

**The effect of footwear on muscle activation during
exercise induced-fatigue**

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Abstract

This study investigates the effect of wearing two different types of running footwear: minimalist (MIN), 178g, versus shod, (SH), 349g on lower limb muscle activation during exercise-induced fatigue. Ten well-trained long-distance runners (aged 29.0 ± 7.5 ; BMI $23.6 \pm 6.5 \text{ kg m}^{-2}$; Volume oxygen uptake ($\text{VO}_{2\text{max}}$) $61.6 \pm 7.3 \text{ mL min}^{-1} \text{ Kg}^{-1}$) partook in the study. The participants ran seven (7 X 1000 m) high-intensity running intervals at speeds within 94 - 97% of their maximal aerobic capacity. These fatiguing trials were completed on a 200-m, unbanked Mondo surface indoor running track. Electromyography (EMG) was recorded from five lower limb muscles: biceps femoris, medial gastrocnemius, gluteus maximus, tibialis anterior, and vastus lateral. The integrated EMG (iEMG) and root mean square (RMS), EMG frequency and Stride Frequency were compared for the two running conditions. Findings reveal that the only statistically significant main effect of the shoe occurred in iEMG ($F_{(1,9)} = 6.137, P = .035$), and RMS ($F_{(1,9)} = .5.573, P = .043$) for the medial gastrocnemius as a function of wearing MIN. In addition, there was only a main effect of the shoe on the stride frequency ($F_{(1,9)} = 9.151, P = .014$) as a function of wearing MIN. The mean stride frequency values were higher in MIN conditions (92.800 stride/minutes) compared to SH conditions (90.600 stride/minutes). This study suggests that MIN footwear used in this study did not affect EMG amplitude (iEMG and RMS) or EMG frequency of lower limb muscles during exercise-induced fatigue when compared to SH due to low minimalist index. Additionally, results showed that MIN footwear increased stride frequency compared with SH footwear.

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Table of Contents:

Abstract	ii
Acknowledgments	iii
List of Tables	vii
List of Figures	viii
List of Abbreviations	ix
List of Appendices	x
Chapter 1: Introduction	1
Background of study	1
Chapter 2: Literature Review	6
2.1 Running Performance.....	6
2.1.1 Introduction to Running Performance.....	6
2.1.2 Muscle activation in running.....	7
2.1.3 Factors influencing muscle activation during running.....	8
2.1.3.1 Footwear.....	8
2.1.3.1.1 Minimalist	10
2.1.3.1.2 Barefoot.....	10
2.1.3.1.3 Shod Shoes	11
2.1.1.4. Muscle activation during different shoe conditions	12
Tibialis anterior.....	12
TA activation in BF running vs. SH	13
Calf muscle activation	15
BF vs. SH.....	15
Soleus.....	18
Gluteus Maximus.....	19
Peroneus longus	19
Biceps femoris	20
Rectus Femoris	21
Vastus lateralis.....	21
Vastus Medialis	23

MIN vs. SH running	23
2.1.1.5 Summary of footwear and muscle activation	24
2.1.3.2. Running speed	25
2.1.3.3 Different running surfaces (Treadmill vs. Overground Running).....	26
2.1.3.4 Effects of fatigue on muscle activation	28
2.1.3.5 Effects of fatigue on muscle activation during running	31
2.1.4 Does footwear alter the effect of fatigue on muscle activation during running? .	41
Chapter 3: Methods: Materials and Methods	46
3.1 Participants	46
3.2 Laboratory visits.....	48
3.3 EMG measurements	49
3.4 Shoes	49
3.5 Experimental Protocol.....	50
3.6 Data analysis	51
3.6.1 EMG amplitude	51
3.6.2 EMG Frequency	52
3.6.3 Stride Frequency.....	52
3.7 Statistical analysis	53
Chapter 4: Results:.....	55
4.1 Participants' characteristics and training regime	55
4.2 Fatigue confirmation	58
4.3 EMG Amplitude.....	59
4.4 EMG Frequency:	63
4.4.1 Median Frequency and Mean Frequency:	63
4.5 Stride Frequency	64
Chapter 5: Discussion	67
5.1 Confirmation of fatigue onset	67
5.2 The effects of fatigue on muscle activation	69
5.3 The interaction of footwear and fatigue	73
5.4 Speed	75
5.5 Limitations	78

5.6 Conclusion.....	79
References.....	81
Appendix: Borg 6-20 Rate of perceived exertion (RPE) scale.....	111

List of Tables

Table 1: A summary of the participants' characteristics.....	47
Table 2a: A summary of the participants' interval durations (minutes) during first, middle and last interval for SH condition.	56
Table 2b: A summary of the participants' interval durations (min/s) during first, middle and last interval for MIN conditions.....	57
Table 3a: Statistical results for root mean square EMG of lower limb muscle activation.....	59
Table 3b: Statistical results for integrated EMG of lower limb muscle activation.....	61
Table 4a: Statistical results for Median Frequency EMG of lower limb muscle activation.....	63
Table 4b: Statistical results for Mean Frequency EMG of lower limb muscle activation.....	64

List of Figures

Figure 1. RMS medial gastrocnemius EMG as a function of shoe type (MIN vs. SH.) collapsed across the first and last minute of the running intervals.....	60
Figure 2. Average medial gastrocnemius integrated EMG (iEMG) as a function of shoe type (MIN vs. SH.) collapsed across the first and last minute of the running intervals...	62
Figure 3. Sample data for the stride frequency using EMG data.....	65
Figure 4. Stride Frequency as a function of shoe type (MIN vs. SH.) collapsed across the first and last running intervals.....	66

List of Abbreviations

ANOVA	Analysis of variance	RPE	Rate of perceived exertion
BiF	Biceps femoris Muscle	SD	Standard deviation
BF	Barefoot (running/shoes)	SH	Shod running shoes
CV	Coefficient of variation	SOL	Soleus Muscle
EMG	Electromyography	TA	Tibialis anterior muscle
Gmax	Gluteus maximus muscle	VL	Vastus lateralis muscle
HR	Heart rate	VM	Vastus medialis muscle
HRpeak	Heart rate Peak	VO _{2max}	Volume oxygen uptake
iEMG	Integrated Electromyography		
LG	Lateral gastrocnemius muscle		
MG	Medial gastrocnemius muscle		
MIN	Minimalist footwear		
MVC	Maximum voluntary contraction		
PL	Peroneus longus muscle		
RF	Rectus femoris muscle		
RMS	Root mean square		

List of Appendices

Appendix: Borg 6-20 Rate of perceived exertion (RPE) scale	111
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Chapter 1: Introduction

Background of study

Running is a physical activity that many people across the world practice and enjoy. Research indicates that running has numerous health benefits, such as improved cardiovascular function, increased bone mass, weight maintenance, lowered potential for hypertension and stroke, and heightened positive mood (Shipway & Holloway, 2010; Hanson, Berg, Deka, Meendering, & Ryan, 2011). Despite these benefits, it is estimated that annually between 37% -79% of people who engage in running suffer running-related injuries (van Gent, Siem, van Middelkoop, van Os Bierma-Zeinstra, & Koes, 2007).

Other research estimates that annual injury percentages range between 50% -79% (Tam, Coetzee, Ahmed, Lamberts, Albertus-Kajee, & Tucker, 2017) or even from 39% - 85% (Bovens, Janssen, Vermeer, Hoerberigs, Janssen, & Verstappen, 1989; Watson & DiMartino, 1987). No research indicates annual injury levels of less than 37%, which means that over one-third of runners experience an injury every year. Over 50%) of these running injuries involve the knee, and the most common injuries occur in the knee joint (Rolf, 1995; Taunton, Ryan, Clement, McKenzie, Lloyd-Smith, & Zumbo, 2003).

Moreover, research shows that long-distance runners suffer the most injuries, mainly caused by chronic overuse of lower extremities (Cauthon, Langer & Coniglione, 2013). In their study published in 1982, Koplán and colleagues discovered that over one-third of both male and female runners who frequently participated in 10 km races had experienced some type of running injury in the previous year (Koplán, Powell, Sikes,

Shirley, & Campbell, 1982). In 1986, Jacobs and Berson set the incidence of injury rate at just under 50 %, citing ankle and knee pain as well as shin splints as the main complaints (Jacobs et al., 1986). More recently, research published by Van Middelkoop and colleagues revealed that the incidence of injury was 54.8 % only for male runners in the Rotterdam marathon that was held in 2005. For those runners, injuries primarily affected their feet, knees, and calves (Van Middelkoop, Kolkman, Van Ochten, Bierma-Zeinstra & Koes, 2008).

Sophisticated changes in running shoe technology are aimed at preventing injuries to runners. Both medical professionals and the sports shoe industry encourage runners to consider the footwear they use and customize it for their needs (Cheung & Rainbow, 2014). Nevertheless, there is very little proof that heavily cushioned running shoes prevent, or even mitigate running-related injuries (Knapik et al., 2010a). Additionally, despite improvements to running shoes, rates of injury have remained more or less constant over the years (Cauthon et al., 2013). Furthermore, there is a growing trend for runners to wear minimalist footwear (MIN) rather than today's complex, sophisticated and increasingly expensive running shoes, which might actually exacerbate, or even cause running injuries (Lieberman, Venkadesan, Werbel, Daoud, D'Andrea, Davis, & Pitsiladis, 2010). Despite the fact that running in MIN footwear results in different running biomechanics compared to shod footwear (SH) (Bonacci et al., 2013), switching from SH footwear to MIN footwear has become common (Lussiana, Hébert-Losie & Mourot, 2015). At present research is still inconclusive as to the benefits of wearing MIN footwear over more traditional running shoes.

It has been estimated that around one-third (31-34%) of people who run on a regular basis are now wearing MIN or running BF (Cohler, & Casey, 2015; Rothschild, 2012). According to some research, this approach leads to an increased stride frequency, decreased stride length and altered joint angles (Bonacci et al., 2013; De Wit, De Clercq, & Aerts, 2000; McCallion, Donne, Fleming, & Blanksby, 2014). Many runners have embraced the latest low-tech footwear, hoping that innovations that incorporate impact regulations and protective mechanisms will help decrease the injury rate (Cohler et al., 2015).

Several researchers still believe that they cannot definitively claim that MIN reduces the rate of injury in regular runners (Bergstra, Kluitenberg, Dekker, Bredeweg, Postema, Van den Heuvel, & Sobhani, 2015; Perkins, Hanney, & Rothschild, 2014; Ryan, Elashi, Newsham-West, & Taunton, 2014). On the contrary, there have been several research studies reporting that MIN footwear would actually cause some changes in running biomechanics rather than running in SH, such as an increase of the loading in lower limb parts (Willy & Davis, 2013). MIN footwear may also increase peak plantar pressure under almost the entire foot (Bergstra et al., 2015), and joint power (Bonacci et al., 2013). An additional issue caused by MIN shoes or BF running is stress fractures (Cauthon et al., 2013; Ridge et al., 2013; Salzler, Bluman, Noonan, Chiodo & de Asla, 2012).

A limitation of many footwear-related studies is that they examine runners under non-exertion conditions, even though footwear is more critical to performance (and overuse injury) if runners are strongly over-exerting themselves (Butler, Hamill, & Davis,

2007). There is a need for evidence-based conclusions and guidelines on how footwear affects a runner's experience of fatigue, especially during the end-stages of running. A recent research study explored fatigue in relation to running exertion and footwear by measuring kinematic, kinetic and neuromuscular changes. The findings showed that runners who typically wore running shoes risked injury if they ran BF while experiencing more extreme levels of fatigue (Tam et al., 2017). This advocates that fatigue is an essential factor to consider when investigating the effects of footwear on running. Fatigue can prevent a runner's body from appropriately dealing with the muscular/mechanical changes that occur during running. Despite the potential influence that fatigue may have on running performance and injury risk, there is almost no literature that has examined how footwear may alter the effect of fatigue during running. The author is particularly interested in how muscle activation is affected. The purpose of this thesis was to investigate the effect of fatigue on lower limb muscle activation two different types of running footwear (MIN & SH).

The data used for this thesis were collected as part of a larger study that examined the effects of footwear on running economy (Blair 2018). The author assisted with data collection for the Blair study and subsequently took a portion of that data and used it to complete the current thesis. Blair's study consisted of three main parts - pre-fatigue trials, exercise-induced fatigue trials and post-fatigue trials. The pre and post-fatigue trials were used by Blair to assess the effects of footwear on running economy. This data, along with fatigue markers (rate of perceived exertion, peak heart rate and blood lactate) and running speed during the exercise-induced fatigue trials, were all reported in Blair's thesis (Blair,

2018). The present thesis is based on an analysis of muscle activation data collected during the exercise-induced fatigue trials only. These data were not examined by Blair and represent a primary examination. In addition to the muscle activation data, running speed and fatigue markers (as reported by Blair, 2018) are also described and used in the present study as they were needed to assist with data interpretation and discussion of results.

Chapter 2: Literature Review

2.1 Running Performance

2.1.1 Introduction to Running Performance

In addition to the use of technology such as motorized vehicles, people generally move using two distinct forms of movement, walking and running, which share similar core kinetics and kinematics (Cappellini, Ivanenko, Poppele, & Lacquaniti, 2006). Unlike driving a car, neither form of ambulation requires any extra special skills and can be practiced by most humans from a very young age. Nevertheless, these activities still require the cooperation of several different muscle groups and input from the nervous system, the cardiovascular system, etc., all working together towards the goal of human locomotion (Slocum & James, 1968). A recent heightened interest in jogging, running, and running-related sports has inspired a new field of running-related research. Sports shoe manufacturers also want to appeal to this growing market, so there is a great deal of research being done on ways to optimize running performance through footwear.

Another topic of concern that led to an increase in research is the rising number of running-related injuries. Although running is a fairly standard mode of travel, repetitive running for exercise or competition demands an understanding of the body's biomechanics and pathophysiology. Knowledge about the biomechanical factors that affect the body during running is useful for two main reasons: (1) knowing and applying optimal running mechanics could increase performance outcomes, and (2) a basic understanding of running-related injury mechanisms could lower the risk of injuries (Novacheck, 1998; Williams, 1985). Assessing biomechanical factors could

also reveal a body's abnormal loads, thereby helping decrease potential risk factors for injury (McClay & Manal, 1999). Having a basic understanding of the biomechanics of running could not only improve runner performance, but also help prevent common running-related injuries (Taunton, Ryan, Clement, McKenzie, Lloyd-Smith, & Zumbo, 2002).

