 and 5.00 m(^{-1}), Maximal runs: 5.50, 5.75, 6.00 and 6.50 m(^{-1}) and all-out maximal run over 30 m.</td>
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<tr>
<td>Lake &amp; Cavanagh (1996)</td>
<td>17 male untrained runners (comfortably able to run for 30 mins at 3.36 ms(^{-1}))</td>
<td>Exp group (n=9): 6 wk running training. Weekly mileage increased from 15 miles to 20 miles. Con group (n=8)</td>
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<td>Lussiana, Fabre.</td>
<td>14 male recreational</td>
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<tr>
<td>McMahon, Valiant &amp; Frederick (1987)</td>
<td>6 males</td>
<td>Groucho training – running with increased knee flexion</td>
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<td>Messier &amp; Cirillo (1989)</td>
<td>22 female novice runners</td>
<td>Exp group (n=11): verbal and visual feedback on running technique (before and during each run). 15x20 min runs over 5 weeks. Con group (n=11): No feedback. 15x20 min runs over 5 weeks.</td>
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<tr>
<td>Mojock, Kim, Eccles and Panton (2011)</td>
<td>12 female long distance runners</td>
<td>Each participant completed each group. Exp group: static stretching prior to 60 min run. SRT, quad, PF and gluteus stretch. Con group: quite sitting prior to 60 min run.</td>
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<td>Morgan et al. (1994b)</td>
<td>9 (6 male and 3 female) recreational runners</td>
<td>Screening process to identify participants with a metabolically uneconomical freely chosen stride length (FCSL). FCSL, ± 6.7%, ± 13.4% and ± 20% of leg length from FCSL.</td>
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<tr>
<td>Morgan, Martin, Baldini &amp; Krahenbuhl (1990)</td>
<td>16 male trained runners</td>
<td>RE and gait assessed prior to and 1, 2 and 4 days after a prolonged maximal run.</td>
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<tr>
<td>Nigg, Stefanyshyn, runners</td>
<td>25 male runners</td>
<td>Two different shoes. Shoe A was medium hardness and</td>
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<td>Participants</td>
<td>Conditions</td>
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<tr>
<td>Cole, Stergiou &amp; Miller (2003)</td>
<td>mainly elastic. Shoe B was softer and more viscous.</td>
<td>(unspecified muscle), GM, TA</td>
</tr>
<tr>
<td>Nummela, Keranen &amp; Mikkelsson (2007)</td>
<td>25 male endurance athletes (10 distance runners, 8 orienteers and 7 triathletes)</td>
<td>Different speeds: 5, 5.4, 5.8, 6.2, 6.6 and 7 m s⁻¹ and maximal sprinting.</td>
</tr>
<tr>
<td>Perl, Daoud &amp; Lieberman (2012)</td>
<td>15 (13 male, 2 female) experienced BFT/MS runners</td>
<td>Compared RE and running mechanics during forefoot and rearfoot striking in MS and SH. Controlled SL.</td>
</tr>
<tr>
<td>Slawinski, Demarle, Koralsztein &amp; Billet (2001)</td>
<td>6 well-trained runners (gender unspecified)</td>
<td>8 wk supra threshold training, 2 x interval sessions and 3 x continuous sessions per week. Volume of interval sessions were increased during training.</td>
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<tr>
<td>Squadrone &amp; Gallozzi (2009)</td>
<td>8 male runners</td>
<td>Barefoot, minimalist shod and shod treadmill running</td>
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<tr>
<td>Svedenhag &amp; Sjodin (1994)</td>
<td>14 elite middle and 12 elite long distance runners</td>
<td>Maximal and submaximal (4 velocities) tests.</td>
</tr>
<tr>
<td>Thomas, Fernhall, Blanpied &amp; Stillwell (1995)</td>
<td>14 female distance runners</td>
<td>Submaximal 5 km run at 80-85% VO₂max. Gait assessed 5 min into run (G1) and 1 min prior to end of run (G2).</td>
</tr>
<tr>
<td>Tseh et al. (2008)</td>
<td>9 female recreational runners</td>
<td>4 running conditions: Normal (NL), hands clasped behind back (BK), hands clasped onto of head (HD) and exaggerated vertical oscillation (↑VOSC) (+4 SD)</td>
</tr>
<tr>
<td>Warne &amp; Warrington (2012)</td>
<td>4 week minimalist training</td>
<td>Stride rate</td>
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<tr>
<td>Williams &amp; Cavanagh</td>
<td>Male elite runners</td>
<td>Running mechanics assessed and correlated with RE.</td>
</tr>
</tbody>
</table>
Williams & Cavanagh (1987) 31 well-trained runners (gender not specified) 3 running groups: low, medium and high VO$_{2submax}$ GCT, strike index, SpL, step width, step time, VOSC trunk, hip, thigh, knee, ankle angles and velocities, wrist excursion CoP, Fx, Fy, Fz, energy transferred between segments of whole body and segments of lower extremity and trunk, mechanical power, power from CoM, net positive power muscle efficiency TD shank angle, TO PF and net positive power explain 54% of variance in RE. Compared to high and medium, low group had: lower F$_{zimpact}$, slower minimum knee flexion velocity during stance, greater trunk lean, less VOSC (non-significant), less plantarflexion at TO. High group had the worst energy transfer between the lower extremity and trunk.

Williams, Cavanagh & Ziff (1987) 14 female elite runners Running mechanics assessed and correlated with RE. SL, GCT, strike index, VOSC Hip, thigh, knee, abduction and ankle angles and velocities CoP, Fx, Fy, Fz RE inversely related knee extension at TO (r = -0.55), maximum DF angle (r = -0.60) and vel. (r = -0.59). Better economy associated with less extension, faster rate of DF and greater DF angle. RE positively related to maximal knee flexion vel. (r = -0.59). Better economy associated with less rapid rate of knee flexion.

