INVESTIGATING FOOTWEAR BIOMECHANICS CONCEPTS IN ‘HEALTH AND WELL-BEING’ FOOTWEAR

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Declaration

As a PhD by published works, this thesis comprises seven papers which have previously been accepted in, or submitted to, peer-reviewed journals and have been, or are to be, published in the public domain. The papers have been written in collaboration with co-authors and the extent to which the author contributed to each paper is defined in Part C and verified by collaborating authors in Appendix A. Abstracts relating to the work included have been presented at international conferences and these abstracts and posters are included in Appendix C.

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Abstract

'Health and well-being' footwear positions itself in the footwear market between high street footwear and specialist therapeutic footwear. Manufacturers in this footwear category promote benefits when compared with standard footwear. However, the full exploration and validation of such proposed benefits requires scientific exploration through the application of footwear biomechanics concepts and techniques. The studies herein were undertaken to assess these biomechanical concepts in 'health and well-being' footwear, particularly in FitFlop^{TM} footwear. The studies are experimental studies with repeated measures designs. A total of 128 individual participants volunteered, 28 of which were included in two publications. Variables were quantified using an in-shoe plantar pressure measurement system (with a bespoke insole), electromyography, 3D motion capture, force plates, accelerometers, a modified questionnaire and a custom-made mechanical drop-test device. The research identified that 'health and well-being' footwear can be manipulated to increase shock absorption, namely reducing the heel-strike transient magnitude (-19%) compared with a flip-flop. 'Health and well-being' footwear does induce instability at specific phases of the gait cycle, which is specific to the outsole shape of the footwear. For example the MBT shoe increased muscle activity relating to controlling sagittal plane motion. The biomechanics of gait are also altered compared to standard footwear styles, such as reducing the frontal plane motion of the foot in stance (-19%) and the magnitude (-86%) and duration (-98%) of gripping with the Hallux in swing compared with a flip-flop. The tested 'health and well-being' footwear was subjectively rated equally as comfortable as a control shoe with increased regional pressures in the midfoot (≈25%) and decreased peak pressures in the heel (-22%). Therefore 'health and well-being' footwear may influence the biomechanics of wearers however further exploration of meaningful differences and individual population differences is required. The studies emphasise the importance and relevance of testing walking, as well as running, footwear to the wider footwear biomechanics field and demonstrate how this may be integrated into research and development processes within a footwear company.

Chapter 1 Thesis Overview

1.1 Introduction

The motivation to undertake the body of work contained in this thesis was multi-faceted. The first motivation was that the work formed part of a project with a commercial footwear company which aimed to undertake product development through testing of their footwear and to benchmark this footwear against relevant competitors for marketing purposes. An aim therefore was to provide data on the influence of FitFlop™ footwear on walking and standing compared with relevant comparator footwear, thus, to contribute to the research and development and marketing of the company's products. This provides a 'real life' example of the integration of biomechanical data and knowledge to add value in the footwear product cycle. To achieve this aim, research questions were developed relevant to both the academic and footwear industry communities following systematic and objective critical appraisals of existing literature and data. In addition to results and interpretation being provided to the company, peer-reviewed publications were accepted, contributing to the wider research field.

Additional motivation for the work was to undertake research on the biomechanics of walking footwear, which is scarce in the existing body of literature. The field of footwear biomechanics is dominated by testing and development of running footwear or specialist therapeutic footwear (e.g. footwear for adults with diabetes). Hence there was a need demonstrate the importance of biomechanical testing in walking footwear. A wide range of experimental designs, protocols, instrumentation and analysis were used. New approaches/protocols were also developed to specifically address the quantification of characteristics of walking footwear which cannot be validly measured with existing protocols, which were designed for running. A further aim, therefore, was to modify testing protocols and methods used for running footwear in order to provide relevant data for walking footwear. This, ultimately, would provide testing methodologies and protocols that could be incorporated into footwear research and development within footwear companies or footwear technology centres in the future.