The amount of oxygen utilized by runners at standard their running speed is referred to as "running economy" (McLaughlin, Howley, Bassett Thompson, Fitzhugh, et al., 2010). Runners who use lower amounts of oxygen at a certain running speed (economical runners) are likely better suited to long-distance running activities than runners who consume higher amounts of oxygen (McLaughlin et al., 2010). Over the years, researchers have investigated the energy costs for walking and running (e.g., Refs, Di Prampero, 1986; Margaria, Cerretelli, Aghemo, & Sassi, 1963). It has been stated that running requires more oxygen consumption than walking when performed above 2m/s (Margaria, 1976). To explore the ways in which energy might be conserved during running, we first need to understand how energy is consumed during running. More details about this topic can be found in other publications (e.g., Refs, Margaria, Cerretelli, Aghemo, & Sassi, 1963).

2.1.2 Muscle activation in running

For muscle contraction to produce movement, there is a need for neural activity to be transformed into actions or movements in the environment. Movement is not only affected by the nervous system, but also by the inertia, loading and biomechanical changes of body parts when they are used (Watson & Ritzmann, 1997). Knowing more

about muscle activation of the lower limb muscles when running is crucial for optimizing human performance and preventing lower limb injuries. Because the lower limb muscles make a significant contribution to running performance, they tend to be more active during running than walking (Ellis, Sumner & Kram, 2014); however, there are many factors that influence the amount of muscle activation during running.

2.1.3 Factors influencing muscle activation during running

2.1.3.1 Footwear

According to the most recent research, the type of footwear worn during running has a significant effect on lower extremity kinematics and kinetics (Butler et al., 2007). Some of this research on impact forces, therefore, involves not only the loading rate, time and magnitude, but also the musculoskeletal system's muscular responses and reactions (Brüggemann, Brüggemann, Heinrich, Müller, & Niehoff, 2011; Wang, Zhang & Fu, 2017). During the process of running, lower limb muscles must offer sufficient propulsion, joint positioning, stiffness, and stability to push the body forward. Therefore, excessive loading of the lower extremities will cause alterations in muscle activation (Ervilha, Mochizuki, Figueira, & Hamill, 2017). A wide range of studies on muscle activation has examined activation of the rectus femoris (RF), biceps femoris (BiF), gluteus maximus (Gmax), (TA), vastus lateral (VL), soleus (SOL), and peroneus longus (PL), as well as MG and lateral gastrocnemius (LG) (e.g. Refs, Divert, Mornieux, Baur, Mayer and Belli, 2005; Komi, Golihofer, Schmidtbleicher, & Fric 1987).

In general, a great deal of research has investigated how muscle activation is altered by wearing different kinds of running footwear. The research has mainly looked at electromyography (EMG) alterations with respect to timing (Ervilha et al., 2017; Von Tscharnner et al., 2003) and amplitude (Ervilha et al., 2017; Kasmer, Ketchum & Liu, 2014; Khowailed, Petrofsky, Lohman, & Daher, 2015; Olin & Gutierrez 2013; Von Tscharnner et al., 2003) as well as changes that can occur with regard to the co-activation of lower-limb musculature (Ervilha et al., 2017). In spite of the general agreement in the above mentioned research in relation to alterations to muscle activation, a small number of studies indicate that either few or no alterations occur in the timing (Ahn, Brayton, Bhatia & Martin, 2014) or magnitude (Burke & Papuga, 2012). It is possible that the wide diversion in findings could be due to the types of footwear worn during the research trials.

Research into footwear cushioning has shown how the muscle activation of lower limbs is “tunable” by applying different kinds of midsole materials and hardness to absorb the impact force during the heel strike (Butler et al., 2007; Wakeling, Pascual & Nigg, 2002). For example, Nigg and Wakeling (2001) demonstrated alterations in the EMG of vastus medialis and hamstrings when runners wore two different running shoes, one with a medium-hard heel and the other with a soft, viscous heel. The alterations in EMG were evident in how the different footwear affected the input signals (Nigg, & Wakeling, 2001). An overview is provided of these three commonly used approaches to running footwear (MIN, SH and BF) in the next section, followed by an examination of the effect of wearing them on muscle activation during running.

2.1.3.1.1 Minimalist

MIN footwear is defined as a type of shoe that has the following characteristics: zero arch support, ultra-thin flexible soles, and heels that are generally 0 to 4 mm thicker than the forefoot midsole (Fleming, Walters, Grounds, Fife, & Finch, 2015; Gavilanes-Miranda, De Gandarias & Garcia, 2012). MIN footwear is also known as the type of footwear that provides the experience of natural movement with the lack of motion control and stability features, and also provides high flexibility, high stack, and low weight (Esculier, Dubois, Dionne, Leblond & Roy, 2015). There are three categories of MIN: 1) low or reduced heel-to-toe drop footwear, such as New Balance®, Minimus™ or Vibram®; 2) footwear featuring a somewhat thicker sole and minimal cushioning, such as Nike® Free™; and 3) footwear modelled after traditional racing flats but still referred to as “MIN”, such as Brooks® Pure™ (Squadrone, Rodano, Hamill and Preatoni, 2015; Sernjuk, 2016). The purpose of MIN is to be as thin as possible while still providing protection to the runner’s feet and giving the runner the impression of BF without the risk of injuries that might occur when running BF (Bowles, Ambegaonkar, Cortes, & Caswell (2012). Therefore, MIN footwear was developed to imitate the style of BF running, yet the specific mechanics used are not well known when wearing these types of running shoes (Standifird, Mitchell, Hunter, Johnson and Ridge, 2013).

2.1.3.1.2 Barefoot

BF running is the act of running without wearing footwear, but BF running also includes running while wearing foot coverings such as compression, diving, or yoga socks (Divert, Mornieux, Freychat, Baly, Mayer and Belli, 2008; Fleming et al., 2015).

The advantages and disadvantages of BF running have been a subject of interest for runners and the research community (Hein & Grau, 2014). Due to the forefoot-strike contact pattern that typically occurs, BF running has gained much fame (Lieberman, 2012). Running BF or in conventional track racing flats result in a running technique whereby the front part of the foot hits the ground first, followed by the heel and next, a toe-off with the forefoot (Mullen, Cotton, Bechtol, & Toby, 2014). BF running is linked to smaller strides and higher cadence (Altman & Davis, 2012). Reduction of stride length results in less load felt by the body and may prevent the runner from impact-related injuries (Edwards, 2009).

2.1.3.1.3 Shod Shoes

SH footwear is known to have more cushioning material for providing protection, support, comfort, and correct movement patterns (Altman et al., 2012; Divert et al., 2005; Warne & Warrington, 2014). The forefoot midsole of SH running footwear is approximately 11 mm thinner than the heel (Fleming et al., 2015; Gavilanes et al., 2012). Research shows that wearing SH (i.e., arch-supported or motion-controlled) footwear during running prevents feet from functioning correctly. Some authors believe that although SH footwear might give the illusion of enhanced support and stability, the reality is that it prevents the longitudinal arch from “flattening” during standing, which then lessens the impact absorption capacity of the feet (Bowles, Ambegaonkar, Cortes, & Caswell, 2012). This inhibition of movement can lead to atrophy in foot muscles due to not needing to use different muscles for different ground types (e.g., hard, soft, sloping). As a result, runners may not change their stride to suit the real (rather than the perceived)

environment, which may magnify the load that joints must deal with (Goble, Wegler & Forest, 2013). It is logical to believe that some running injuries are associated with the impact stage when the body collides with the surface, as there is no scientific proof verifying that SH footwear reduces injuries or improves long-term health in runners (Rixe, 2012).

Considering that most runners wear SH shoes while running, many scientific studies have investigated the influence of the shoes' properties on one's style of running (De Wit, De Clercq & Aerts, 2000). It is highly likely that the heavily cushioned footwear that runners wear today has altered how people run to a form that clashes with the way our bodies have evolved to deal with running stresses (Davis, 2017). During running, the lower extremities provide reasonable stability, joint positioning, stiffness, and propulsion in order to move the body. The main functions of running footwear are to ease the strength of action that is exerted by runners on the ground in addition to absorbing the reaction force in order to protect the muscular and skeletal systems (Gavilanes-Miranda et al., 2012). However, when footwear such as SH footwear contains more cushioning, there is more damping of the impact force (Gavilanes-Miranda et al., 2012). Cushioning and stability might conflict since more cushioning can result in less stability (Perry, 1992).

2.1.1.4. Muscle activation during different shoe conditions

Tibialis anterior

When running in SH running shoes using rearfoot strike, TA has two primary roles. Just prior to heel strike, TA is activated to place the foot in a dorsiflexed position to

prepare for the foot strike. Immediately following a heel strike, TA contracts eccentrically to control ankle plantar flexion and prevent the foot from slapping on the ground (Inman, 1969; Von Tscharner, Goepfert, & Nigg 2003). Research examining how TA differs in MIN and/or BF running has produced conflicting results, with some researchers reporting decreased TA activation during BF running (Lucas-Cuevas et al. 2016; Snow, Basset & Byrne 2015; Strauts, Vanicek & Halaki., 2016; Von Tschar et al., 2003; Olin and Gutierrez 2013). Moreover, some reported no effect (Ervilha et al. 2017; Lindberg and Olle, 2012) and still others have reported increased TA activation (Von Tscharner et al., 2003; Da Silva et al., 2016). To fully appreciate the effects of footwear on TA activation, this paper reviews these studies in greater detail.

TA activation in BF running vs. SH

Although there are some discrepancies in the literature, by far the most common finding is that TA activation decreases when running BF compared to SH (Lucas-Cuevas et al. 2016; Snow et al., 2015; Strauts et al., 2016; Olin and Gutierrez 2013; Von Tscharner et al., 2003). All five of the studies that have determined this have examined treadmill running. Lucas-Cuevas et al. (2016) had runners run at 60% of their maximal aerobic speed in one of three conditions (natural BF, BF with a forced heel-strike, and SH). Activation of TA was reduced in the BF condition compared to the other two. The work of Strauts et al. (2016) examined the acute effects of a short bout of BF running. Research has shown decreased TA activation at several intervals throughout the running stride, and Fleming et al. (2015) reported similar findings in their examination of the influence of BF running on muscle activation and lower limb kinematics in habitual SH

runners. Additionally, Olin and Gutierrez (2013) examined muscle activation in BF and SH running conditions. All these studies examined recreational runners, but similar findings have been reported for competitive distance runners (Snow et al., 2015). While these studies examined TA activation primarily during stance, similar findings were found for activation during both pre-heel strike and post-heel strike periods. Both Von Tscharnner et al. (2003) and Standifird et al. (2013) reported that TA activation was lower during the pre-heel strike phase when running BF compared to SH. Post heel strike activation of TA did not differ in either study. While these five studies had consistent findings, still others show contradictory results.

Two studies have reported no effect of BF running on TA activation (Ervilha et al., 2017; Lindberg and Olsson, 2012). A key difference in Ervilha's work is that data collection occurred during overground running rather than running on a treadmill. As treadmill and overground running place different demands on lower limb musculature (Ervilha et al., 2017), it is somewhat inappropriate to directly compare these studies, which examined runners using different running surfaces. While the work of Lindberg and Olsson (2012) examined treadmill running, they failed to normalize EMG, making the results of this work difficult to interpret. In addition to those who have reported no effect of footwear on TA activation, De Silva Azevedo et al. (2016) indicate that TA activation was higher in BF running than in SH. These authors performed a training study that did not directly compare BF to SH activation but focused instead on how a BF training program would affect muscle activation. Examination of their results suggests TA activity was higher in the BF condition. As the authors did not explicitly examine

these values, the statistical significance of these differences cannot be determined.

Although there is still controversy in the literature on how shoe conditions affect TA activation during running, it appears most studies in the literature suggest that the TA is less activated during BF running compared to SH running.

Calf muscle activation

Research examining MG muscle activation during running in different running conditions (MIN, BF and SH) has delivered inconsistent outcomes. Some studies have determined there is an increase in MG activation in BF running compared to SH running (Ervilha et al., 2017; Fleming et al., 2015; Lucas-Cuevas et al., 2016; Olin & Gutierrez, 2013; Snow et al., 2015; Standifird et al., 2013). Some studies have reported a decrease in MG muscle activation in BF running compared to SH running (Da Silva et al., 2016; Lindberg & Olsson, 2012; Strauts et al., 2016). Still other research has reported no difference between conditions (Ahn, Brayton, Bhatia, & Martin, 2014). The next section provides more details on how these studies reported MG muscle activation.

BF vs. SH

Even though there is some disagreement in the literature about MG muscle activation when running BF compared to SH, most studies that have examined MG have found an increase in muscle activation when running BF compared to SH. Eight studies found this result, wherein six experiments examined treadmill running (Fleming et al., 2015; Lucas-Cuevas et al., 2016; Olin & Gutierrez., 2013; Snow et al., 2015; Standifird et al., 2013), and one looked at overground running (Ervilha et al., 2017). Lucas-Cuevas et al. (2016) examined the effect of three different running conditions (BF running, BF with

a rearfoot strike, and SH running) on muscle activation. In their study, LG and MG muscle activity were higher in the BF compared to the SH condition. Snow et al. (2016) also studied alterations in muscle activation during running in BF and SH in eight experienced MIN shoe runners who ran on a treadmill for 10 min for each running condition. The authors found that during the stance and swing phases, BF running led to greater muscle activation in MG.

Ervilha et al. (2017) looked at muscle activation patterns in nine habitual SH runners who were examined during the following running conditions: BF with forefoot strike pattern, SH with a forefoot strike pattern and SH with a rearfoot strike. They observed that MG EMG was higher for BF with the forefoot strike pattern and SH with the forefoot strike pattern compared to the SH with the rearfoot strike pattern condition. Similarly, Fleming et al. (2015) reported an increase in MG and LG muscle activation during the pre-activation phase in BF running compared to SH. Olin & Gutierrez (2013) examined eighteen SH recreational runners who ran for three distinct 7-minute experiments. They found that average EMG for MG was higher in BF running compared to SH running. Even though these studies recruited different types of runners such as recreational runners and long-distance runners, EMG results for MG have been found to be similar in all of them.

In a previous investigation comparing muscle activity in BF, SH and Vibram FiveFingers (VF®) running in three distinct time periods, the researchers reported no differences found in MG in the pre-activation phase in BF compared with other running conditions (Standifird et al., 2013). Moreover, another previous investigation reported no

significant differences between BF and SH. Ahn, Brayton, Bhatia, Martin (2014) examined forty subjects (twenty-one recreational runners and nineteen well-trained runners). Subjects ran on motorized treadmills at 2.5, 2.8, 3.2, and 3.5 m/s in BF (wearing five-toed socks) and SH (wearing traditional running shoes) conditions. The researchers did not find any significant differences in MG activation between the two running conditions. These results might be different due to the different types of participants with different characteristics.