Wilson et al. (2010) 10 male distance runners Participants performed two conditions. Exp condition: static stretching followed by 60 min run. Con condition: quite sitting followed by 60 min run. SRT Energy cost (calorie expenditure) greater in Exp condition. Distance covered greater in Con condition than Exp condition. SRT improved in Exp condition.

BF = biceps femoris; BFT = barefoot; CoM = centre of mass; Con group = control group; CoP = centre of pressure; DF = dorsiflexion; EMG = electromyography; Exp group = experimental group; FCSL = freely chosen stride length; F$_x$ = force in medio-lateral direction; F$_y$ = force in anterio-posterior direction; F$_z$ = force in vertical direction; F$_{zimpact}$ = vertical impact force; GA = gastrocnemius; GCT = ground contact time; GM = gastrocnemius medialis; GMax = gluteus maximus; GRF = ground reaction force; %LL = percentage of leg length; MS = minimalist shod; PF = plantarflexion; PL = peroneus longus; RE = running economy; RF = rectus femoris; rot = rotation; ROM = range of motion; Sh = shod; SL = stride length; SO = soleus; SpL = step length; SRT = sit-and-reach test; TA = tibialis anterior; TD = touchdown; TO = toe-off; vel = velocity; V$E$ = minute ventilation; VL = vastus lateralis; VM = vastus medialis; $\dot{V}O_2$ = oxygen consumption; $\dot{V}O_{2submax}$ = maximal oxygen consumption; $\dot{V}O_{2submax}$ = submaximal oxygen consumption; VOSC = vertical oscillation.
Appendix B: MATLAB script detailing the calculation used to compute knee and ankle joint moments

- Several functions were created to increase the speed of the script:
  [anthropo_coef; step_finding_fp; markers_unit_vectors]
- When running the script a loop was created so that all trials could be analysed at once

```matlab
clear all
close all

scrsz = get(0,'ScreenSize');

%% BASIC VARIABLES DEFINITION

%% gravity vector
g=-9.81; G=[0;0;-9.81];

%% FILTERS CHARACTERISTICS
% Double finite differentiation amplifies noise of measurement data
f_cut=20; % Cutoff frequency

% Butterworth filter characteristics for force data filtering
fp_sampling_rate=960; % sampling rate
[b_GRF_butter, a_GRF_butter] = butter(2,2*f_cut/fp_sampling_rate,'low');

% Butterworth filter characteristics for kinematic data filtering
kin_sampling_rate=120; % sampling rate
[b_kin_butter, a_kin_butter] = butter(2,2*f_cut/kin_sampling_rate,'low'); % Frequencies values are specified in normalized terms between 0.0 and 1.0, %where 1.0 corresponds to half the sampling frequency: %f/2 ie the Nyquist frequency

%% FILENAME DEFINITION
filename='Data/Post/pHBr5';

%% SUBJECT VARIABLES DEFINITION
% units body mass BM in kg, body height in cm
no_subj=7;
BM=70.5;
BW=BM*g;
BH=163.3;
correction_widths = importdata('Data/correction_widths.mat');
%foot: right or left?
rl_foot=filename(end-1);

%% ANTHROPOMETRIC PARAMETERS from Shan & Bohn 2003. coefficients definition
[segCOM_coef_foot,M_coef_foot,L_coef_foot,I_coef_foot,...
  segCOM_coef_shank,M_coef_shank,L_coef_shank,I_coef_shank]=anthropo_coef('f');

%% FORCE PLATE DATA
```
GRF_tmp=importdata(['filename ' 'VFC']);
fprintf('Reading Peak file %s 
', filename)

%% Peak Force data filtering
for ii=[1 2 3 9]
    GRF_tmp(:,ii)= filtfilt(b_GRF_butter,a_GRF_butter,GRF_tmp(:,ii));
end
free_Tq = Tq_min_adj(GRF_tmp(:,9));
GRF=GRF_min_adj(GRF_tmp(:,1:3));

%% Window for data calculation
%step detection force plate 2
loading_rate_fp1=diff(GRF)/-BW*960;
step_fp1=step_finding_fp(GRF,loading_rate_fp1);
TO_fp1=step_fp1+30+find(GRF(step_fp1+30:end)<10,1,'first');
step_matrix_fp1=step_fp1:TO_fp1;
stance_time=(TO_fp1-step_fp1)/960;
GRF=GRF(step_matrix_fp1,:);
l=length(step_matrix_fp1);
free_Tq=[zeros(1,l); zeros(1,l); free_Tq(step_matrix_fp1,:)';];

%%Centre of pressure data
COP=[GRF_tmp(step_matrix_fp1,7);GRF_tmp(step_matrix_fp1,8);zeros(1,l)];

%%%%% KINEMATIC DATA
%% interpolation to 960 Hz
xi=1/960:1/960:length(GRF_tmp)/960;
x=1/120:1/120:length(GRF_tmp)/960;

kin_tmp=importdata(['filename ' '3QS']);
[ll,col_kin]=size(kin_tmp);
for ii=1:col_kin
    kin_tmp(:,ii)= filtfilt(b_kin_butter,a_kin_butter,kin_tmp(:,ii));
end
kin=interp1(x,importdata(['filename ' '3QS']),xi,'cubic');
clear GRF_tmp

%% Markers data & unit vectors
ankle_width = correction_widths(1,no_subj);
[calc1,calc2,ach1,ach2,shin,med_knee,lat_knee,hip,Toe,mid_knee,...
    lat_mal,MTP5,ankle,forefoot,COM,foot_COM,...
    Z_foot,x_foot,y_foot,z_shank,x_shank,y_shank,...
    ankle_ach2,Toe_calc2,z_thigh]...
    =markers_unit_vectors(kin,ankle_width,rl_foot);