There has been a recent growth in the 'health and well-being' footwear market. This is footwear that is marketed to the general public as being more comfortable (e.g. Hotter™), or to challenge stability (e.g. Masai Barefoot Technology^{TM}), or to mimic barefoot walking (e.g. Vivobarefoot^{TM}). However, the biomechanical investigation and influence of this footwear category on wearers gait is yet to be fully determined and thus any proposed benefits remain largely unsupported by scientifically rigorous data. This footwear category adapts traditional aesthetic expectations of casual footwear and has features such as thicker soles, rocker soles, wide fitting uppers and secure fitting, which adapt aesthetics in order to reportedly deliver specific functional aims. This enables a modification of traditional footwear design to meet directed outcomes as long as any benefits can be demonstrated and conveyed to wearers. Consequently, 'health and well-being' footwear is the first non-therapeutic or sport footwear category that has attempted, or had the opportunity, to fully embed biomechanical principles and testing in the research and development processes, unlike standard high-street retail footwear. The number of manufacturers and ranges of footwear in the 'health and well-being' footwear category is increasing. However any benefits of specific products need to be quantified and conveyed to wearers, as opposed to companies relying on claiming benefits. There may be health benefits from some specific aspects of these footwear styles, however further research is required to establish this. The final motivation of the thesis, therefore, was to apply concepts in footwear biomechanics 'health and well-being' footwear to explore the functionality of this footwear category, which can be quantified with available physiological and biomechanical techniques.

1.1.1 Footwear Biomechanics Concepts

The development of the concepts to be explored in this thesis was based on the outcomes of a literature review undertaken at the outset of the Knowledge Transfer Partnership (KTP). This literature review considered the material available relating to 'health and well-being' footwear in addition to a broad assessment of footwear biomechanics literature. Dissemination material from footwear companies including marketing material press releases, research studies and technical sections on websites were reviewed. Additionally, recent literature relating to 'health and well-being' footwear was reviewed including scientific studies undertaken in magazines and peer-reviewed research in journal articles. Anecdotal testimonials relating to use of footwear and alleviation of symptoms were also considered. From this analysis of available material it was determined that literature pertaining to 'health and well-being' footwear could be categorised as quantifying:

- Shock absorption properties of footwear

- Variables denoting instability in footwear
- Gait modifications and changes in response to specific footwear styles.
- Footwear comfort and associated variables

These concepts encompass the majority of proposed or reported benefits from ‘health and well-being’ footwear and running footwear companies. They also represent concepts and variables which can be readily quantified utilising biomechanical techniques. These topics were described as “Shock Absorption”, “Instability”, “Gait Modifications” and “Comfort” to define the sub-sections of the research within this thesis.

1.1.1.1 “Shock Absorption”

Footwear biomechanics research focuses on running footwear and the protocols utilised in this field are well validated and reviewed to assess specific requirements relating to athletic footwear (e.g. cleated footwear or running cushioning systems). For example, impact testing in footwear focuses on the ASTM (American Society for Testing and Materials) F1614-06 protocol, which impacts the footwear with 5 Joules of energy, as quantified in running impacts (Cavanagh et al., 1984). Footwear research uses this impact energy to quantify impacts in different thickness and hardness midsoles, heel flares, military footwear and worn footwear for running (Dixon et al., 2003; Frederick et al., 1984). These protocols are replicated in the testing of footwear or insoles for walking in footwear biomechanics literature (Nordin and Dufek, 2012) and by the Shoe and Allied Trade Research Association (SATRA), the U.K. footwear testing body. The first footwear biomechanics concept to be explored within Paper 1 investigated shock absorption properties in walking footwear. A protocol was developed as part of this work to test walking footwear. This promotes the concept to the field that making an adaptation to the current running protocols is more appropriate for testing or assessing walking shoes. Simply using the same methodology as traditionally utilised for running footwear is not sufficient. The paper developed a methodology for this approach and then utilised both human and mechanical testing to compare impact characteristics in a range of walking shoes (e.g. trainers, flip-flops). The protocol was then implemented in Paper 2 alongside walking data to compare impact in a range of different hardness and thickness footbeds.