Some studies have demonstrated that MG activation decreased during BF running compared to SH running. Strauts et al. (2016) looked at muscle activity in six recreational distance SH runners. Subjects ran on a treadmill for 5 min, then completed three ten-minute intervals of BF running, completing the last minute in SH. Results indicated that EMG was significantly lower for LG in BF compared to SH (pre-condition). Similar results have been illustrated by Lindberg & Olsson (2012), who recruited eight runners (three elite competitive runners and three recreational runners). The runners were required to run 10km/h and 14km/h in each condition. They determined that both LG and MG muscle activation decreased in the BF condition compared to the SH condition. Similar findings have been observed by Da Silva et al. (2016). This study investigated the influence of six weeks of progressive running training during BF running on impact force and muscle activity in habitual SH runners. When looking at EMG results, GL pre-activation decreased in BF pre-training compared to SH pre-training.

Although many studies used treadmill running as a running surface, they still demonstrated dissimilar findings. Contradictory outcomes may be attributed to variations

in individuals' characteristics, training status, and testing protocols. Despite the discrepancy in the literature, the majority of research has concluded that BF running results in an increase in MG activation.

Soleus

Like the MG, research examining SOL activation in BF and SH running has resulted in contradictory outcomes. In total, four studies have compared SOL activation during BF and SH running, with one reporting an increase in muscle activation for SOL (Ervilha et al., 2017), and two reporting no differences in muscle activation (Erik Lindberg & Olle Olsson, 2012; Standifird et al., 2013). The remainder reported a decrease in SOL muscle activation in BF compared to SH (Strauts et al., 2016). Ervilha et al. (2017) investigated nine habitual SH runners who ran straight for 20 m on a track at a self-selected pace. There was an increase in muscle activation of the SOL in BF running using a forefoot footfall pattern compared to SH running using a forefoot pattern. In this study, results for SOL activation in BF running were different compared to other results that showed decreased SOL activation. This might be due to overground running, which might require more SOL activation when BF compared to SH running.

There was an additional study carried out by Standifird et al. (2013). This produced differing results from previous studies. Ten recreational runners performed overground running using three different varieties of running shoes: Vibram FiveFingers (VF®), BF and SH shoes (Nike Air Pegasus). The study found no significant difference in SOL activation among shoe types. Furthermore, Lindberg & Olsson (2012) examined seven subjects who ran on treadmills in BF and SH; their results showed no significant differences for SOL activation in BF compared to SH (Lindberg & Olsson, 2012).

Strauts et al. (2016) studied EMG in six recreational runners when they ran BF and SH. EMG significantly decreased for SOL in BF running compared to SH running (SH pre condition). The literature contains some disagreement about how footwear affects SOL EMG or calf muscles, but it seems that most research found an increase in MG activation; however, SOL activation results vary during running in BF compared to SH running.

Gluteus Maximus

Even though the GMax has a seemingly small role during walking, it has been found that GMax activation increases noticeably and changes its timing during running (Lieberman, Raichlen, Pontzer, Bramble, & Cutright-Smith, 2006). Only one study by Snow et al. (2015) examined GMax differences between BF and SH running by incorporating MIN shoe users in the study. Interestingly, researchers found a significant increase in GMax muscle activation in BF running during stance compared to SH running. In addition, during the swing phase, they found no differences between the two conditions for the same muscle. Even though the GMax muscle plays a significant role during running, it is still unclear whether the muscle activation of GMax alters as an effect of changing footwear during running. This is largely due to a lack of relevant information in the literature.

Peroneus longus

One previous study by Standifird et al. (2013) compared peroneus longus (PL) in BF, SH, and Vibram FiveFingers (VF®) during overground running. Authors found no difference in the activation of PL among the three running conditions.

Biceps femoris

Those who have studied BiF activation during running in different footwear conditions have reported varied results. Some studies reported a decrease in muscle activation for the BiF during BF running compared to SH running (Strauts et al., 2016; Tam et al., 2017) while other authors have reported increased activation (Da Silva et al., 2016; Fleming, Walters, Grounds, Fife, & Finch, 2015).

Four studies were found which investigated muscle activation in BiF, wherein three of these studies used treadmills to collect data (Da Silva., 2016; Fleming et al., 2015; Strauts et al., 2016). One study completed the protocol for overground running (Tam et al., 2017). Strauts et al. (2016) recruit six habitual SH runners for the study sample. All subjects wore the same type of SH running footwear for the SH condition. The researchers compared the peak and mean EMG across all conditions and reported that the mean EMG levels across the running cycle significantly diminished for BiF muscle in BF running compared to the SH pre condition. Furthermore, Tam et al. (2017) examined twenty-two subjects (ten trained and twelve well-trained) during an overground running protocol. Their results show that, during stance, BiF activation reduced in BF compared to SH in the trained group.

Moreover, when Da Silva et al. (2016) examined six habitual SH runners (three women and three men), they compared muscle activation pre- and post-sixteen weeks of progressive BF running training. Results show that before training, EMG root mean square (RMS) in BF conditions had higher values than RMS EMG in SH conditions during the stance phase. Fleming et al. (2015) examined ten runners on treadmills;

researchers found similar results regarding the effect on BF running on muscle activation in six muscles. BiF activation in the pre-activation phase increased in BF running compared to SH running. There have been mixed results for BiF activation during running in BF and SH conditions, but most of the findings show that muscle activation for the muscle mentioned increases in BF conditions compared to SH running conditions.

Rectus Femoris

Several studies have investigated muscle activation of the RF muscle during running in different conditions with varying results (Da Silva, 2016; Fleming et al., 2015; Gavilanes-Miranda et al., 2012). Da Silva (2016) found that the RF RMS EMG was lower during the stance phase of BF running before training. Gavilanes-Miranda et al. (2012) found no significant differences in RF activation in BF jogging compared to SH jogging. Only one study, conducted by Fleming et al. (2015), reported that RF activation increased during BF running pre-activation compared to SH running.

Vastus lateralis

Research studies that reported muscle activation for vastus lateralis (VL) during running in different running conditions have provided two different overarching findings. While a small number of these studies has found that VL activation increased in BF running (Da Silva et al., 2016; Fleming et al., 2015; Snow et al., 2015), other studies reported a decrease in muscle activation for VL during running in BF compared to SH (Ervilha et al., 2017; Tam et al., 2017).

Two studies used overground running (Ervilha et al., 2017; Tam et al., 2017) while a few studies had their participants run on treadmills (Da Silva et al., 2016; Snow et

al., 2015; Fleming et al., 2015). Snow et al. (2015) examined eight male runners who ran on a motorized treadmill for 10 minutes under two running conditions (BF and SH). After participants warmed up in their selected speed and preferred foot strike pattern, participants performed 10 min running in each condition at 70 % of their maximum aerobic speed, and then sat for 3 min as a rest period between the two conditions. The results show that VL activation increased during the swing phase in BF running compared to SH running. Similarly, Fleming et al. (2015) examined a group of habitual SH runners. Participants completed 1-minute bouts of running trials at three velocities in both BF and SH conditions. Their results indicate that VL activation increased in BF compared to SH in the three velocities in all phases (pre-activation, absorptive phase, and propulsive phase). Moreover, Da Silva et al. (2016) examined six experienced distance runners who were habitual SH runners and compared pre/post- training for both the SH and the BF conditions. In their results, VL activation (RMS) in BF running was higher compared to SH running activation (RMS).

Ervilha et al. (2017) studied lower limb muscle activation during BF and SH running conditions in nine habitual SH runners, who ran straight for 20 minutes at a self-selected speed on a treadmill. The researchers compared EMG data across the following running conditions: BF using forefoot footfall pattern, SH using forefoot pattern and SH with the rearfoot pattern. Examination of the data provided in the paper indicates that VL EMG was lowest during BF running using forefoot footfall conditions compared to the other two conditions. Similarly, in work by Tam et al. (2017), twenty-two participants partook in a 10-km fatiguing trial. Although again the authors did not specifically

compare muscle activation in BF vs. SH, data presented in the paper suggest that VL activation in BF running trials was lower than VL activation in SH running in both trained and well-trained participants. Based on the review of VL activation research, two findings are evident: some studies found an increase in muscle activation, whereas other studies found a decrease in VL. This contradiction is possibly due to the fact that these studies used different running protocols and different running surfaces. This makes it ultimately hard to draw a conclusion on whether muscle activation for VL increases or decreases during running in either BF or SH running conditions.

Vastus Medialis

Research that reported muscle activation for vastus medialis (VM) during running in different footwear conditions exhibited different findings as well. Two research studies showed the results for VM. One study used treadmills in their protocol (Strauts et al., 2016), while the other study collected the data from jogging overground (Gavilanes-Miranda et al., 2012). Strauts et al. (2016) found that muscle activation for VM decreased in BF running compared to SH. Gavilanes-Miranda et al. (2012) determined similar results when they compared EMG of BF running and SH running in six low-extremity muscles, wherein they showed VM activation decreased in BF running compared to SH running for the right leg. They also found that muscle activation for VM did not differ between the three running conditions for the left leg.

MIN vs. SH running

While there is a considerable amount of research studies that has examined changes in muscle activation for BF vs. SH running, relatively little literature exists

comparing MIN to the SH condition. One research study by Kasmer, Ketchum & Liu (2014) examined four experienced runners (one runner ran in MIN shoes and three runners in SH shoes). Their findings revealed that median frequency did not change in the pre-run trials when compared to post-run trials in the TA in both footwear conditions. The RF median frequency was higher in the post-run condition when running in MIN shoes. The findings also revealed that there were significant larger RMS values in the post-run trials compared to RMS values in the pre-run trials in the hip abductors in the MIN shoe type during the first loading response. In both pre- and post-run conditions, there was a higher RMS value in the SH shoe type compared to the MIN shoe type in the TA muscle. Other than the Kasmer study, the author was unable to find any other research that has examined and compared muscle activation in MIN and SH.

2.1.1.5 Summary of footwear and muscle activation

Given the remarkable increase in the number of runners committing to running and the lack of relevant literature, there is a clear need to develop research on the effect of MIN footwear on muscle activation (e.g., Refs, Bonacci et al., 2013; Cheung et al., 2014; Rothschild, 2012). There is a wide divergence of findings which, when combined with an overall lack of research comparing muscle activation when running wearing SH .vs MIN in the field, leads to difficulties when trying to find a link between footwear type and muscle activation. In particular, there is still a dearth of literature regarding muscle activation when running wearing MIN (Standifird et al., 2013).

2.1.3.2. Running speed

Another variable known to affect muscle activation during running is running speed; running speed alters ankle and knee kinematics (Fredericks, Swank, Teisberg, Hampton, Ridpath, & Hanna, 2015). It is highly likely that muscle contributions to propulsion probably change as well, depending on velocity (Hamner, Seth & Delp, 2010). According to integrated EMG (iEMG) measurements, muscle activation increases in tandem with running speed (Ross, Leveritt, & Riek, 2001).

EMG of the MG and SOL muscles has been found to be higher in running than in maximal isometric trials (Mero, Kuitunen, & Komi, 2001). The pre-activity increased EMG of VL and MG more when the speed increased (Kyröläinen, Avela & Komi, 2005). For the gastrocnemius, faster running speed resulted in higher activation just prior to heel contact. (Komi et al., 1987). Kyröläinen et al. (2005) found that RF EMG increased noticeably with increasing running speed. This increase occurred in two events: one before and at the start of the contact phase for the knee extensor muscle, and the other at the early stage of the swing phase when starting the hip flexion.

Moreover, Lindberg and Olsson (2012) did a comparison of different speeds under two different running conditions. They discovered no significant changes for TA muscle activation when they observed speeds of around 12 to 14 km/h in either running condition. A few of the participants showed a slight decrease in EMG when they ran at 14 km/h, whereas some other participants showed rising levels of EMG during exertions of the same muscle at 14 km/h. Additionally, there was a significant increase in SOL

muscle activity at 14km/h when compared to muscle activity at 10 km/h for SH running. Lindberg et al. (2012) also looked at muscle activity in the BF running condition with respect to speed. The authors found that there was an increase in muscle activity at the higher speed for BF running, but results were not significant. Results also show that LG muscle activity increased when running speed increased; however, MG muscle activity results exhibited no significant difference in running speed for either condition. It seems that as running speed increases, there is a much stronger force production requiring significant increases in EMG of two-joint muscles throughout the subsequent running cycle (Kyröläinen et al., 2005).

2.1.3.3 Different running surfaces (Treadmill vs. Overground Running)

Over the past several decades, a broad range of kinematic and kinetic studies have compared features of overground and treadmill running (Waldhelm, Fisher, 2016; Schache et al., 2001). In general, results indicate running kinematics during overground and treadmill running are similar with only slight differences (Waldhelm et al., 2016). These differences include a reduction in stride length, increase in stride rate, and a decrease in the period of non-support when running overground as compared to treadmill running (Monte, 1976). Moreover, some kinetics were also found to be different for the treadmill and overground running. These include joint moments, peak values of ground reaction force and joint power trajectories (Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007).

In earlier research efforts, there were only minor differences reported when comparing overground to treadmill surfaces in relation to spatiotemporal variables, kinetics and kinematics for running and walking (Cronin & Finni, 2013; Riley et al., 2007; Strathy, Chao & Laughman, 1983). At the same time, there is an immense amount of data supporting the equivalency of overground and treadmill surfaces in terms of creating similar gait patterns (Cronin, & Finni, 2013; Matsas, Taylor & McBurney, 2000). Even so, just because the locomotive effort required while on overground and treadmill running surfaces gives more or less the same kinematics, this does not imply that joint movements or muscle activation patterns are the same for both modalities (Lee & Hidler, 2008). Baur, Hirschmüller, Müller, Gollhofer, and Mayer (2007) provided evidence for detailed differences between overground and treadmill running regarding mainly earlier and longer EMG of PL, along with decreased amplitude in the SOL during treadmill running (Baur et al., 2007).

Based on their research results, Waldhelm et al. (2016) proposed that the treadmill's moving belt could lead to potential adaptations in the neuromuscular system as well as muscle activity differences. Treadmill runners consistently exhibit decreased vertical oscillations in their center of mass, in addition to reductions in forefoot loading; these findings indicate that the triceps surae, either all or in part, might be less active in treadmill running (Waldhelm et al., 2016). The study's results were supported by another previous investigation (Baur et al., 2007). These authors reported reductions in SOL muscle activation at the push-off phase in treadmill running. In the same research study, the findings suggested that there was a higher activity of the SOL and peroneus longus

muscles in the weight-acceptance phase. This may indicate that treadmill runners require greater stability compared to overground runners (Baur et al., 2007; Waldhelm et al., 2016). Differences in these findings could result from differences inherent in the mechanical properties of the treadmills used for the research testing or a general lack of participant's familiarity with treadmill running (Schache et al., 2001; Waldhelm, Fisher, 2016). While it appears that the biomechanics in lower limbs is in fact affected by differences in running surfaces, results pertaining to why this occurs are still inconclusive. What has been conclusively determined is that running on different surfaces affects muscle activation.