%% Method
% 1. Define the global frame and a local frame in each rigid body.
% 2. Define the free body diagram for each rigid body (Figure 3.3b) with constraint forces at
   the joints.
% 3. Choose the set of generalized coordinates to express the movement in.
% 4. Formulate the equations of motion for each segment.
% 5. Implement the force constraint equations directly in the equations of
   motion. These are simply (according to Newton’s third law):
% 6. Define the constraint equations

%%%%%
% ANKLE MOMENT CALCULATION

%% Calculation of subject's anthropometric parameters using above coef. (Shan & Bohn 2003)
y=b0+b1 * BM+b2 * BH. Mass in kg, body height in cm
M_foot=(M_coef_foot(1)+ M_coef_foot(2)*BM+M_coef_foot(3)*BH);
W_foot=M_foot*g;

% conversion in meters
L_foot=(L_coef_foot(1)+ L_coef_foot(2)*BM+L_coef_foot(3)*BH)*10^-2;
COM_coef_foot=segCOM_coef_foot*L_foot;

%% Preallocation of variables

for i=1:l
    % rotation matrix from global to local foot reference frame
    Rf(:,:,i)=[x_foot(step_matrix_fp1(i),1),x_foot(step_matrix_fp1(i),2),x_foot(step_matrix_fp1(i),3);...%
y_foot(step_matrix_fp1(i),1),y_foot(step_matrix_fp1(i),2),y_foot(step_matrix_fp1(i),3);...%
z_foot(step_matrix_fp1(i),1),z_foot(step_matrix_fp1(i),2),z_foot(step_matrix_fp1(i),3)];
    % Euler angles calculated from the rotation matrix of the foot
    % reference frame. 2 different sets of euler angles are calculated:
eul_f_y(i)=asin(Rf(3,1,i));
eul_f_x(i)=-atan2(Rf(3,2,i)/cos(eul_f_y(i)),Rf(3,3,i)/cos(eul_f_y(i)));
eul_f_z(i)=-atan2(Rf(2,1,i)/cos(eul_f_y(i)),Rf(1,1,i)/cos(eul_f_y(i)));
    % To choose the right set of orientation angles, we assume that
    % -pi/2 < eul_f_y < pi/2
    if eul_f_y(i)<-pi/2 && eul_f_y(i)>pi/2
        eul_f_y(i)=eul_f_y_1;
        eul_f_x(i)=-atan2(Rf(3,2,i)/cos(eul_f_y_1),Rf(3,3,i)/cos(eul_f_y_1));
        eul_f_z(i)=-atan2(Rf(2,1,i)/cos(eul_f_y_1),Rf(1,1,i)/cos(eul_f_y_1));
    else
        eul_f_y(i)=eul_f_y_2;
        eul_f_x(i)=-atan2(Rf(3,2,i)/cos(eul_f_y_2),Rf(3,3,i)/cos(eul_f_y_2));
        eul_f_z(i)=-atan2(Rf(2,1,i)/cos(eul_f_y_2),Rf(1,1,i)/cos(eul_f_y_2));
    end
end
end
% GRF expressed in local reference frame
GRF_ref_f(:,i)=Rf(:,:,i)*GRF(:,i);
% Foot COM after rotation of the referential: it will be used as the
% origin of the local reference frame
COM_foot_ref_f(:,i)=Rf(:,:,i)*COM_foot(:,i);
% Ankle centre of rotation in local reference frame
ankle_ref_f(:,i)=Rf(:,:,i)*ankle(step_matrix_fp1(i),:);
% Centre of pressure in local reference frame
COP_ref_f(:,i)=Rf(:,:,i)*COP(:,i);
% Free torque in local reference frame
free_Tq_ref_f(:,i)=(Rf(:,:,i)*free_Tq(:,i))';
end
% After rotation the centre of pressure still lies on the top of the force
% plate
COM_foot_1st_der=diff(COM_foot_ref_f,1,2)*960;
COM_foot_2nd_der=diff(COM_foot_ref_f,2,2)*960^2;
for ii=1:3
    COM_foot_2nd_der(ii,:)=filtfilt(b_kin_butter,a_kin_butter,COM_foot_2nd_der(ii,:));
end
% Vectors
ankle_2_COP_f=COP_ref_f-ankle_ref_f;
ankle_2_COM_f=COM_foot_ref_f-ankle_ref_f;
eul_f=[eul_f_z; eul_f_y; eul_f_x];
% Euler angle 1st derivative
eul_f_1st_der=diff(eul_f,1,2)*960;
eul_f_2nd_der=diff(eul_f,2,2)*960^2;
% Angular velocity in foot reference frame
wf=nan(3,l-1);
for i=1:l-1
    wf(1:3,i)=[0;0;eul_f_1st_der(3,i)+[cos(eul_f_z(i+1)) -sin(eul_f_z(i+1)) 0;sin(eul_f_z(i+1)) cos(eul_f_z(i+1)) 0;0 0 1]*[0;eul_f_1st_der(2,i);0]+...
             [cos(eul_f_z(i+1)) -sin(eul_f_z(i+1)) 0;sin(eul_f_z(i+1)) cos(eul_f_z(i+1)) 0;0 0 1]*[cos(eul_f_y(i+1)) 0 sin(eul_f_y(i+1)));0 1 0;sin(eul_f_y(i+1)) 0
             cos(eul_f_y(i+1))]*[eul_f_1st_der(1,i);0;0];
end
% Angular acceleration in foot reference frame
wwf=nan(3,l-2);
% Ankle moments components
Ma_GRF=nan(3,l-2);
Ma_weight=nan(3,l-2);
Ma_eff=nan(3,l-2);
Ma_rot=nan(3,l-2);
% Intersegmental force in the reference frame of the foot
Finter_ankle=nan(3,l-2);
% Ankle moment
for i=1:l-2
    wwf(1:3,i)=[0;0;eul_f_1st_der(3,i+1)]+...
end
\[-\sin(eul_f_z(i+2)), -\cos(eul_f_z(i+2)), 0; \cos(eul_f_z(i+2)), -\sin(eul_f_z(i+2)), 0; 0, 0, 0\]
\*\[eul_f_1st_der(2,i+1);0\]+...
\[-\cos(eul_f_z(i+2))*\sin(eul_f_y(i+2)), 0, \cos(eul_f_y(i+2))*\cos(eul_f_z(i+2)); -\sin(eul_f_y(i+2))*\sin(eul_f_z(i+2)), 0, \cos(eul_f_y(i+2))*\cos(eul_f_z(i+2)); 0, 0, 0\]
\*\[eul_f_1st_der(1,i+1);0;0\]+...
\[0;0;0;0;0;0\]
%% Moments due to rotational acceleration
% the inertia matrix of the body with respect to a frame rigidly
% attached to the segment considered remains constant: we use directly
% l_foot
% Euler's dynamic equations expressed in this frame