1.1.1.2 “Instability”

Numerous footwear companies have developed “unstable” footwear styles which aim to reduce the stability of the wearer and increase muscle activation in the wearer. The original premise of this footwear style appeared to be to make walking more like barefoot or more demanding for the wearer (e.g. Masai Barefoot Technology™, Reebok EasyTone™). Research papers relating to instability have critiqued and compared a range of commercially available footwear styles which promote themselves as unstable (Porcari et al., 2009). Identifying differences between these designs and technologies is informative for clinicians and wearers alike to provide a comparison of what footwear is available and relate it to their specific symptoms or aims. Currently this footwear category is termed "unstable" as opposed to considering the specific features that are producing the instability, the nature of the instability and which wearer's symptoms or aims specific footwear might be most appropriate for. The focus of research in this footwear category is rocker-shoe styled footwear and more specifically Masai Barefoot Technology (Buchecker et al., 2012; Landry et al., 2010; Nigg et al., 2010). Other research has utilised bespoke modified footwear that cannot be related back to specific commercial styles for use by the general population (Hömme et al., 2012). Some recent research has tested commercial footwear, but does not present the brand names or shoe features such that the wearer or clinician cannot draw conclusions to drive a purchase or prescription from the publication (Germano et al., 2012). The second footwear biomechanics concept to be investigated within this research (in Papers 3&4) was instability; explicitly, the quantification and comparison of instability in single-leg standing and walking in a range of commercially available unstable footwear. The footwear has been identified by name to enable wearers and clinicians to make full use of study findings and the comparison of any findings has been related back to the footwear midsole and outsole features.

1.1.1.3 "Gait Modifications"

Gait modification to fashion footwear styles have been reported including high-heels (Lee et al., 2001; McBride et al., 1991) and flip-flops (Carl and Barrett, 2008; Shroyer et al., 2010). Despite the popularity of the footwear style, localised heel pain and other conditions such as overuse injuries of the tibialis anterior and toes are implicated for the wearers of flip-flops by podiatrists (American College of Foot and Ankle Surgeons, 2007). Flip-flops defy recommendations for footwear by being thin, not supporting the medial arch, not protecting the toes, having a loose fitting upper and having no pitch from heel to toe (Barton et al., 2009; McPoil, 1988). However at the outset of this research (2009) there exists minimal data concerning this footwear style and how it influences gait. The literature in this field compares walking kinematics in flip-flops (Shroyer, 2009) and quantifies plantar pressures (Carl and Barrett, 2008). However, the work undertaken does not present plantar pressures and has some methodological weaknesses, including not controlling walking velocity when quantifying plantar pressures and comparing kinematics from 2D digitised video data. Additionally, the work does not compare flip-flops to relevant control conditions, such as a different design of toe-post footwear, which may remove some of the concerns that clinicians currently voice. Thus, a more thorough exploration is required. The third footwear biomechanics concept of 'health and well-being' footwear, and thus this body or work, was to describe and define walking in flip-flops and how this affects or modifies plantar pressures and gait. Papers 5&6 raise areas for future study investigating toe-post footwear and highlight some biomechanical implications of the footwear which may relate to pathologies or predispose wearers to the lower limb overuse injuries widely reported by healthcare professionals.

1.1.1.4 "Comfort"

The assessment of comfort is an aspect of footwear that is widely studied utilising subjective measures such as questionnaires (Mills et al., 2011; Mündermann et al., 2002) and interviews (Kouchi, 2011), and objective measures such as plantar (Che et al., 1994; Jordan et al., 1997) and dorsal pressures (Hagen et al., 2010). Quantifying aspects that relate to foot comfort in a shoe is essential for footwear manufactures and designers to produce footwear which is favourable for their consumers. Literature pertaining to comfort reports quantification of objective measures such as ground reaction force at impact (Lake and Lafortune, 1998; Whittle et al., 1994), plantar pressures (Che et al., 1994; Jordan et al., 1997) and subject features such as foot size and sensitivity (Miller et al., 2000). The measurement of subjective outcomes generally use a comfort questionnaire which was specifically designed and validated for runners and running footwear with varied insoles/orthotics in a trainer (Mündermann et al., 2002; Zifchock and Davis, 2008). Hence the fourth footwear biomechanics concept to be investigated within this research was comfort through developing a 'comfort protocol' including a modification of a well-published comfort scale. Comfort was quantified in two footwear styles subjectively and objectively with results subsequently being compared and discussed relating back to footwear and wearer features. This concept, again, addresses the requirement for walking shoe and walking gait specific protocols in footwear biomechanics for realistic testing and development, which could be integrated into footwear product cycles.