2.1.3.4 Effects of fatigue on muscle activation

The concept of fatigue has been intensely studied, but no single, universally accepted definition of the term has emerged, due to a wide range of paradigms applied in its study (Enoka & Stuart, 1992). Commonly, the term fatigue is used as an expression to describe a decline in physical performance related to an increase in the challenge of performing a task (MacIntosh, Gardiner & McComas, 2006), so-called peripheral fatigue. . There are many potential mechanisms for fatigue. A novel interpretation of fatigue has emerged in literature. It proposes that the integration of peripheral afferents from active skeletal muscles alters the regulation of motor control by the central nervous system (Noakes & Gibson, 2004). This type of fatigue is referred to as central fatigue. Both types of fatigue (central and peripheral) are known to take place during whole-body exercise (e.g., running, cycling, rowing, skiing, hiking/trekking). Given that fatigue has an effect on muscle activation (Mizrahi, Voloshin, Russek, Verbitski, & Isakov, 1997), it is

not surprising that whole-body exercise can affect muscle activation. Before reviewing the literature related to fatigue and muscle performance, I will provide definitions of these fatigue categories.

Two types of fatigue are generally recognized in the literature. The term central fatigue refers to sensations /information coming from the periphery (i.e. from cardiac, skeletal and pulmonary tissues) that modulates central nervous system responses (such as motor cortex drive and limbic system). Based on this feedback, the central nervous system regulates the recruitment and derecruitment of motor units in active muscles during exercise. Therefore, the process of skeletal muscle recruitment during a state of fatigue is controlled by the central nervous system. Central fatigue becomes a dominant factor for exercise and ensures that no failure of cell homeostasis takes place during voluntary exercise, the so-called the central governor (Noakes & Gibson, 2004; Noakes, Gibson & Lambert, 2005). Peripheral fatigue, on the other hand, refers to a very objective criterion that diminishes physical performance, which in turn, is linked to muscle fiber contractile alteration. In peripheral fatigue muscle action potentials are known to be affected by build-up of metabolites such as potassium, inorganic phosphate, sodium, calcium and hydrogen that lead to muscle contractile apparatus failure(Allen, Lamb & Westerblad, 2008; Noakes & Gibson, 2004). From this point onwards in this thesis, central and peripheral fatigue will refer strictly to the above mentioned definitions.

During bouts of exercise, both peripheral and central fatigue can contribute to performance deficits. The degree to which both types of fatigue contribute to

performance deficit is mainly reliant on the nature of the exercise. Generally speaking, peripheral fatigue is the dominant contributor behind early deficits in a bout of exercise. Peripheral fatigue is the dominant contributor to exercise deficits that occur early in the exercise bout, provided the muscle mass being exercised is relatively small (that is when the cardiorespiratory system has little contribution to the fatigue). In contrast central fatigue plays a more dominant role as the length of the exercise period increases and muscle mass involvement increases (Thomas et al., 2015; Rossman, Venturelli, McDaniel, Amann & Richardson, 2012). The role of central fatigue is evident in runners who run longer distances or in those who run high intensity interval training as they will most often experience a decrease in performance due to its presence (Vuorimaa, Virlander, Kurkilahti, Vasankari, & Häkkinen, 2006). Given the changes in voluntary activation that are known to occur with fatigue, it is logical to conclude that running related fatigue affects muscle activation. Research in this topic will now be reviewed.

The literature offers overwhelming evidence that fatigue affects running biomechanics, and these effects can occur after only fifteen minutes of exertion (e.g., running or fast trot) (Brown, Zifchock, & Hillstrom, 2014; Derrick, Dereu & Mclean, 2002). Many investigations have reported a decline in the amplitude of EMG signals (iEMG values and root mean square) with fatigue-inducing running (Billaut & Smith, 2010; Jewell, Boyer & Hamill, 2017; Jewell, Hamill, von Tscharnner & Boyer, 2019; Lepers, Pousson, Maffiuletti Martin & Van Hoecke, 2000; Millet et al., 2002; Mendez-Villanueva, Hamer, & Bishop, 2007, 2008; Mizrahi, Verbitsk & Isakov, 2000; Newton, 2010; Nicol, Komi & Marconnet, 1991; Paavolainen, Nummela, Rusko & Häkkinen,

1999; Racinais, Bishop, Denis, Lattier, Mendez-Villaneuva, & Perrey, 2007; Vuorimaa et al., 2006; Weist, Eils & Rosenbaum, 2004). Muscle activation can also be affected by fatigue that occurs during exercise such as intermittent sprinting (Bishop, 2012). Ultimately, fatigue can lead to modifications in running kinematics and kinetics in long-distance running because of the repetitive activity (Winter, Gordon & Watt, 2017).

Intense exercise-induced muscle fatigue is measurable according to alterations found in EMG signals. For instance, Weist, Eils & Rosenbaum (2004) discovered decreased levels of EMG in MG and TA, during intensive treadmill training. In general, changes in kinetics, kinematics, and EMG are well-known fatigue-induced phenomena that occur in injured and healthy runners alike (Brown Zifchock & Hillstrom, 2014). During physical exertions like running, these kinds of changes can lead to running-related injuries and reduced performance levels. Intense exertion leads to a failure to produce the necessary power output that is required for a given activity (Hashish, Samarawickrame, Baker, & Salem, 2016; Phinyomark, Thongpanja, Hu, Phukpattaranont & Limsakul, 2012).

2.1.3.5 Effects of fatigue on muscle activation during running

Alterations in EMG signals during running have been studied by numerous researchers to evaluate muscle fatigue (Al-Mulla, Sepulveda & Colley, 2011; Enoka & Duchateau, 2008; Hanon, Thepaut-Mathieu & Vandewalle, 2005; Tam et al., 2017; Weist et al., 2004). A broad range of studies has assessed running performance by examining the changes that take place in neuromuscular characteristics at fatigue onset (Paavolainen,

Nummela, Rusko, & Häkkinen, 1999). Nicol et al. (1991) tested nine experienced endurance (marathon) runners for neuromuscular performance. The study found that a reduction in VL iEMG during maximal contractions after a marathon run led to decreases in maximal force production in both isometric and dynamic situations (Nicol et al., 1991).

Other studies reported different results regarding the influence of fatigue on VL muscle activation during running (Abe, Muraki, Yanagawa, Fukuoka, & Niihata, 2007; Camic, Kovacs, Enquist, VanDusseldorp, Hill, Calantoni, & Yemm, 2014; Hanon, Thepaut-Mathieu, Hausswirth, & Chevalier, 1998; Paavolainen et al., 1999; Weist et al., 2004). Most of these studies were conducted on treadmills, but some were performed using overground running (Paavolainen et al., 1999; Tam et al., 2017). Two studies found a decrease in VL muscle activation following fatigue (Paavolainen et al., 1999; Tam et al., 2017), while other studies found an increase in VL EMG after fatigue (Abe et al., 2007; Camic et al., 2014; Hanon et al., 2005; Hanon et al., 1998). Another study by Weist et al. (2004) found that fatigue exercise did not affect VL muscle during running.

Abe et al. (2007) examined VL muscle activation in seven male novice distance runners during 90-minute prolonged running tests. The iEMG during the tests was separated into eccentric and concentric phases. The study found a significant decrease in the average iEMG value of the VL in the concentric phase at the 90th minute compared to the 10th minute during the test, but there was no significant change in the average iEMG value of the eccentric phase during the test. Camic et al. (2014) examined the VL muscle in fifteen recreational runners while they were performing an incremental run on

a treadmill. The experiment started at a speed of 4.8 km h^{-1} and then it was increased by 1.6 km h^{-1} every 2 minutes. The runners continued until they reached approximately 90% of their age-predicted maximal heart rate (HR_{peak}). During the second visit, each runner completed an incremental test to the point of exhaustion. All participants exhibited a linear increase in EMG amplitude of VL, associated with the time at the fatiguing test velocities. Hanon et al. (2005) studied iEMG of VL during discontinuous four-minute stages of incremental run participants reach the level of exhaustion. The authors found a significant increase in the EMG amplitude for five successive running bursts of activity at the end compared to the start. They found that the increases in EMG amplitude, which took place during constant-velocity running on the treadmill, replicated fatigue-induced rises in muscle activation required to sustain a specific pace (Hanon et al., 2005).

A study by Weist et al. (2004) looked at the effect of fatigue on VL muscle activation. Thirty runners partook in the study and completed a maximally exhaustive run above their anaerobic thresholds. The authors found no differences in VL EMG. Paavolainen et al. (1999) studied VL EMG before and after fatiguing long-distance running (10 km) on a 200-m indoor track. Although some decreases in average iEMG (averaged across three muscles) were reported following fatigue, no individual results for VL were provided (Paavolainen et al., 1999). Although research produced different findings on how fatigue affects muscle activation during running, most studies have revealed that VL activation increases when running following fatiguing.

Relatively few studies have investigated the effect of fatigue on VM muscle activation during running. Two studies examined VM EMG during treadmill running, but the results from these studies were diverse (Kellis et al., 2009; Weist et al., 2004). Whereas Kellis et al. (2009) found an increase in VM EMG amplitude, a previous investigation reported no difference in VM EMG amplitude after fatigue (Weist et al., 2004). Specifically, Kellis et al. (2009) examined fifteen female runners who ran at 3.61 m/s on a treadmill. The participants ran immediately before and after the fatigue protocol, which contained an isokinetic knee extension/flexion protocol and, on another visit, an ankle plantarflexion/dorsiflexion fatigue procedure. The study found that, after fatigue, VM RMS increased. Furthermore, Weist et al. (2004) looked at the EMG of VM in thirty experienced runners. The participants performed an exhaustive run after an eight-minute warm-up at their comfortable pace. EMG signals for at least 20 steps were collected every two minutes during the exhausting run. The authors reported no significant difference in VM activation after the fatiguing run.

Weist et al (2004) studied the influence of fatigue on GMax EMG while participants performed an exhaustive run on a treadmill. The researchers found that there were no significant differences in GMax muscle activation after fatigue (Weist et al., 2004). Due to a dearth of investigations of GMax EMG after fatigue, firm conclusions cannot be drawn about how fatigue would affect these muscles during or after running.

One research study, carried out by Tam et al. (2017), looked at the effect of fatigue on gluteus medius. Twelve well-trained and ten trained runners were recruited for

the study, with participants performing a 10-km fatiguing trial. EMG signals were measured during overground running in two running conditions, BF and SH. Irrespective of the footwear participants wore, the trained group exhibited a diminished median frequency of gluteus medius after fatigue compared to the well-trained group.

Additionally, three studies looked at the EMG of RF after fatigue (Hanon et al., 2005; Mizrahi et al., 2001; Weist et al., 2004). Although data were collected from treadmill running trials, the findings from these studies reported different outcomes regarding the effect of fatigue on RF EMG. Two studies found an increase in EMG after fatigue (Hanon et al. 2005; Mizrahi et al., 2001), while Weist et al. (2004) found no difference in RF EMG amplitude after a fatiguing run. Hanon et al. (2005) examined the EMG of RF during a fatiguing protocol on a treadmill. Nine well-trained runners performed a running protocol that consisted of four-minute stages, during which running speed was increased incrementally until the participants reached exhaustion. The EMG measurements were recorded throughout ten bursts of activation. The results show that EMG amplitude significantly increased throughout the running trials. However, RF exhibited signs of fatigue earlier than the other muscles (Hanon et al., 2005).

Similar results were found by Mizrahi et al. (2001), who looked at EMG in fourteen recreational runners. Participants ran above their level-running anaerobic threshold speed for 30 minutes on a treadmill. The RF iEMG significantly increased during running, starting at the 20-minute mark, with an increase in median power frequency at the 15-minute mark (Mizrahi et al., 2001). However, when Weist et al.

(2004) examined the EMG amplitude of the RF, they noted no changes in muscle activation, nor did they find that fatigue affected its amplitude. Another study examined quadriceps EMG during fatiguing running (Mizrahi et al., 1997). Twenty-two runners participated in the study, performing a treadmill run for 30 minutes at a speed corresponding to each participant's anaerobic threshold level. The researchers found no significant variations in time during the running protocol (Mizrahi et al., 1997). There appears to be a discrepancy in the literature regarding the effect of fatigue on RF EMG, but this might be due to the recruiting of participants, protocol setting, or training status.

Two previous investigations studied the effect of fatigue on PL activation during treadmill running (Cheung et al., 2010; Weist et al., 2004). Cheung et al. (2010) tested 20 female recreational runners who ran 10 km. Participants wore two different types of footwear over two visits. The researchers found that PL EMG increased with the increase of the running distance. Moreover, the median frequency declined for both running conditions and shoe conditions with mileage, which demonstrates the presence of fatigue. However, another study was carried out by Weist et al. (2004), who had 30 experienced runners run above their anaerobic threshold on a treadmill. The study found a significant decrease in PL EMG amplitude with fatigue running compared to non-fatigue running. Because not many running investigations have examined the effect of fatigue on PL, it is still unclear how PL EMG alters muscle response during fatigue running.

Research examining TA muscle activation during fatigue running has reported varying outcomes (Cheung et al., 2010; Hanon et al., 2005; Tam et al., 2017; Weist et al.,

2004). Three studies reported an increase in TA activity after fatigue (Cheung et al., 2010; Hanon et al., 2005; Tam et al., 2017), whereas one study found no significant difference in TA EMG. Three out of four of these studies collected data from treadmill running (Cheung et al., 2010; Hanon et al., 2005; Weist et al., 2004), while one study used overground running (Tam et al., 2017). Hanon et al. (2005) examined nine trained runners. The participants ran in four-minute stages, with speed gradually increasing until participants could not keep running. The EMG signals were assessed during ten bursts of activation. The results show that TA EMG significantly increased at the final stage of fatigue running.

Similar results were found by Cheung et al. (2010), who examined 20 recreational runners. Their findings show that the EMG of TA was higher as running mileage increased. Additionally, Tam et al. (2017) looked at TA activity in twelve well-trained and ten trained runners who performed a 10-km fatiguing trial. Their results demonstrated that TA EMG increased in both groups over time after a fatiguing run. These findings contrast with those of Weist et al. (2004), who found no difference in TA activity during a fatiguing run. However, the differences in results might be due to EMG being collected and analyzed during the stance when TA is not highly active during running (Weist et al., 2004). Overall, it seems clear that TA activation amplitude tends to increase during fatigue running since most research has revealed this outcome.