\[
\begin{bmatrix}
    I_{foot_{xyz}(1)} \cdot \omega(1,i) - (I_{foot_{xyz}(2)} - I_{foot_{xyz}(3)}) \cdot \omega(2,i) \cdot \omega(3,i) \\
    I_{foot_{xyz}(2)} \cdot \omega(2,i) - (I_{foot_{xyz}(3)} - I_{foot_{xyz}(1)}) \cdot \omega(3,i) \cdot \omega(1,i) \\
    I_{foot_{xyz}(3)} \cdot \omega(3,i) - (I_{foot_{xyz}(1)} - I_{foot_{xyz}(2)}) \cdot \omega(1,i) \cdot \omega(2,i)
\end{bmatrix}
\]
%% Intersegmental force bw foot and shank
\[
F_{inter_ankle(1:3,i)} = GRF_{ref_f(:,i+2)} - Rf(:,:,i+2) \cdot [0;0;W_{foot}] + M_{foot} \cdot COM_{foot_{2nd_der(:,i)}};
\]
%% Global ankle moment in ankle reference frame
% Moment of the GRF
\[M_{GRF(:,i)} = \text{cross(ankle}_2\_COP_f(:,i+2),GRF_{ref_f(:,i+2)})\];
% Moment of the Weight of the links
\[M_{weight(:,i)} = \text{cross(ankle}_2\_COM_f(:,i+2),Rf(:,:,i+2) \cdot [0;0;W_{foot}])\];
% Moments of the effective forces acting at the COM of the links
\[M_{eff(:,i)} = \text{cross(ankle}_2\_COM_f(:,i+2),M_{foot} \cdot COM_{foot_{2nd_der(:,i)}})\];
end
free_Tq_ref_f=free_Tq_ref_f(:,3:end);
fprintf(' Stance time: %f sec. 
', stance_time);
%% ANKLE MOMENT OUTPUT
\[M_{a}=\text{free_Tq_ref_f-M_{GRF-M_{weight}}+M_{eff-M_{rot}};\]
if rl_foot=='\text{l}''
\[M_{a}=\text{[M(1,:);M(2,:);M(3,:)]};\]
end

%%% KNEE MOMENT CALCULATION
%%% Calculation of subject's anthropometric parameters using above coef. (Shan & Bohn 2003)
y=b0+b1 \cdot BM+b2 BH. Mass in kg, length in cm
M_{shank}=M_{coef_{shank}(1)} + M_{coef_{shank}(2)} \cdot BM + M_{coef_{shank}(3)} \cdot BH;
W_{shank}=M_{shank} \cdot g;
%conversion in meters
L_{shank}=L_{coef_{shank}(1)} + L_{coef_{shank}(2)} \cdot BM + L_{coef_{shank}(3)} \cdot BH;*10^-2;
COM_{coef_{shank}}=segCOM_{coef_{shank}} \cdot L_{shank};
% I_tmp1= eul_f_xix components of the matrix of inertia
I_shank_xyz=[I_coef_shank(1,1)+I_coef_shank(1,2)*BM+I_coef_shank(1,3)*BH;
I_coef_shank(2,1)+I_coef_shank(2,2)*BM+I_coef_shank(2,3)*BH;
I_coef_shank(3,1)+I_coef_shank(3,2)*BM+I_coef_shank(3,3)*BH]*10^-4;

% Shank centre of mass definition in global reference frame
COM_shank=(ankle(step_matrix_fp1,:)+COM_coef_shank*z_shank(step_matrix_fp1,:))';

%% Preallocation of variables
eul_s_z=NaN(1,l);
eul_s_y=NaN(1,l);
eul_s_x=NaN(1,l);
Rs=NaN(3,3,l);
mid_knee_loc=NaN(3,l);
knee_ref_s=NaN(3,l);
ankle_ref_s=NaN(3,l);
COM_shank_loc=NaN(3,l);
COM_shank_R=NaN(3,l);
COP_ref_s=NaN(3,l);
COM_shank_ref_s=NaN(3,l);
COM_foot_ref_s=NaN(3,l);
x_shank_ref_f=NaN(3,l);
y_shank_ref_f=NaN(3,l);
z_shank_ref_f=NaN(3,l);
GRF_ref_s=NaN(3,l);
free_Tq_ref_s=NaN(3,l);

for i=1:l
    % Rotation matrix from global to local shank reference frame
    Rs(:,i)=[x_shank(step_matrix_fp1(i),1),x_shank(step_matrix_fp1(i),2),x_shank(step_matrix_fp1(i),3);
             y_shank(step_matrix_fp1(i),1),y_shank(step_matrix_fp1(i),2),y_shank(step_matrix_fp1(i),3);
             z_shank(step_matrix_fp1(i),1),z_shank(step_matrix_fp1(i),2),z_shank(step_matrix_fp1(i),3)];
    y_shank(step_matrix_fp1(i),1),y_shank(step_matrix_fp1(i),2),y_shank(step_matrix_fp1(i),3);
    z_shank(step_matrix_fp1(i),1),z_shank(step_matrix_fp1(i),2),z_shank(step_matrix_fp1(i),3)];