1.1.2 Approach

This thesis is a working example of how footwear biomechanics data can be integrated into a company's footwear design, development and marketing processes to provide knowledge transfer and ultimately economic value. Additionally, this work emphasises the importance of footwear testing and development in a commercial footwear environment, not just for athletic shoes, but for footwear produced for daily wear by the general population. A recent survey suggests only 2 of the 50 million adults in the UK take part in athletic activities (including running and jogging) for at least 30 minutes at least once a week (Sport England, 2012). The activity of walking and walking footwear are more relevant to general and clinical populations than running footwear. However, given that most footwear research focuses on athletic footwear, this work fills the gap in knowledge and emphasises the importance of considering the appropriateness and function of all footwear. The publications contained in this thesis provide valuable and detailed information to footwear consumers, technologists, researchers and manufacturers alike around the concepts of quantifying and comparing gait kinematics in footwear styles, instability, shock absorption and comfort in walking footwear. These are relevant footwear biomechanics concepts to apply to this relatively new category of 'health and well-being' footwear. Furthermore, the work provides a novel protocol to assess the shock absorption properties of walking footwear, reliable plantar-pressure data when walking in flip-flop style footwear and a comparison of the nature of the instability from walking and standing in commercially relevant instability footwear. The scope of this thesis therefore includes quantifying the immediate influence of 'health and well-being' footwear on the biomechanics of wearers with a focus on four specific considerations.

The nature of the research within this thesis is quantitative in relation to the data collection and data analysis. The research approaches for data collection were trials with repeated measures designs with healthy volunteer subjects undertaken in gait laboratories at the University of Salford. The research utilised an array of methodologies in order to quantify gait in walking footwear in representative populations (Table 1.1). The methodologies were generally drawn from standard gait laboratory practices, footwear testing research, industry standards, and the application of wider biomechanical techniques (e.g. balance measurement) to variables of interest. Modifications were undertaken to general protocols to increase the relevance to walking footwear (e.g. mechanical test device and comfort questionnaire). Testing utilised 3D motion capture, electromyography, force plates, in-shoe pressure measurement, accelerometers, a bespoke mechanical impact device, foot switches and a questionnaire. Data was captured utilising Qualisys (Gothenburg, Sweden), MyoResearch XP (Noraxon Inc., Scottsdale, Arizona, USA) and Medilogic (T&T Medilogic, Gmbh, Germany) software packages. Data processing and analysis was undertaken in Visual 3D (C-Motion Inc., Rockville, Maryland, USA), Matlab (MathWorks, Cambridge, UK) and Microsoft Excel (Microsoft, Washington, USA) using custom-written models, pipelines, scripts and templates written by the author. Statistical comparisons were undertaken using Statistical Package for Social Sciences V17 (SPSS Inc., Chicago, U.S.A.).

Table 1.1 Research equipment and participant overview.