Like research on TA, research on the effect of fatigue on BiF has also produced varying results. Three studies looked at BiF during a fatiguing run. Two reported a

decrease in BiF EMG after the onset of fatigue (Paavolainen et al., 1999; Weist et al., 2004) while the third found that BiF activation increased after fatigue onset (Hanon et al., 2005). Weist et al. (2004) examined 30 experienced runners. The participants ran above their anaerobic threshold on a treadmill. The results show that the BiF exhibited a significant reduction in EMG amplitude. Paavolainen et al. (1999) did examine the BiF EMG during pre/post-fatigue running. As discussed above, these authors reported only average iEMG across 3 muscles, so that no conclusions can be drawn about BiF activation. In contrast, Hanon et al. (2005) found an increase in BiF activity in nine well-trained participants when they performed a running test involving four-minute stages with increasing speed until participants felt exhausted. Since there is little research on the effect of fatigue on BiF activation, it is challenging to give a clear conclusion regarding how fatigue can alter BiF EMG.

The SOL EMG has only been investigated by a few researchers (Girard, Millet, Micallef and Racinais, 2012; Hashish, 2016; Weist, 2004). The authors reported a decrease in SOL EMG during fatigue running. Two studies collected data from overground running (Girard et al., 2012; Hashish, 2016), while one study collected data from treadmill running (Weist et al., 2004). Thirty experienced runners were recruited by Weist et al. (2004), with participants running on the treadmill at speeds above each runner's anaerobic threshold. Then, the speed was increased every two minutes by two km/h until runners became exhausted. This was followed by a six-minute run as a cool-down. The researchers found that SOL EMG was significantly diminished during the fatiguing runs (Weist et al., 2004). Similar results were found by Girard et al. (2012),

who examined SOL EMG in eleven well-trained athletes. The participants performed a 5-km running time trial on an indoor Tartan Track®. The results showed that, after exercise, SOL EMG (RMS) was significantly reduced. Meanwhile, Hashish et al. (2016) looked at the median frequency in SOL EMG, with 21 recreational runners participating in the study. The participants performed novice barefoot running pre- and post-exertion on an outdoor concrete running surface. The EMG signals were collected during pre- and post-exertion trials. The results in this study show that SOL median frequency is significantly reduced due to fatigue and that after fatigue, SOL EMG declines when runners experience fatigue during running.

Research examining MG EMG during fatigue running has also reported varying results. Some studies determined there was no difference in MG EMG between pre- and post-fatigue running (Hanon et al., 2005; Hashish et al., 2016; Mizrahi et al., 1997), whereas other studies reported a decrease in MG activation after fatigue (Paavolainen et al., 1999; Weist et al., 2004) as well as a decrease in median frequency in a well-trained group (Tam et al., 2017). A study by Kellis et al. (2009) reported an increase in MG EMG, and a study by Tam et al. (2017) found an increase in median frequency in the trained group. Hanon et al. (2005) examined nine well-trained runners. The participants performed running tests comprised of four-minute stages, with increases in speed until exhaustion. EMG signals were measured in ten bursts of activity. Among other muscles tested in this investigation, the results demonstrate that MG was the only muscle that did not show an increase in EMG between the beginning and final stages (Hanon et al., 2005). Similarly, Hashish et al. (2016) examined the MG median frequency during

exhausting overground running in 21 recreational runners. The authors found that fatigue had no impact on MG median frequency. Mizrahi et al. (1997) examined 22 runners, who performed 30-minute running tests on a treadmill at speeds corresponding to their individual anaerobic threshold values. Results of the normalized EMG indicate that no differences were found in either frequency or time domain for MG.

Weist et al. (2004) studied MG EMG in 30 runners who completed an exhaustive running test on a treadmill. The researchers discovered that MG EMG amplitude significantly decreased after the fatiguing run. Paavolainen et al. (1999) also investigated MG EMG following a fatiguing long-distance run. Again, no conclusions can be drawn from this study due to the nature of the analysis performed. Kellis et al. (2009) studied MG EMG in fifteen female runners. EMG signals were collected during a treadmill running at a speed of 3.61 m/s pre-and post- a fatiguing program. The fatiguing program consisted of an isokinetic knee extension/flexion on one day and an ankle plantar flexion/dorsiflexion on a different day. The researchers found that the two different fatigue programs resulted in an increase in the EMG amplitude of MG.

Tam et al. (2017) looked at the median frequency of MG after a fatiguing protocol. Their results show that median frequency increased in the trained group (Tam et al., 2017). Although different studies have reported contrasting outcomes regarding the effect of fatigue on MG EMG, it seems that, overall, this activity is not affected by fatigue. Because EMG of the lower limb muscles likely varies according to the type of running or running protocol being performed (e.g., sprinting, short-distance running,

marathon running, long-distance running, etc.), different studies have applied varying types of fatiguing protocols in investigations of running-related fatigue. The varied protocols used in the literature make it challenging to reach a definite verdict about the extent of muscle activation pattern responses to fatigue across running types.

2.1.4 Does footwear alter the effect of fatigue on muscle activation during running?

Relatively few researchers have examined running footwear as a factor that may alter the effect of fatigue on muscle activation. During running, runners might adapt to mechanical stress and fatigue by altering running biomechanics (Giandolini et al., 2016), although increasing muscle fatigue may then increase injury risk due to some abnormal changes in joint motion (Tam et al., 2017). However, it is not yet understood if the footwear has a major effect on fatigue during running.

One recent study by Cheung & Ng (2010) examined twenty female recreational runners who had excessive rearfoot pronation. Participants were tested when they ran 10 km on a treadmill in two visits. Participants wore Adidas Supernova (Adidas AG, Herzogenaurach, Germany) motion control running footwear for one visit and neutral running footwear (Adidas Supernova cushion) for the other visit. EMG data of TA and PL were collected. The total running distance was divided into ten equivalent stages, and EMG was recorded in every single stage in order to compare the EMG recordings. EMG RMS and median frequency were then compared for the two conditions. Fatigue was determined in the differences between the first checkpoint and the last checkpoint. The

authors reported that TA mean values across all EMG signals in the neutral condition were greater than mean values in the motion control footwear condition.

PL EMG was found to be higher in the neutral footwear than in the motion control footwear condition. When compared at early running stages and late running stages, results showed that PL EMG increased with the running mileage regardless of the type of footwear. This study also found a significant relationship between TA RMS and running mileage for the motion control shoes and in both muscles examined for the neutral footwear conditions, while for PL, only the neutral footwear condition exposed a correlation between RMS and running mileage. In both running conditions, the correlations were more obvious in the neutral footwear than in the motion control footwear condition. Median frequency significantly decreased when comparing the first and the last checkpoint in the 10 km running bout. Comparing the running footwear conditions, there was a significantly higher shift in median frequency in the PL muscle during running in the neutral footwear. However, for the TA muscle, there was a significant difference between footwear conditions, as there was a decrease in median frequency noticed in the neutral footwear conditions. The study concluded that running in motion control footwear tends to display more steady activation patterns and higher resistance to fatigue for both TA and PL in runners who have excessive rearfoot pronation. These findings suggest that motion control footwear might benefit runners' performance by increasing the endurance and delaying the occurrence of fatigue in leg muscles (Cheung & Ng, 2010). Though the focus of this study was not to study the effect of SH and BF running on muscle activation during a fatiguing run, it appears that there is

some evidence that footwear might have an effect on muscle activation and fatigue during running. Hence, this study offers some support for the questions being asked in this thesis.

A more recent study by Tam et al. (2017) compared EMG in lower limb muscles during acute fatigue in two different groups: trained and well-trained runners. Both groups ran a 10km fatiguing trial overground in both BF and SH conditions. Pre/post-fatigue trials were collected to assess the effect of fatigue on muscle activation and a number of other variables. These trials were collected on a treadmill. The study found that BiF muscle activity during the pre-activation portion of swing decreased in post-fatigue trials for BF running in both trained and well-trained groups; however, BiF muscle activity did not change during SH running. Both VL and BiF muscle activity declined in post-fatigue trials during the stance phase in trained and well-trained groups and for both footwear conditions. The study also found an increase in TA muscle activity in post-fatigue trials in both footwear conditions and for both trained and well-trained groups.

MG median frequency declined in the trained group in post-fatigue trials, but no differences were found between BF and SH conditions. Additionally, VL median frequency was also reduced in BF running during post-fatigue in the well-trained group, but no differences were found in SH conditions or in the trained group. LG median frequency increased in post-fatigue trials in the trained group in both BF and SH running conditions. However, it decreased in the well-trained group in both BF and SH running

conditions. These findings could result from fatigue affecting muscle activation during running (Tam et al. 2017). This study presents some effects of wearing different footwear (BF and SH) on fatigue during running. However, there is still a need to understand how fatigue and footwear interact when other types of footwear, such as MIN, are worn.

In reviewing all the factors that appear to have a direct influence on muscle activation, running fatigue is clearly one of the most important factors. However, there is a gap in the literature regarding the effects of different types of footwear on EMG during fatiguing tasks. Therefore, the examination of exercise-induced fatigue protocols should be of interest in this field to further understand the mechanisms underlying the relationship between peripheral fatigue and footwear. Using exercise-induced fatigue protocols could also lead to a better understanding of lower limb muscle activation patterns during running.

Based on this review of the literature, the following research questions were examined in this thesis:

1. Does the onset of peripheral fatigue in runners alter lower limb muscle activation amplitude? Is this effect altered by the type of shoe (MIN vs. SH) worn when running?
2. Does the type of footwear (MIN vs. SH) affect the median frequency of lower limb muscles when a fatiguing running trial is performed?

The proposed hypotheses are:

1. The onset of peripheral fatigue will have an effect on the lower limb muscle activation, and there will be a decrease in tibialis anterior (TA) muscle activation amplitude and an increase in medial gastrocnemius (MG) activation amplitude in MIN conditions.
2. There will be less peripheral fatigue exhibited in the lower limb muscles when running in MIN conditions. Although changes in median frequency will occur in the MIN condition, we expect the reductions to be fewer than those observed in the SH condition.

Chapter 3: Methods: Materials and Methods

3.1 Participants

Ten male long-distance runners volunteered to take part in this study (See Table 1 for participant demographics). All participants were healthy and free of acute or chronic musculoskeletal injuries. Informed consent was obtained in writing from all participants prior to starting the study. The participants' training regimes were strictly screened in advance to guarantee that their training status was sufficient to ensure they could complete the study. In general, the participants were required to train at least five out of seven days /week at an intensity exceeding 70% of their maximal aerobic speed on at least one of those training days. As well, the participants had to report running for a minimum of 50k over the course of each seven consecutive training sessions and adhere to a weekly training program. They were also categorized into "regional classes" in accordance with the online calculator for the USA Masters Track and Field (http://www.usatfmasters.org/fa_agegrading.htm). In the given categorization, regional class runners achieved 70-79.9% of the sex- and age-normalized world record required to qualify for a 10k race. Participants in this study scored an average of 74% (Blair, 2018).

Table 1. A summary of the participants' characteristics and training profile.

Participants' measure	Mean / SD
Age	28.3 +/- 8.4 (years)
Weight	71.1 +/- 4.9 (kg)
Height	176.4 +/- 6.5 (cm)
VO _{2max}	61.6+/-7.3 (ml min ⁻¹ kg ⁻¹)
HR _{max}	190.3+/-9.0 (b. min ⁻¹)
Structured training	4.4+/-4.7 (yr)
Maximal Aerobic Speed	1 8.0+/-1.1(km hr ⁻¹)
Weekly training sessions	7.1+/-2.7
Weekly interval sessions	1.5 +/- 0.7
Training load	1 04.0+/-63.5 km/week)
10K personal best	36:02 +/- 4:22 (mm: ss)

3.2 Laboratory visits

The study was completed over three laboratory visits. During the first visit, a volume oxygen uptake (VO_{2max}) test was performed to determine the endurance capacity and maximal aerobic speed. The maximal aerobic speed is essential for interval training and subsequent experiment sessions. The VO_{2max} test was completed in the lab on a motor-driven treadmill at a constant 1% slope. Participants performed a five-minute warm-up at their self-selected speed prior to the test. The incremental test started at an initial speed of $7 \text{ km}\cdot\text{h}^{-1}$, after which the speed was gradually increased by $1 \text{ km}\cdot\text{h}^{-1}$ every 2 minutes until subjects reached volitional exhaustion (Basset and Boulay, 2003; Uger and Boucher, 1980). A five-minute rest was given to the participants prior to going through a verification phase. The verification phase comprised running at 105% of the speed that was reached at VO_{2max} up until volitional exhaustion; it was essential to confirm that every participant reached his VO_{2max} (Rossiter, Kowalchuk, and Whipp, 2006). A recovery period followed the test until the participants' heart rates declined to $120 \text{ b}\cdot\text{min}^{-1}$. The maximal aerobic speed corresponded to the speed reached at VO_{2max} as per Basset and Boulay (2003).

On day two and three, participants completed a fatiguing running trial wearing either a traditional running shoe or a shoe classified as a minimalist shoe (described below). The order of shoe conditions was randomly assigned for the two visits. Thus, the two visits had an identical method design except for the type of running shoe, which was switched between visits. The fatiguing trials that occurred on days two and three were completed on a running track (described below).

3.3 EMG measurements

Prior to the running trials, participants were prepared for the collection of electromyographic (EMG) data. All EMG signals were collected by using a Delsys Trigno wireless acquisition system (Delsys Incorporated, Natick, Massachusetts). The EMG was sampled at 2000 Hz; CMR of 80db; a bandpass filter (20 Hz -450 Hz cut-off) was used to remove any movement artifacts.

The wireless electrodes were affixed to five muscles on each participant's right leg: BiF, MG, GMax, TA, and VL, as described by Criswell (2011). For each targeted muscle, the skin was prepared by shaving the hair over the belly of the muscle with a razor. Before the electrodes were placed on the skin, the skin was gently rubbed with medical sandpaper and then cleaned with an alcohol swab. After the skin was dry and free of hair, the electrodes were applied and secured on the skin with medical tape. The participants then performed maximum voluntary isometric contractions (MVCs) of each muscle. Each maximum contraction lasted for about five seconds, and the contractions were performed twice for each muscle. One-minute rest was provided for the participants between each contraction. Participants were verbally motivated while performing the contractions, to encourage them to exert maximum effort. Following the MVC trials, the participants warmed up for five minutes at a self-selected speed, on a motor-driven treadmill with a constant 1% slope.