    % Euler angles calculated from the rotation matrix of the shank
    % reference frame. 2 different sets of euler angles are calculated:
eul_s_y_1=asin(Rs(3,1,i));
eul_s_y_2=pi-asin(Rs(3,1,i));
eul_s_x(i)=-atan2(Rs(3,2,i)/cos(eul_s_y_1),Rs(3,3,i)/cos(eul_s_y_1));
eul_s_z(i)=-atan2(Rs(2,1,i)/cos(eul_s_y_1),Rs(1,1,i)/cos(eul_s_y_1));
    else
        eul_s_y(i)=eul_s_y_2;
eul_s_x(i)=-atan2(Rs(3,2,i)/cos(eul_s_y_2),Rs(3,3,i)/cos(eul_s_y_2));
eul_s_z(i)=-atan2(Rs(2,1,i)/cos(eul_s_y_2),Rs(1,1,i)/cos(eul_s_y_2));
    end

    % GRF expressed in local reference frame
    GRF_ref_s(:,i)=Rs(:,i)*GRF(:,i);
    % Shank COM after rotation of the referential: it will be used as the
    % origin of the local reference frame
    COM_shank_ref_f(:,i)=Rf(:,i)*COM_shank(:,i);
COP_ref_s(:,i)=Rs(:,:,i)*COP(:,i);
COM_shank_ref_s(:,i)=Rs(:,:,i)*COM_shank(:,i);
COM_foot_ref_s(:,i)=Rs(:,:,i)*COM_foot(:,i);

% Knee centre of rotation in local reference frame
knee_ref_s(:,i)=Rs(:,:,i)*mid_knee(step_matrix_fp1(i),:);
ankle_ref_s(:,i)=Rs(:,:,i)*ankle(step_matrix_fp1(i),:);
free_Tq_ref_s(:,i)=Rs(:,:,i)*free_Tq(:,i);
end

% After rotation the centre of pressure still lies on the top of the force plate
COP_ref_s(3,:)=zeros(1,l);

% 1st and 2nd derivative of the shank COM
COM_shank_1st_der=diff(COM_shank_ref_s,1,2)*960;
COM_shank_2nd_der=diff(COM_shank_ref_s,2,2)*960^2;
COM_foot_2nd_der=diff(COM_foot_ref_s,2,2)*960^2;
for ii=1:3
    COM_shank_2nd_der(ii,:)=filtfilt(b_kin_butter,a_kin_butter,COM_shank_2nd_der(ii,:));
    COM_foot_2nd_der(ii,:)=filtfilt(b_kin_butter,a_kin_butter,COM_foot_2nd_der(ii,:));
end

% Vectors
knee_2_COP_s=COP_ref_s-knee_ref_s;
knee_2_COM_s=COM_shank_ref_s-knee_ref_s;
ankle_2_COM_s=COM_foot_ref_s-ankle_ref_s;

% Matrix of Euler angles in shank reference frame: eul_s
eul_s=[eul_s_z; eul_s_y; eul_s_x];
% Euler angle 1st and 2nd derivative
eul_s_1st_der=diff(eul_s,1,2)*960;
eul_s_2nd_der=diff(eul_s,2,2)*960^2;

% Angular velocity in shank reference frame ws
ws=nan(3,l-1);
for i=1:l-1
    ws(1:3,i)=[0;0;eul_s_1st_der(3,i)]+[cos(eul_s_z(i+1)) -sin(eul_s_z(i+1)) 0;sin(eul_s_z(i+1)) cos(eul_s_z(i+1)) 0;0 0 1]*[0;eul_s_1st_der(2,i);0]+...
        [cos(eul_s_z(i+1)) -sin(eul_s_z(i+1)) 0;sin(eul_s_z(i+1)) cos(eul_s_z(i+1)) 0;0 0 1]*[cos(eul_s_y(i+1)) 0 sin(eul_s_y(i+1));0 1 0;sin(eul_s_y(i+1)) 0;cos(eul_s_y(i+1))]*[eul_s_1st_der(1,i);0;0];
end

% Angular acceleration in shank reference frame wws
wws=nan(3,l-2);
Mk_rot=nan(3,l-2);
Mk_GRF=nan(3,l-2);
Mk_weight=nan(3,l-2);
Mk_eff=nan(3,l-2);
Finter_knee=NaN(3,l-2);
for i=1:l-2
    wws(1:3,i)=[0;0;eul_s_1st_der(3,i+1)]+...
        [-sin(eul_s_z(i+2)), -cos(eul_s_z(i+2)), 0; cos(eul_s_z(i+2)), -sin(eul_s_z(i+2)), 0; 0, 0, 0]...*
        [0;eul_s_1st_der(2,i+1);0]+...
        ...
    Mk_rot(i)=Mk_rot(i)+1;


\[-\cos(eul_s_z(i+2))\sin(eul_s_y(i+2)), 0, \cos(eul_s_y(i+2))\sin(eul_s_z(i+2));
-\sin(eul_s_z(i+2))\sin(eul_s_y(i+2)), 0, \cos(eul_s_y(i+2))\sin(eul_s_z(i+2));
-\cos(eul_s_y(i+2)), 0, -\sin(eul_s_y(i+2))\]

\n
\n
\[
[eul_s_1st_der(1,i+1);0;0]
+
[0:0:eul_s_2nd_der(3,i)]
+[
\cos(eul_s_z(i+2)) \sin(eul_s_z(i+2)) \sin(eul_s_y(i+2)) 0;\sin(eul_s_z(i+2)) \cos(eul_s_z(i+2)) \cos(eul_s_y(i+2)) 0;0 0 1]
+\cos(eul_s_y(i+2)) 0 \sin(eul_s_y(i+2));0 1 0;\sin(eul_s_y(i+2)) 0
\cos(eul_s_y(i))][eul_s_2nd_der(1,i);0;0];
\]