Footwear Biomechanics Concept	**“Shock Absorption”**	**“Instability”**	**“Gait Modifications”**	**“Comfort”**
Data collection	3D motion capture (Qualisys), Accelerometer, Force plates (AMTI), Mechanical Impact Device.	3D motion capture (Qualisys), Electromyography (Noraxon), Force plates (AMTI).	In-shoe pressure (Medilogic), 3D motion capture (Qualisys), Electromyography (Noraxon), Force plates (AMTI).	In-shoe pressure (Medilogic), Accelerometer (Noraxon), Foot Switch (Noraxon), Comfort questionnaire, Mechanical Impact Device.
Analysis Software	Visual 3D, Matlab	Visual 3D, Microsoft Excel	Visual 3D, Matlab, Microsoft Excel	Visual 3D, Matlab
Participants	N = 13: Paper 1 N = 13: Paper 2	N = 15: Paper 3 N = 15: Paper 4	N = 20: Paper 5 N = 40: Paper 6	N = 40: Paper 7
	2 Male 11 Female	15 Female	20 Male 40 Female	40 Female

1.2 Timeframes

In order to give context to the literature reviews and rationale/justification for the work the timeframes of the studies within this body of work are relevant (Figure 1.1). The literature reviews address the existing literature base and footwear research and technology state, which led to the study definition and aims. Further literature and interpretation was included in the paper drafting process and throughout the review and publication process. Following this, the critique of the papers places the research in the existing literature field and reviews the addition to the knowledge base from this body of work.

Footwear Biomechanics Concept	Study Conception/ Literature Review	Data Collection	Paper First Submission	Paper Accepted
"Shock Absorption"	July 2010	October 2010	April 2013/ January 2014	January 2014/ In review
"Instabilty"	August 2010	July 2010/ July 2011	August 2012/ January 2013	January 2013/ March 2013
"Gait Modifications"	May 2009/ June 2010	September 2009/ June- November 2010	September 2012/ June 2012	March 2013/ September 2014
"Comfort"	June 2012	September 2012- February 2013	October 2013	To submit

Figure 1.1 Timeline and timeframes for the studies and papers within the thesis as of September 2014.

1.3 Main Objectives

The main objective of this body of work was to measure aspects of 'health and well-being' footwear related to footwear biomechanics concepts which have been related to the footwear. The data and research included in this these formed part of a Technology Strategy Board funded research project (KTP) with a commercial footwear company (FitFlop ltd). This aimed to undertake product testing for research and development and marketing purposes. Another objective therefore was to provide data on FitFlop™ footwear for the company to utilise for research and development and marketing purposes. The nature of this body of research as a collection of work aimed to provide data on the influence of FitFlop™ footwear on walking and standing, demonstrate the importance of biomechanical testing in walking footwear, modify testing from running footwear protocols for walking footwear and footwear biomechanics concepts relating to 'health and well-being' footwear (Figure 1.2).