3.4 Shoes

Each participant ran in a different type of running shoe for each visit. The SH shoes were Brooks "Glycerine 13" (Brooks, Seattle, Washington). They had a weight of

349 g and a heel-to-toe drop of 12 mm. The MIN shoes used were Altra One 2.5 (Orem, Utah). They had a weight of 178 g and a heel-to-toe drop of 0 mm. Both types of running shoe had no anti-pronation or anti-supination elements in the outsole. Participants were provided with identical running shoes to ensure that all participants were exposed to the same conditions during the study.

3.5 Experimental Protocol

Prior to running on the track, participants completed a series of running trials on a treadmill. The data from these trials were used to calculate the running economy and were not part of this thesis. As such, no further discussion of this portion of the protocol will be included in this thesis.

Following the treadmill running trials, participants were instructed to run on a 200 m unbanked mondo-surface indoor running track. They performed a minimum of seven 1000 m intervals at speeds within 94 - 97% of their maximal aerobic speed as determined on day one of testing. They were given a three-minute recovery between bouts. The duration for each interval was recorded (not including the recovery time). EMG data were collected for the duration of each of the running intervals. HR data were collected during running intervals using a Suunto, Ambit2 heart rate monitor (Suunto OY, Vantaa, Finland). At the end of each of the intervals, participants were asked about the rate of perceived exertion (RPE) to evaluate the level of central fatigue. The RPE was determined using Borg's category-ratio scale of 6-20 (Borg, 1973), shown in the appendix.

Blood lactate was also sampled during three of the rest periods (the first, third, and last interval) using a lactate analyzer (EKF Lactate Scout). Since lactate is a good indicator of the fast-glycolytic pathway that plays a major role in energy production during high-intensity bouts, it is one of the parameters that helps to quantify the existence of peripheral fatigue. Running intervals were repeated until the participants reached an RPE of 19. If this did not occur, the participants were asked to perform “all-out” running intervals until they reached an RPE of 19.

3.6 Data analysis

Methods used to analyze blood lactate, RPE, and HR were previously described by Blair (2018). No reanalysis of this data was done for the present thesis – results were taken directly from Blair’s work. The EMG analyzed was completed using custom-designed software written in MATLAB R2018b (MathWorks Incorporated, Natick, Massachusetts). The mean of the raw data was subtracted prior to starting the EMG analysis in order to remove any bias in the signal.

3.6.1 EMG amplitude

MVC trials were examined to determine the peak activation (MVC_{max}) of each muscle. Peak activation was determined using a 100ms moving window. This determined the RMS EMG for each MVC trial:

$$RMS = \sqrt{\frac{1}{T} \sum EMG(t)^2}$$

Once the RMS was determined, the peak activation value was selected and used to normalize the same muscle for the EMG data that were previously obtained from the fatiguing trials. This was done by dividing the raw EMG by the MVC_{max} .

The amplitude of the normalized EMG signal was determined in two ways – RMS EMG and iEMG. Both RMS EMG and iEMG were calculated for each minute of the running intervals. RMS was determined using the formula above while iEMG was calculated using the trapezoid rule.

3.6.2 EMG Frequency

Mean Frequency and Median Frequency:

Mean Frequency and Median Frequency were calculated using a function written in the MATLAB software for each minute of the running intervals.

3.6.3 Stride Frequency

Stride Frequency and Coefficient of Variation:

Stride Frequency represents how many strides a runner takes per minute. Since there were no kinematic data recorded in this study, the only variable we had to quantify Stride Frequency was EMG. We examined EMG for all tested muscles to determine which muscle could be used to identify strides consistently. We determined that VL exhibited peak activation during each contact phase. Based on these results, we calculated RMS EMG, using a 100 ms moving window, as shown in the example in **Figure 3**. Next, we applied a function in MATLAB to count the number of peaks during a 30-second time frame in the third minute of each interval. The stride frequency could then be determined

by doubling the peak number (i.e., multiplying the number of peaks by 2). Finally, random checks were performed on the RMS EMG using MATLAB to ensure the accuracy of the peaks we found.

The coefficient of variation is frequently utilized to compare outcomes from two different tests with different values or measures. The coefficient of variation of the stride frequency was calculated using stride frequency results for each of the intervals. The coefficient of variation (CV) calculation was done by dividing the standard deviation (SD) by mean (\bar{x}):

$$CV = (SD / \bar{x}) * 100.$$

3.7 Statistical analysis

Statistical analysis for heart rate, RPE, and blood lactate was completed previously by (Blair,2018). While the results of this analysis will be discussed in this thesis the reader is directly to Blairs thesis for a detailed description of the analysis of these variables. Statistical analysis for this study was completed using SPSS (IBM SPSS Statistics 25, IL, USA). Data were checked for the normality and sphericity. A two-way repeated measure analysis of variance (ANOVA), using time and shoe type as factors, was used to assess for differences. The ANOVA was completed on EMG amplitude (iEMG & RMS) and mean and median frequency, stride frequency and the coefficient of variation of stride frequency. EMG amplitude and frequency measures were compared between the first minute of the first interval and the last minute of the last interval. Stride frequency data were compared between the first and last intervals only. Additionally, an

ANOVA was previously performed on interval pace during the exercise-induced fatigue between the two shoe conditions (MIN VS. SH) and the time (first, fourth and last interval) (Blair, 2018).

Chapter 4: Results:

4.1 Participants' characteristics and training regime

A total of 10 individuals completed the study. The $\dot{V}O_{2\max}$ of the included participants ranged from the 95th to 99th percentiles (American College of Sports Medicine [ACSM], 2013). Participants were ranked significantly higher than typical recreational runners. Moreover, the participants' levels of aerobic fitness were affirmed in the velocity achieved before the point of exhaustion (18.1 km h^{-1}) (Blair, 2018). **Table 2a & 2b** show pace times (first, third, and last interval) from all subjects for both shoe conditions (MIN & SH).

Table 2a. A summary of the participants' target interval times (94% and 97% of maximum aerobic speed) during the shod condition. Also provided are the actual interval times participants ran during the first, middle, and last intervals in the shod condition. All times are in minutes: seconds.

Subject	Target Interval Time (94%)	Target Interval Time (97%)	Time (First Interval)	Time (Middle Interval)	Time (Last Interval)
2	3:34	3:39	3:28	3:34	3:31
3	3:24	3:18	3:28	3:27	3:35
4	4:11	4:03	4:01	4:06	4:01
5	3:30	3:23	3:23	3:26	3:25
6	3:29	3:36	3:24	3:30	3:27
9	3:19	3:13	3:11	3:09	3:10
10	3:22	3:15	3:09	3:15	3:18
11	3:36	3:29	3:44	3:28	3:39
12	3:18	3:24	3:26	3:15	3:20
13	3:36	3:29	3:30	3:42	3:29

Table 2b. A summary of the participants' target interval times (94% and 97% of maximum aerobic speed) during the MIN condition. Also provided are the actual interval times participants ran during the first, middle, and last intervals in the MIN condition. All times are in minutes: seconds.

Subject	Target Interval Time (94%)	Target Interval Time (97%)	Time (First Interval)	Time (Middle Interval)	Time (Last Interval)
2	3:34	3:29	3:26	3:26	3:19
3	3:24	3:18	3:24	3:23	3:11
4	4:11	4:03	3:49	4:00	3:50
5	3:30	3:23	3:20	3:18	3:14
6	3:36	3:29	3:33	3:31	3:29
9	3:19	3:13	3:04	3:07	3:08
10	3:22	3:15	3:22	3:14	3:08
11	3:36	3:29	3:36	3:28	3:34
12	3:24	3:18	3:17	3:15	3:02
13	3:36	3:29	3:33	3:33	3:50

4.2 Fatigue confirmation

As reported by Blair (2018), there were no interaction effects or main effects of footwear for either RPE, blood lactate or HR_{peak} . Despite the absence of significant blood lactate interactions, the blood lactate did indicate a significant main effect of time ($F_{(4, 32)} = 57.376$ and $p = 0.001$). The post-hoc test that was conducted indicated that blood lactate levels increased significantly at each time interval tested. RPE also indicated statistical significance, with a significant main effect of time ($F_{(4, 36)} = 95.947$ and $p = 0.001$). Results of the *post hoc* analysis revealed that each RPE score was significantly different compared to other RPE scores. This reflected the same trend as with blood lactate. As well, HR_{peak} showed a significant main effect of time ($F_{(2, 18)} = 21.954$ and $p = 0.001$), with pairwise comparisons clearly indicating a significant difference between the first interval HR_{peak} and the other two. While there was no effect of time on running pace, statistical analyses indicated that there was a main effect of shoe type on pace ($F_{(1, 9)} = 5.710$ and $p = 0.041$). Specifically, the analyses performed by Blair (2018) indicated that the participants ran faster during MIN intervals than SH ($3:25 \pm 0:15$ min/ interval vs. $3:28 \pm 0:17$ min/interval respectively).

4.3 EMG Amplitude

Root Mean Square (RMS) and Integrated EMG (iEMG):

No statistically significant main effects of time or interaction effects were noted between the first and last time point for RMS EMG (**Table 3a**). The only statistically significant main effect of shoes ($p < 0.05$) occurred for MG, as shown in **Figure 1**. There were higher RMS values observed in MIN conditions compared to those in SH conditions.

Table 3a: Statistical results for root mean square EMG between the first and last interval . Significant differences are identified in **bold** and occurred with $p < 0.05$. df indicates degrees of freedom

Muscle	Shoe			Time			Interaction (shoe*time)		
	df	F-statistic	p-value	df	F-statistic	p-value	df	F-statistic	p-value
BiF	(1,9)	1.274	.288	(1,9)	1.027	.337	(1,9)	.908	.366
Gmax	(1,9)	.129	.728	(1,9)	.791	.397	(1,9)	.415	.535
TA	(1,9)	.035	.856	(1,9)	1.267	.289	(1,9)	.026	.875
VL	(1,9)	.383	.551	(1,9)	1.153	.311	(1,9)	.463	.513
MG	(1,9)	5.573	.043	(1,9)	1.353	.275	(1,9)	.198	.667

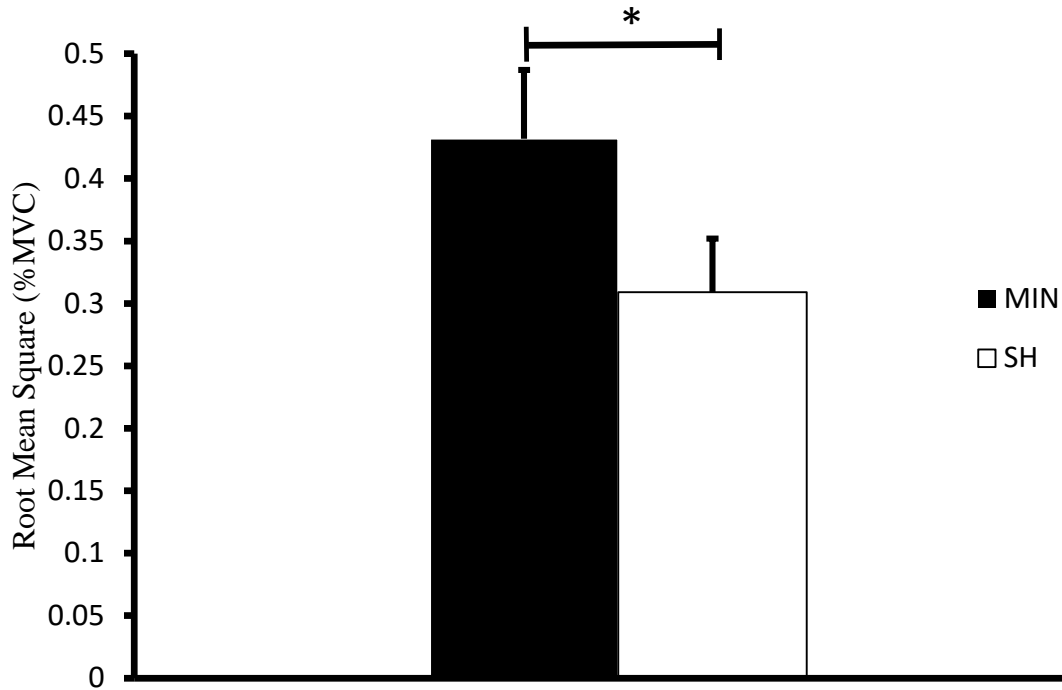


Figure 1. Medial gastrocnemius root mean square EMG as a function of shoe type (MIN vs. SH) collapsed across the first and last minute of the running intervals (* $p = 0.043$).

Findings of the current study revealed that no statistically significant differences were observed between the first and last minute for iEMG (**Table 3b**). Only one muscle showed statistically significant differences ($p < 0.05$), and only with the effect of wearing different shoes for MG. There were higher iEMG values observed in MIN conditions compared to SH conditions, as shown in **Figure 2**.

Table 3b: Statistical results for integrated EMG between the first and last interval. Significant differences are identified in **bold** and occurred with $p < 0.05$. df indicates degrees of freedom.

Muscle	Shoe			Time			Interaction (Shoe*Time)		
	df	F-statistic	<i>p</i> -value	df	F-statistic	<i>p</i> -value	df	F-statistic	<i>p</i> -value
BiF	(1,9)	.963	.352	(1,9)	1.099	.322	(1,9)	.672	.433
Gmax	(1,9)	1.391	.268	(1,9)	.048	.831	(1,9)	1.391	.268
TA	(1,9)	.023	.884	(1,9)	1.328	.279	(1,9)	.371	.558
VL	(1,9)	.445	.521	(1,9)	1.415	.265	(1,9)	.460	.515
MG	(1,9)	6.137	.035	(1,9)	.034	.858	(1,9)	1.170	.307

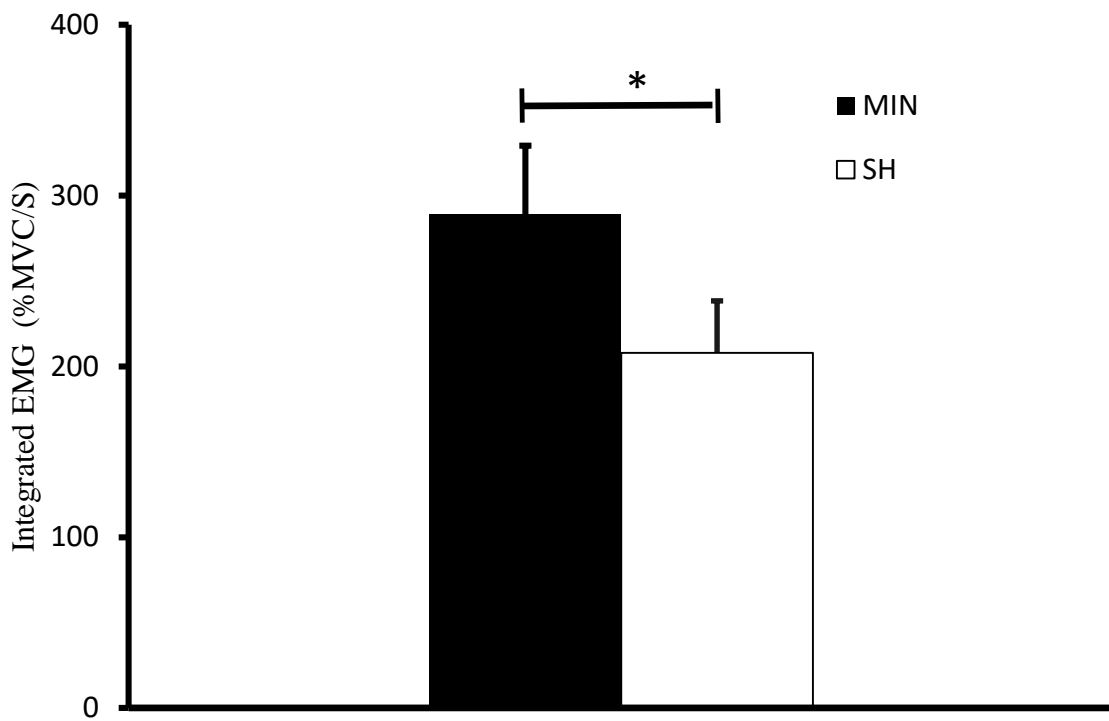


Figure 2. Medial gastrocnemius integrated EMG as a function of shoe type (MIN vs. SH) collapsed across the first and last minute of the running intervals (* $p = 0.035$).