%% Moments due to rotational acceleration
% the Euler's inertia matrix of the body with respect to a frame rigidly
% attached to the segment considered remains constant: we use directly
% I_shank
% Euler's dynamic equations expressed in this frame
Mk_rot(1:3,i)=
[I_shank_xyz(1)*wws(1,i)
- (I_shank_xyz(2)
- I_shank_xyz(3))*ws(2,i)*ws(3,i);
I_shank_xyz(2)*wws(2,i)
- (I_shank_xyz(3)
- I_shank_xyz(1))*ws(3,i)*ws(1,i);
I_shank_xyz(3)*wws(3,i)
- (I_shank_xyz(1)
- I_shank_xyz(2))*ws(1,i)*ws(2,i)]+Ma_rot(1:3,i);

%% Intersegmental force bw shank and thigh in the shank reference frame
Finter_knee(1:3,i)=-GRF_ref_s(:,i+2)
- Rs(:,:,i+2)*[0;0;W_foot+W_shank]

% Moment GRF
Mk_GRF(:,i)=cross(knee_2_COP_s(:,i+2),GRF_ref_s(:,i));

% Moment of the Weight of the links
Mk_weight(:,i)=cross(knee_2_COM_s(:,i+2),Rs(:,:,i+2)*[0;0;W_shank])
+cross(ankle_2_COM_s(:,i+2),Rs(:,:,i+2)*[0;0;W_foot]);

% Moments of the effective forces acting at the COM of the links
Mk_eff(:,i)=
cross(knee_2_COM_s(:,i+2),M_shank*COM_shank_2nd_der(:,i))+(Rs(:,:,i)*Ma_eff(:,i));
end

free_Tq_ref_s=free_Tq_ref_s(:,3:end);

Mk=free_Tq_ref_s-Mk_GRF-Mk_weight+Mk_eff+Mk_rot;
if rl_foot=='l'
    Mk=[Mk(1,:);Mk(2,:);Mk(3,:)];
end

%% DISPLAY ANK KNEE MOMENTS IN MATLAB WINDOW AND ON A FIGURE
display_ank_knee_moments(Ma,Mk,fp_sampling_rate,scrsz,1)
Appendix C: Descriptions of the spatiotemporal, kinematic and kinetic parameters measured

<table>
<thead>
<tr>
<th>Variable</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step length (m)</td>
<td>Distance between successive foot strikes of the opposite foot</td>
</tr>
<tr>
<td>Vertical oscillation (m)</td>
<td>Vertical range of motion of the hip marker</td>
</tr>
<tr>
<td>Ground contact (s)</td>
<td>Time spent in stance</td>
</tr>
<tr>
<td>Peak knee flexion&lt;sup&gt;a&lt;/sup&gt; (º)</td>
<td>Maximum knee flexion during stance</td>
</tr>
<tr>
<td>TO knee extension (º)</td>
<td>Knee angle as the foot leaves the ground</td>
</tr>
<tr>
<td>Peak knee flexion velocity (º·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Maximum knee flexion velocity during stance</td>
</tr>
<tr>
<td>Peak knee extension velocity (º·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Maximum knee extension velocity during stance</td>
</tr>
<tr>
<td>TD dorsiflexion&lt;sup&gt;b&lt;/sup&gt; (º)</td>
<td>Ankle angle as the foot strikes the ground</td>
</tr>
<tr>
<td>Peak dorsiflexion&lt;sup&gt;a&lt;/sup&gt; (º)</td>
<td>Maximum ankle angle during stance</td>
</tr>
<tr>
<td>TO plantarflexion&lt;sup&gt;b&lt;/sup&gt; (º)</td>
<td>Ankle angle as the foot leaves the ground</td>
</tr>
<tr>
<td>TD plantarflexion velocity&lt;sup&gt;b&lt;/sup&gt; (º·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Ankle velocity as the foot strikes the ground</td>
</tr>
<tr>
<td>Peak plantarflexion velocity&lt;sup&gt;b&lt;/sup&gt; (º·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Maximum plantarflexion velocity during stance</td>
</tr>
<tr>
<td>TD inversion&lt;sup&gt;c&lt;/sup&gt; (º)</td>
<td>Rearfoot angle as the foot strikes the ground</td>
</tr>
<tr>
<td>Peak eversion&lt;sup&gt;a&lt;/sup&gt; (º)</td>
<td>Maximum rearfoot angle during stance</td>
</tr>
<tr>
<td>TO inversion&lt;sup&gt;c&lt;/sup&gt; (º)</td>
<td>Rearfoot angle as the foot leaves the ground</td>
</tr>
<tr>
<td>TD inversion velocity (º·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Rearfoot velocity as the foot strikes the ground</td>
</tr>
<tr>
<td>Peak inversion velocity (º·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Maximum inversion velocity during stance</td>
</tr>
<tr>
<td>Peak inversion velocity (º·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Maximum inversion velocity during stance</td>
</tr>
<tr>
<td>Peak medial force (N)</td>
<td>Peak $F_{ML}$ force towards the centre line of the body</td>
</tr>
<tr>
<td>Peak lateral force (N)</td>
<td>Peak $F_{ML}$ force away from the centre line of the body</td>
</tr>
<tr>
<td>Medial impulse (Ns)</td>
<td>Product of the time spent applying medial force and the force generated</td>
</tr>
<tr>
<td>Lateral impulse (Ns)</td>
<td>Product of the time spent applying lateral force and the force generated</td>
</tr>
<tr>
<td>Net $F_{ML}$ impulse (Ns)</td>
<td>The sum of the medial and lateral impulses</td>
</tr>
<tr>
<td>Peak propulsive force (N)</td>
<td>Peak $F_{AP}$ force in the direction of the run (positive force)</td>
</tr>
<tr>
<td>Peak braking force (N)</td>
<td>Peak $F_{AP}$ force in the opposing direction of the run (negative force)</td>
</tr>
<tr>
<td>Propulsive impulse (Ns)</td>
<td>Product of the time spent applying acceleration force and the force generated</td>
</tr>
<tr>
<td>Braking impulse (Ns)</td>
<td>Product of the time spent applying acceleration force and the force generated</td>
</tr>
<tr>
<td>Net $F_{AP}$ impulse (Ns)</td>
<td>The sum of the acceleration and braking impulses</td>
</tr>
<tr>
<td>Impact peak&lt;sup&gt;a&lt;/sup&gt; (N)</td>
<td>Initial $F_V$ peak prior to maximum peak</td>
</tr>
<tr>
<td>Active peak (N)</td>
<td>Maximum $F_V$ peak</td>
</tr>
<tr>
<td>Net $F_V$ impulse (Ns)</td>
<td>The total $F_V$ impulse during stance</td>
</tr>
<tr>
<td>Average loading rate (N·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Average loading rate value between 20 and 80% of the time to impact peak</td>
</tr>
<tr>
<td>Instantaneous loading rate (N·s&lt;sup&gt;-1&lt;/sup&gt;)</td>
<td>Peak loading rate value between 20 and 80% of the time to impact peak</td>
</tr>
</tbody>
</table>