4.4 EMG Frequency:

4.4.1 Median Frequency and Mean Frequency:

There was no significant difference ($p < 0.05$) in median frequency between the first minute of the first trial versus the last minute of the last trial point for all muscles examined, as shown in **Table 4a**.

Table 4a: Statistical results for Median Frequency EMG between the first and last trial. Significant differences are identified in **bold** and occurred with $p < 0.05$. df indicates degrees of freedom

Muscle	Shoe			Time			Interaction (Shoe*Time)		
	df	F-statistic	p-value	df	F-statistic	p-value	df	F-statistic	p-value
BiF	(1,9)	.229	.644	(1,9)	.188	.675	(1,9)	.736	.413
Gmax	(1,9)	1.013	.340	(1,9)	2.119	.179	(1,9)	.159	.699
TA	(1,9)	3.967	.078	(1,9)	.607	.456	(1,9)	.035	.855
VL	(1,9)	1.878	.204	(1,9)	1.795	.213	(1,9)	2.416	.155
MG	(1,8)	.177	.685	(1,8)	.749	.412	(1,8)	.282	.610

Likewise, there was no significant difference ($p < 0.05$) for mean frequency between the first and last time points for all muscles examined, as shown in **Table 4b**.

Table 4b: Statistical results for Mean Frequency EMG between the first and last trial. Significant differences are identified in **bold** and occurred with $p < 0.05$. df indicates degrees of freedom.

Muscle	Shoe			Time			Interaction (Shoe*Time)		
	df	F	<i>p</i> -value	df	F	<i>p</i> -value	df	F	<i>p</i> -value
BiF	(1,9)	.021	.888	(1,9)	.046	.835	(1,9)	1.414	.265
Gmax	(1,9)	1.965	.194	(1,9)	3.136	<i>.110</i>	(1,9)	.327	.582
TA	(1,9)	3.275	.104	(1,9)	1.785	2.14	(1,9)	.657	.439
VL	(1,9)	4.112	.073	(1,9)	3.105	<i>.112</i>	(1,9)	1.763	.217
MG	(1,8)	.218	.653	(1,8)	.012	<i>.915</i>	(1,8)	1.366	.276

4.5 Stride Frequency

Stride Frequency and Coefficient of Variation:

Statistical analysis revealed that there was only a significant main effect of the shoe on the stride frequency ($F_{(1,9)} = 9.151$, $P = .014$). As shown in **Figure 4**, the mean Stride Frequency values were higher in MIN condition (92.80 strides/minute) compared to SH condition (90.60 strides/minute). No other statistically significant differences were observed in the tests of within-subjects' effects for the effect of time ($F_{(1,9)} = 2.647$, $P = .138$) or the interaction between shoe and time ($F_{(1,9)} = .153$, $P = .705$).

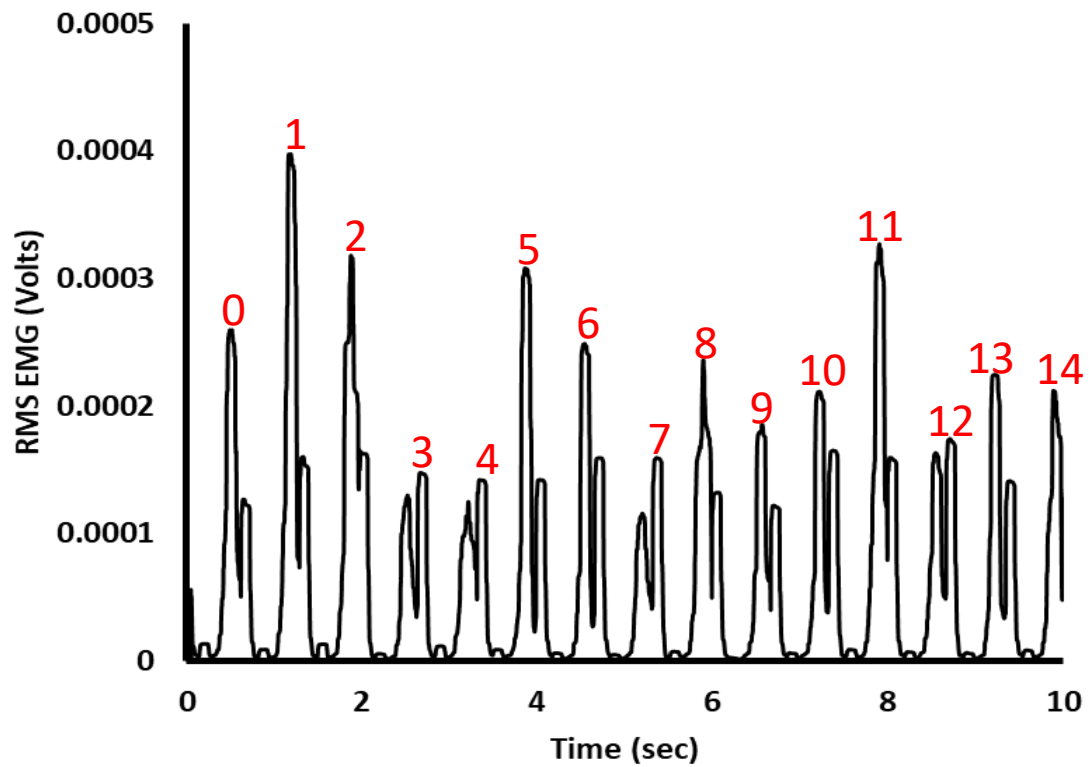


Figure 3. Sample data to illustrate how stride frequency was estimated using EMG data that was collected from vastus lateralis. The RMS EMG was calculated for 30sec of data taken from the middle minute of each fatigue interval (Note for clarity only 10sec of data is presented in this figure). Each peak of activation was associated with ground contact (as confirmed during pilot testing, which used video and EMG) and used to indicate the occurrence of a stride. Strides are indicated as numbers above the peaks in the figure above.

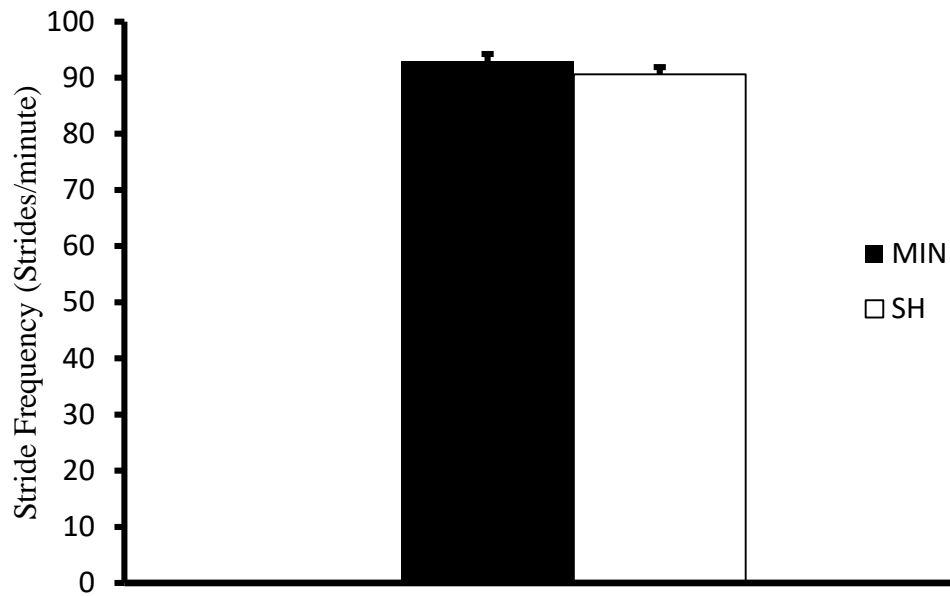


Figure 4. Stride frequency as a function of shoe type (MIN vs. SH) collapsed across the first and last running intervals. Stride frequency was significantly greater in the MIN condition.

No statistically significant ($p < 0.05$) differences were found between the first minute and last minute for the coefficient of variation of stride frequency for shoes ($F_{(1,9)} = 1.058, P = .330$), time ($F_{(1,9)} = .347, P = .570$), shoe * time ($F_{(1,9)} = .753, P = .408$).

Chapter 5: Discussion

This study examined the effects of running-designated footwear on muscle activation in five lower-limb muscles during exercise-induced fatigue. The participants were ten well-trained long-distance runners who wore two different types of footwear (MIN and SH) during two separate runs. The findings revealed that the footwear did not affect muscle activation as a function of peripheral fatigue and that only MG activation was affected as a result of wearing different running footwear. The findings also showed that there was an effect on stride frequency as a function of wearing different running footwear. These results could reflect that the MIN footwear used in the study was not minimalist enough, based on the Minimalist Index. Additionally, despite the fact that the intervals induced central fatigue, it is thought that the participants may not have reached a sufficient level of peripheral fatigue due to the interval nature of the training (intermittent runs) and the relatively long rest periods between intervals.

5.1 Confirmation of fatigue onset

In this study, participants ran at least seven intervals with a running intensity of 94-97% of their maximal aerobic speed. Participants were instructed to continue running intervals until they reached RPE of 19, which was considered as the fatiguing point, in addition to other physiological parameters such as, HR_{peak} , and blood lactate. All three of these measures are reliable indicators, which have been used to assess biological stressors (both physiological and psychological) during training or competition (Tucker & Noakes, 2009; Bourdon et al., 2017). The RPE integrates several variables to assess central fatigue

during exercise (Borg, 1982; Hampson, Gibson, Lambert and Noakes, 2001). In this study, RPE scores increased gradually during running intervals. Participants were considered to be fatigued and stopped exercising when they reached an RPE of 19. This RPE level meant that the participants experienced a level of exertion similar to as if they ran on a treadmill as fast and as hard as they could (Elsangedy et al., (2013). During high-intensity exercise, afferents from skeletal muscles and from various physiological systems are sent to the central nervous system to inform it about the potential physiological changes occurring during exercise. So, RPE plays a major role in the anticipatory regulation during exercise, not only as the integrator of afferent feedback from different physiological systems in humans (Tucker, 2009), but as a guard against homeostatis disruption (Tucker & Noakes, 2009). Based on this observation, the RPE of 19 was chosen to be the stop point in this study, and it was ultimately used to confirm the fatigue status of the participants. Observing blood lactate levels was the second method used to assess fatigue in participants. Increases in blood lactate levels during exercise cause the inhibition of contractile processes and diminish exercise performance (Cairns, 2006). In his results, Blair (2018) reported that participants in the study exhibited a significant time effect on blood lactate data collected during the first, fourth, and last interval, thereby confirming fatigue onset. The HR_{peak} was also used to assess fatigue, as it is a known method for monitoring changes during fatigue-induced training recovery status and to assess fatigue onset (Buchheit et al., 2014). Blair's study (2018) found a significant main effect of time on HR_{peak} between the first, fourth and last interval. These three parameters have been widely used to confirm the occurrence of whole-body fatigue,

and these results, as reported in the study by Blair (2018), were used for this study as well. Based on these results, the participants in the present study were considered to be experiencing central fatigue.

5.2 The effects of fatigue on muscle activation

Though the preceding measurements indicated that participants were exhibiting central fatigue at the end of the running intervals, this study was unable to confirm the hypothesis that muscle activation amplitude or frequency would decrease as a result of peripheral fatigue. Similarly, the study found no evidence to support the hypothesis that footwear type would have an impact on the effect of peripheral fatigue on muscle activation.

We consider the lack of effect of fatigue on muscle activation (amplitude and frequency) somewhat surprising. To better understand this, it is necessary to re-examine the relevant literature. Experts generally accept that fatiguing exercise causes a reduction of EMG amplitude during maximal levels of the MVC (Komi and Tesch, 1979). Tesch et al. (1990) found an increase in EMG amplitude during submaximal dynamic exercise. Similarly, contraction EMG amplitude increased during submaximal isometric contractions (Arendt-Nielsen and Mills, 1988; Maton, 1981). In the course of submaximal isometric muscle contraction, the rise in EMG amplitude emanates from recruiting additional motor units (Moritani, Nagata and Muro, 1982). The EMG signal can be influenced by various factors during dynamic contractions. These include: shifting in muscle fibers caused by changes in joint angle, rapid changes in motor unit recruitment

patterns, and changes in muscle force; these factors may cause changes in EMG characteristics that are different from those recorded during static contractions (González-Izal, Malanda, Gorostiaga & Izquierdo, 2012).