<sup>a</sup> Time occurrence was calculated as a percentage of stance time
<sup>b</sup> Positive represents dorsiflexion and negative represents plantarflexion
<sup>c</sup> Positive represents inversion and negative represents eversion
Appendix D: Women’s Running Network beginner’s running programme

<table>
<thead>
<tr>
<th>Week</th>
<th>Session</th>
<th>Homework</th>
</tr>
</thead>
<tbody>
<tr>
<td>Week 1</td>
<td>15 mins - j:w:j</td>
<td>None</td>
</tr>
<tr>
<td>Week 2</td>
<td>6x15 sec fast bursts during 10 mins run</td>
<td>10-14 mins - j:w:j</td>
</tr>
<tr>
<td>Week 3</td>
<td>10 mins out, 10 mins back - j:w:j</td>
<td>10-14 mins run</td>
</tr>
<tr>
<td>Week 4</td>
<td>6x20 sec hill runs</td>
<td>15 mins - j:w:j</td>
</tr>
<tr>
<td>Week 5</td>
<td>25mins using loop backs (aim to cover 2.5-3km)</td>
<td>6x30 sec fast bursts during 15 mins run</td>
</tr>
<tr>
<td>Week 6</td>
<td>8x30 sec fast bursts during 12 mins out, 12 mins back - j:w:j</td>
<td>20 mins - j:w:j</td>
</tr>
<tr>
<td>Week 7</td>
<td>30 mins using loop backs (aim to cover 3.5-4km)</td>
<td>10 mins run with 6x30 sec hill runs, followed by 10 mins run</td>
</tr>
<tr>
<td>Week 8</td>
<td>Same as week 3, record distance to see improvement</td>
<td>20 mins run</td>
</tr>
<tr>
<td>Week 9</td>
<td>8x30 sec fast bursts during 30 mins run</td>
<td>24 mins run</td>
</tr>
<tr>
<td>Week 10</td>
<td>5km or timed loops of 3km/4km</td>
<td>5km event</td>
</tr>
</tbody>
</table>

j:w:j denotes jog interspersed with walk
Appendix E: Muscular coactivation as a percentage of stance for each muscle pair, over the three speeds across both days. The ICC, absolute SEM and relative SEM are presented, in addition to the day means (SD) and overall means for each muscle pair.

<table>
<thead>
<tr>
<th>Speed</th>
<th>Muscle pair</th>
<th>Day 1 mean (SD)</th>
<th>Day 2 mean (SD)</th>
<th>ICC</th>
<th>Overall mean (SD)</th>
<th>Absolute SEM</th>
<th>Relative SEM (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>RFBF</td>
<td>28.71 (12.33)</td>
<td>28.49 (11.05)</td>
<td>.684</td>
<td>28.60 (10.13)</td>
<td>7.29</td>
<td>23.99</td>
</tr>
<tr>
<td></td>
<td>VLBF</td>
<td>26.84 (7.47)</td>
<td>29.45 (10.98)</td>
<td>.533</td>
<td>28.15 (7.73)</td>
<td>4.65</td>
<td>15.86</td>
</tr>
<tr>
<td></td>
<td>GLTA</td>
<td>39.74 (11.69)</td>
<td>40.52 (7.38)</td>
<td>.786</td>
<td>40.13 (8.82)</td>
<td>3.85</td>
<td>9.85</td>
</tr>
<tr>
<td></td>
<td>RFGL</td>
<td>30.43 (14.91)</td>
<td>29.32 (9.72)</td>
<td>.635</td>
<td>30.22 (10.22)</td>
<td>9.54</td>
<td>21.62</td>
</tr>
<tr>
<td></td>
<td>BFTA</td>
<td>37.71 (13.28)</td>
<td>31.26 (14.96)</td>
<td>.428</td>
<td>34.49 (11.33)</td>
<td>8.57</td>
<td>24.85</td>
</tr>
<tr>
<td>2</td>
<td>RFBF</td>
<td>27.68 (12.48)</td>
<td>24.32 (11.33)</td>
<td>.892</td>
<td>26.01 (11.38)</td>
<td>4.51</td>
<td>16.48</td>
</tr>
<tr>
<td></td>
<td>VLBF</td>
<td>27.76 (10.73)</td>
<td>25.40 (11.76)</td>
<td>.140</td>
<td>26.58 (8.24)</td>
<td>7.55</td>
<td>28.60</td>
</tr>
<tr>
<td></td>
<td>GLTA</td>
<td>36.76 (9.89)</td>
<td>30.65 (11.86)</td>
<td>.541</td>
<td>33.71 (9.16)</td>
<td>7.25</td>
<td>22.06</td>
</tr>
<tr>
<td></td>
<td>RFGL</td>
<td>29.06 (11.47)</td>
<td>23.19 (11.47)</td>
<td>.700</td>
<td>26.13 (9.11)</td>
<td>4.99</td>
<td>19.11</td>
</tr>
<tr>
<td></td>
<td>BFTA</td>
<td>34.73 (13.73)</td>
<td>29.31 (15.53)</td>
<td>.483</td>
<td>32.02 (11.91)</td>
<td>8.56</td>
<td>26.74</td>
</tr>
<tr>
<td>3</td>
<td>RFBF</td>
<td>27.86 (11.62)</td>
<td>20.22 (8.38)</td>
<td>.606</td>
<td>24.04 (8.89)</td>
<td>9.23</td>
<td>38.47</td>
</tr>
<tr>
<td></td>
<td>VLBF</td>
<td>20.40 (10.06)</td>
<td>21.77 (12.95)</td>
<td>.255</td>
<td>21.08 (8.74)</td>
<td>7.57</td>
<td>33.04</td>
</tr>
<tr>
<td></td>
<td>GLTA</td>
<td>28.27 (14.82)</td>
<td>20.95 (10.66)</td>
<td>.506</td>
<td>25.07 (11.36)</td>
<td>7.12</td>
<td>28.34</td>
</tr>
<tr>
<td></td>
<td>RFGL</td>
<td>28.85 (9.15)</td>
<td>23.66 (9.56)</td>
<td>.309</td>
<td>26.26 (7.24)</td>
<td>6.02</td>
<td>22.93</td>
</tr>
<tr>
<td></td>
<td>BFTA</td>
<td>31.36 (19.89)</td>
<td>20.62 (13.03)</td>
<td>.273</td>
<td>25.99 (12.91)</td>
<td>11.70</td>
<td>45.01</td>
</tr>
</tbody>
</table>