The picture is more complicated when one considers the effects of fatigue on muscle activation during dynamic tasks (e.g., running), and this may contribute to the inconsistency in the literature. Many authors have reported decreases in muscle activation during exhaustive running (Billaut & Smith, 2010; Jewell, Boyer & Hamill, 2017; Jewell, Hamill, von Tscherner & Boyer, 2019; Lepers, Pousson, Maffiuletti Martin & Van Hoecke, 2000; Millet et al., 2002; Mendez-Villanueva et al., 2007, 2008; Mizrahi et al., 2000; Newton, 2010; Nicol et al., 1991; Paavolainen et al., 1999; Racinais et al., 2007; Vuorimaa et al., 2006; Weist et al., 2004). Other studies have identified an increase in muscle activation (Newton 2010; Hausswirth, Brisswalter, Vallier, Smith & Lepers, 2000; Cheung & Ng, (2010). Even so, most authors have found that key muscles exhibited lower activation towards the end of fatigued running. The current study found no changes in muscle activation as a function of fatigue. Similarly, no changes were found in EMG frequency, a result that also conflicts with the majority of studies (Ament, Verkerke, Bonga & Hof, 1996; Ament, Bonga, Hof & Verkerke, 1993; Hashish, Samarawickrame, Baker & Salem, 2016; Hausswirth, Brisswalter, Vallier, Smith & Lepers, 2000; Mizrahi, Verbitsky, Isakov & Daily, 2000; Mizrahi, Verbitsky & Isakov, 2000), which have reported decreases in EMG frequency following fatigue induced by running. Both of these were surprising results and warranted further examination. Three possible reasons have been identified to explain the lack of decline in EMG amplitude

and frequency with fatigue. The first relates to the limitations of using EMG to assess fatigue during dynamic contractions and the two others involve the nature of the exercise – induced fatigue protocol used for the study.

Despite the widespread use of EMG as a method to assess fatigue during dynamic contractions, several authors have highlighted the issues with using common EMG collection and analysis methods for this purpose (see Cifrek, Medved, Tonković & Ostojić, 2009 for example). The main issue that arises when collecting EMG during dynamic movements (e.g., running), is that the movement itself can negatively affect the quality of the EMG signal. For instance, attaching surface EMG electrodes to the skin while collecting EMG signals during dynamic contractions can result in the crosstalk of myoelectric signals from neighbouring muscles on EMG signals (Farina, Merletti, Indino, Nazzar & Pozzo, 2002). Dynamic contractions also result in movement of electrodes relative to the target muscles, which will also affect the obtained signal (Farina, Merletti, Nazzaro & Caruso, 2001). Moreover, during dynamic movements, the force and joint angle vary with time, because EMG signals are assumed to be non-stationary, due to some dynamic factors (e.g., the recruitment and de-recruitment of working motor units surrounding EMG electrodes). This variation causes increased variability in the signal and affects the quality and reliability of EMG signals (Farina et al., 2001; Rogers & MacIsaac, 2011). As such, a more complex, multivariable approach of EMG analysis has been suggested (Cifrek, Medved, Tonković & Ostojić, 2009; Rogers & MacIsaac, 2011). Based on this, the lack of decreasing EMG could be an artifact of the particular method

of EMG analysis utilized in the present study. Such an analysis is currently being considered for the current data set.

Although the importance of the previous issue cannot be downplayed, there are two issues with the fatiguing trials that the author also feels may have contributed. The first relates to the type of fatiguing trials used. Most of the previous running investigations mentioned above used continuous, longer duration running to induce fatigue, a key difference between the present study and most others that have reported decreases in EMG with fatigue. This is in contrast to the high-intensity intervals used in the present study. It is possible that it was this difference in fatigue intervals that contributed to the lack of decrease in EMG in the present study. Research by Vuorimaa et al. (2006) provides some support for this hypothesis. These authors found that when long-distance runners performed different types of running exercise (e.g., continuous, interval or incremental running), the effect of lactate accumulation on performance differed based on the type of running performed. For instance, during interval running, an increase of blood lactate was associated with the increase of exercise intensity, while during continuous running, the blood lactate was only slightly elevated (Vuorimaa et al., 2006). Although participants in the Vuorimaa research were not performing exhaustive runs, this study does indicate that interval running seems to produce a different blood lactate response than continuous running. It is possible that in the current study, when performing exercise-induced fatigue using an interval training protocol, peripheral fatigue may not have developed in the same manner as it would have if the continuous fatiguing

run were performed. This may help to explain why no changes in EMG were observed in the current study.

A second issue related to the interval trials was the length of the rest period between consecutive intervals. The 3- minute recovery time, coupled with the highly trained nature of the participants, may have enabled individual muscles to more fully recover during the rest period. As a result, peripheral fatigue may not have occurred to as great an extent as would have been expected with a shorter recovery time. Although several authors have examined the impact of recovery duration on the physiological response to interval training (Schoenmakers and Reed, 2019; Seiler and Hetlelid, 2005; Smilios et al. 2018) no studies were found that specifically examined how recovery duration affected muscle activation as a result of fatigue. However, Smilios et al. (2018) did report that a longer duration of recovery (4 minutes as opposed to 2 minutes) did result in lower physical and cardiovascular stress during interval training, providing some support of the fact that 3 minutes of recovery was too long to fully allow peripheral fatigue to develop. Hence, additional research is needed in this field.

5.3 The interaction of footwear and fatigue

The study's primary objective was to examine the effect of footwear on muscle activation during running trials that produced fatigue. The hypothesis that MIN footwear would result in less muscle fatigue and a reduction in lower limb muscle activation, compared to these factors observed in the SH footwear condition, was not confirmed. The hypothesis was based on previous studies which confirmed that both MIN footwear and

fatigue have been shown to independently affect running biomechanics (Kasmer et al., 2016; Tam et al., 2017; Siler and Martin, 1991). If either the footwear or the runner's fatigue failed to affect running mechanics, then there would be a reduced likelihood of observing an interaction between the two factors. It is suggested that there are two reasons that likely explain why participants did not exhibit changes in muscle activation during fatiguing trials when wearing MIN footwear versus SH footwear: the magnitude of the effect of MIN footwear on participants, and the nature of runner's fatigue induced by the interval trials. The details of both factors will be discussed more thoroughly below.

To explain the first factor (i.e., the magnitude of the effect of MIN footwear on participants), the MIN footwear utilized in the study may have been insufficiently minimal in relation to the SH footwear. Blair (2018) has previously suggested this factor in his attempt to explain the lack of effect of footwear on running economy, using the same participants as in the present work. Blair suggested the MIN footwear used was not 'minimalist' enough in comparison to the SH footwear. According to Esculier, Dubois, Dionne, Leblond and Roy (2015), the minimalist index is a reliable guide for evaluating footwear's degree of minimalization for running research; this guide uses a grading scale, with higher scores signifying a more minimalist shoe. The MIN footwear utilized in this study had a minimalist index of 70%, which is close to the cutoff between MIN footwear and SH footwear. Ideally, MIN footwear should have a score closer to 100%, to more appropriately characterize the barefoot running condition. The work of researchers that have used MIN footwear with scores above 70% on the minimalist index supports this hypothesis and has reported significant changes in running biomechanics (e.g. strike

pattern, contact time, stride length) compared to SH footwear or BF (Bergstra et al., 2015; Fredericks, 2015; Gillinov, Laux, Kuivila, Hass, & Joy, 2015). These findings can help to explain the importance of the use of the minimalist index when comparing running footwear. Therefore, it can be reasonably assumed that the lack of effect of MIN footwear on muscle activation during exercise-induced fatigue in this study was, at least in part, the result of the low minimalist index of MIN footwear.

A second factor which likely explains the findings of this study is the manner in which fatigue was induced in participants. As discussed in detail previously, the interval runs used to induce fatigue may not have resulted in sufficient peripheral fatigue. In fact, it appears as though participants likely experienced more central fatigue during these runs. The relative absence of peripheral fatigue makes it unlikely that any effect of footwear on fatigue would have been observed – if no peripheral fatigue was present, then how could footwear be expected to affect this fatigue? As mentioned above, future studies using continuous long-duration fatiguing trials using both MIN and SH footwear are likely needed to provide more insight into this issue.

5.4 Speed

An interesting finding from Blair (2018) was that participants ran faster in the MIN condition compared with the SH condition. The average run time to complete all intervals (an average of 7 per participant) was lower in the MIN condition (3:25 ± 0:15 minutes) compared to the SH condition (3:28 ± 0:17 minutes). Blair suggested that this finding could be due to the reduced mass of the MIN shoes, a hypothesis that has been

supported by other authors (Divert et al., 2008; Kasmer et al., 2016). Divert et al. (2008) and Kasmer et al. (2016) concluded that the reduced mass of the shoe was linked to less oxygen consumption. Blair (2018) also suggested that the increase in running speed was due to MIN footwear altering some running kinematics (e.g., contact time); however, because of a lack of kinematic data, the author was unable to confirm this hypothesis. Therefore, the present analysis examined the latter suggestion to determine whether the stride frequency was altered in the MIN condition.

There are two ways that runners increase their running speed; either by altering stride frequency (Högberg, 1952) and/or stride length (Hoyt, Wickler & Cogger, 2000). In the present study, the mean stride frequency values in the MIN condition (92.8 strides/min) were higher when compared with SH conditions (90.6 strides/min). The increase in stride frequency in MIN running suggests that this was at least one of the mechanisms runners used to increase their running speed. These findings are similar to previous researchers' findings which reported an increased stride frequency in MIN running (Lussiana et al., 2013; Squadrone et al., 2015; Bonacci et al., 2013; Divert et al., 2008). The type of running footwear (e.g., MIN or SH) has been found to be linked to changes in stride frequency. For example, Hollander, Argubi-Wollesen, Reer & Zech (2015) found that when running BF, individuals had a higher stride frequency compared to those using uncushioned MIN, cushioned MIN and SH. These findings are in line with the findings of the current study, with a higher stride frequency and faster running speed in the MIN condition. Stride frequency has also been found to be linked to an increase in lower limb muscles (e.g., MG), mainly during the late swing phase, and this seems to be

used for the anticipation of foot-ground contact (Chumanov, Wille, Michalski & Heiderscheit, 2012). While the higher stride frequency provides insight into how participants were able to increase running speed in the MIN condition, this finding also helps to interpret the increased MG activation observed in this condition.

Overall, when running in MIN, runners have a tendency to increase the contribution of their ankles' plantar flexor muscles (e.g., MG muscle) (Paquette, Zhang & Baumgartner, 2013) in order to eccentrically control the positioning of the heel to the ground following a midfoot or a forefoot strike (Lieberman et al., 2010). Based on this, the obvious conclusion to draw about the increased MG activation observed in the MIN condition is that it occurred as a result of a change in foot strike pattern from rearfoot to mid/forefoot strike (De Wit, De Clercq & Aerts, 2000; Divert et al., 2005; Squadrone et al., 2015). This is unlikely to be the case, however, as Blair (2018) reported that about 75% (MIN) and 76% (SH) of participants in the present study used a rearfoot strike pattern. Although Blair's assessment of foot strike pattern was carried out while participants ran on a treadmill, the findings do strongly suggest that most participants likely maintained a rearfoot strike pattern when they ran in the MIN condition. If forefoot striking did not require the increase in MG activation, why then was it observed? According to Hamner, Seth & Delp (2010), the MG muscle has been shown to be one of the main contributors to both forward accelerations of the center of mass (forward propulsion) in addition to the upward acceleration of the body mass center (support) during running. As such, it is hypothesized that the increase observed in MG activation occurred to create the additional forward propulsion needed to increase running speed.

This increase in MG activation, which would typically result in an increased cost of running, was likely made possible by the reduced mass of the MIN shoes (178g vs. 349g for SH), enabling participants to run at a higher intensity without experiencing peripheral fatigue. Blair (2018) and Cheung & Ngai (2016) have suggested that the mass effect of MIN footwear is likely the primary factor involved in improvements in running performance when wearing MIN footwear compared to SH footwear. Clearly, a limitation to this suggestion is the fact that Blair's assessment of footstrike pattern happened during the treadmill running. It is possible that participants used a different foot strike pattern during the interval trials. This would need to be confirmed by replicating the study and recording kinematic data.

5.5 Limitations

One major limitation in this study was that MIN footwear had a relatively low minimalist index, which the author feels affected the results of the current study. As discussed above, the MIN footwear chosen for the present study was likely not minimalist enough to require changes in running kinematics/biomechanics. Using MIN footwear with a higher minimalist index (i.e. closer to 100%) would have likely had a greater effect on lower limb muscle activation, resulting in different fatigue effects when wearing MIN footwear compared to SH footwear.

A second limitation is that fatigue intervals completed in this study appear to have produced central fatigue rather than peripheral fatigue. The lack of peripheral fatigue contributed to show no changes in muscle activation or any interaction between fatigue

and footwear. Inducing peripheral fatigue would likely have resulted in changes in muscle activation and would also help to determine if there is an interaction between fatigue and footwear.

Another limitation of this study is that stride frequency was estimated by using EMG data. The gold standard for such measures would be kinematic data; however, the current investigation did not measure any running kinematic data during fatigue intervals. Because of this, we had to have an alternative method to calculate stride frequency (by counting the number of peaks of RMS EMG). Based on our analysis, we are confident that using the peak activation method to quantify stride frequency is of high accuracy. Future studies, incorporating kinematic data fatiguing trials, would help to confirm this finding.

5.6 Conclusion

The present study examined the interplay of running footwear on lower limb muscle activation during exercise-induced fatigue. The findings revealed that there was no interaction effect between MIN footwear and fatigue on either muscle activation amplitude or EMG frequency. However, stride frequency was higher during running in MIN footwear compared to running in SH footwear, which provides insight into the mechanism by which the increase in speed found by Blair (2018) was achieved. The lack of effect of footwear on peripheral fatigue is suggested to be a reflection of three methodological flaws. First, the MIN footwear used in this work had a low minimalist index (70%), which means it was not sufficiently minimalist to differ from the SH

footwear used in the same study. Secondly, despite the presence of central fatigue, as confirmed using accepted measures, the study's participants may not have experienced an adequate level of peripheral fatigue due to the long rest period between the intervals as well as the interval running protocol utilized in this study. Thirdly, the use of EMG to assess fatigue during dynamic contractions has been questioned by several authors (Farina, 2006; Farina et al., 2001; Gonzalez-Izal et al., 2010; Maclsaac, Parker, Scott, Englehart and Duffley, 2001), and as such may not have been an appropriate measure to assess peripheral fatigue in this study. Consequently, this study was unable to conclude whether MIN footwear had any effect on muscle activation compared to SH during exercise-induced fatigue.

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Appendix: Borg 6-20 Rate of perceived exertion (RPE) scale

<i>Rating of Perceived Exertion Borg RPE Scale</i>		
6		How you feel when lying in bed or sitting in a chair relaxed. Little or no effort.
7	Very, very light	
8		
9	Very light	
10		
11	Fairly light	
12		Target range: How you should feel with exercise or activity.
13	Somewhat hard	
14		
15	Hard	
16		
17	Very hard	How you felt with the hardest work you have ever done.
18		
19	Very, very hard	Don't work this hard!
20	Maximum exertion	