RFBF = rectus femoris-biceps femoris; VLBF = vastus lateralis–biceps femoris; GLTA = gastrocnemius lateralis – tibialis anterior; RFGL = rectus femoris–gastrocnemius lateralis; BFTA = biceps femoris – tibialis anterior.
Appendix F: Means (SDs) of absolute muscular coactivations (ms) across each speed

<table>
<thead>
<tr>
<th>Variable</th>
<th>Speed 1</th>
<th>Speed 2</th>
<th>Speed 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>RFBF</td>
<td>86.0</td>
<td>60.5</td>
<td>43.4</td>
</tr>
<tr>
<td></td>
<td>(33.0)</td>
<td>(28.6)</td>
<td>(18.7)</td>
</tr>
<tr>
<td>VLBF</td>
<td>89.2</td>
<td>63.1</td>
<td>46.5</td>
</tr>
<tr>
<td></td>
<td>(34.5)</td>
<td>(30.0)</td>
<td>(27.3)</td>
</tr>
<tr>
<td>GLTA</td>
<td>122.9</td>
<td>76.1</td>
<td>45.0</td>
</tr>
<tr>
<td></td>
<td>(25.0)</td>
<td>(29.9)</td>
<td>(23.3)</td>
</tr>
<tr>
<td>RFGL</td>
<td>89.0</td>
<td>57.7</td>
<td>51.0</td>
</tr>
<tr>
<td></td>
<td>(32.5)</td>
<td>(22.2)</td>
<td>(22.7)</td>
</tr>
<tr>
<td>BFTA</td>
<td>94.6</td>
<td>73.0</td>
<td>44.3</td>
</tr>
<tr>
<td></td>
<td>(46.2)</td>
<td>(39.2)</td>
<td>(44.3)</td>
</tr>
</tbody>
</table>

A denotes significantly different to speed 1.
B denotes significantly different to speed 2.
C denotes significantly different to speed 3.
Appendix G: Means (SDs) of absolute (ms) and relative (% of stance) muscular activation times during stance across the three speeds

<table>
<thead>
<tr>
<th>Variable</th>
<th>Speed 1</th>
<th>Speed 2</th>
<th>Speed 3</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Absolute</td>
<td>Relative</td>
<td>Absolute</td>
</tr>
<tr>
<td>RF</td>
<td>177.25</td>
<td>58.55</td>
<td>141.83</td>
</tr>
<tr>
<td></td>
<td>(20.95)(^{A,C})</td>
<td>(5.66)</td>
<td>(13.38)(^{A,C})</td>
</tr>
<tr>
<td>VL</td>
<td>139.23</td>
<td>45.91</td>
<td>119.23</td>
</tr>
<tr>
<td></td>
<td>(33.50)(^{C})</td>
<td>(10.15)</td>
<td>(20.01)(^{C})</td>
</tr>
<tr>
<td>BF</td>
<td>165.06</td>
<td>54.85</td>
<td>132.36</td>
</tr>
<tr>
<td></td>
<td>(32.67)(^{B,C})</td>
<td>(11.20)</td>
<td>(31.66)(^{A,C})</td>
</tr>
<tr>
<td>GL</td>
<td>181.31</td>
<td>59.73</td>
<td>143.17</td>
</tr>
<tr>
<td></td>
<td>(29.74)(^{A,C})</td>
<td>(7.82)</td>
<td>(10.17)(^{A,C})</td>
</tr>
<tr>
<td>TA</td>
<td>155.60</td>
<td>51.50</td>
<td>110.94</td>
</tr>
<tr>
<td></td>
<td>(33.96)(^{B,C})</td>
<td>(11.36)(^{B,C})</td>
<td>(24.02)(^{A,C})</td>
</tr>
</tbody>
</table>

\(^{A}\) denotes significantly different to speed 1. \(^{B}\) denotes significantly different to speed 2. \(^{C}\) denotes significantly different to speed 3.