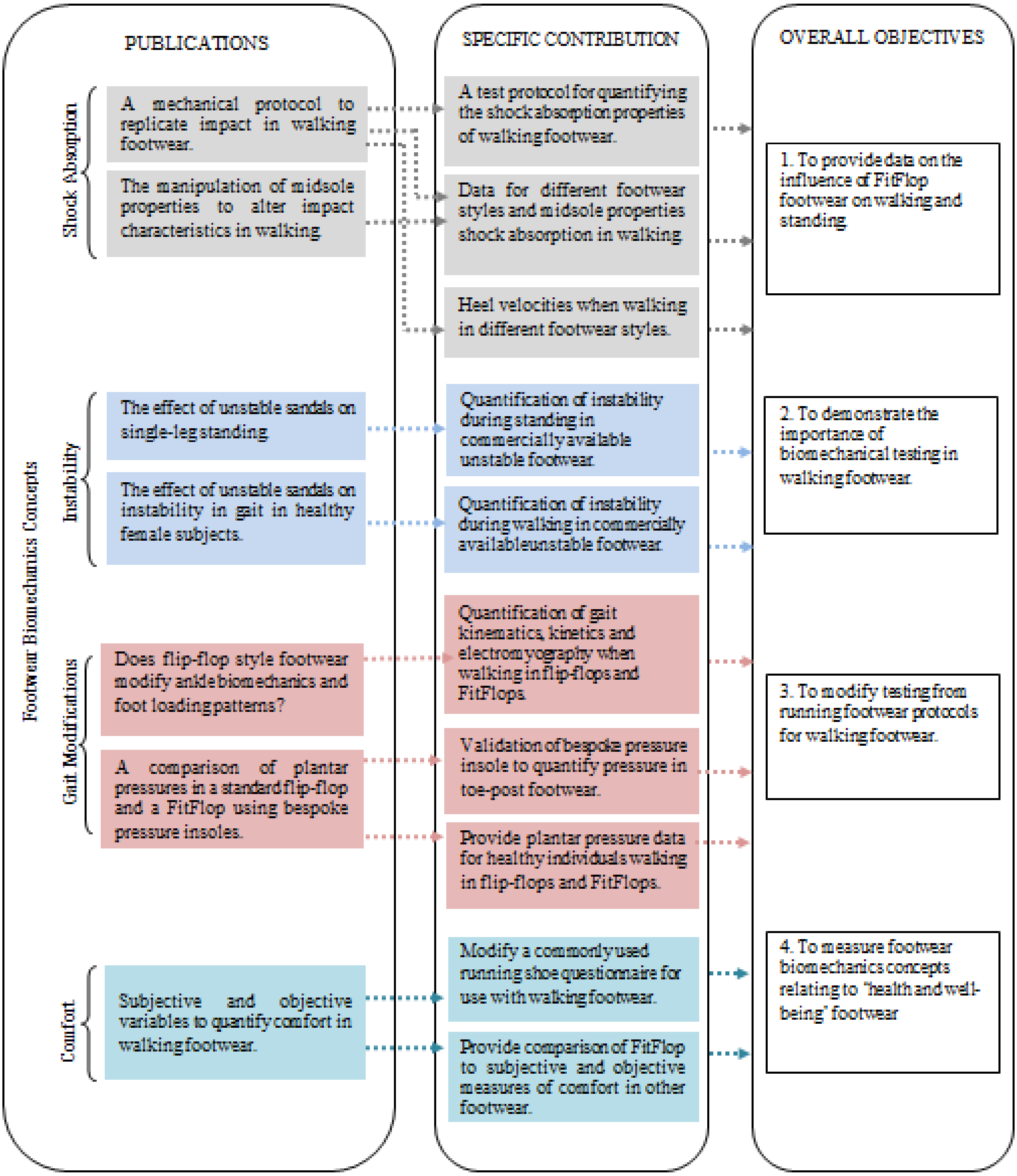


Figure 1.2 Objectives of the thesis

1.4 Thesis Structure

The remainder, and main body, of this thesis has been divided into the three following parts (Figure 1.3).

1.4.1 Chapter 2: Footwear Biomechanics Concept

Chapter two defines footwear biomechanics concepts which are to be investigated within ‘health and well-being’ footwear relating to “Shock Absorption”, “Instability”, “Gait Modifications” and “Comfort”. The main research in empirical literature relating to these areas is discussed, including the methodologies, and findings, relating to footwear. Key points are then drawn from the omissions and insufficiencies of, or extensions to, the existing literature. Additionally, factors relating to embedding the academic knowledge into processes of footwear development and testing are included.

1.4.2 Chapter 3: Publications

Chapter three consists of the publications included within this submission in the format in which they were accepted (or submitted) to the peer-reviewed journals. Additionally presented is a description of the specific contribution from the author to each of the publications including review of the literature and establishing the research questions, study design, data collection, data processing, statistical analysis and paper writing and peer-review response.

1.4.3 Chapter 4: Critique

Chapter four of the thesis critiques the work presented in the publications. Subsections critically appraise aspects relating to the research design then specific methodological choices within each experimental design. Finally, the findings of the research are discussed and conclusions for the body of work made. The contribution that the literature has made to the wider field of footwear biomechanics is highlighted in addition to consideration of the novelty and contribution of each publication. Continuation of the research is reviewed with recommendations for future research throughout the critique.

Figure 1.3 Structure of the thesis

Chapter 2 Footwear Biomechanics Concepts

2.1 Introduction and Definition of Footwear Biomechanics Concepts

The ‘health and well-being’ footwear category is increasing and evident by over 25 companies producing unstable footwear alone (Nigg et al., 2012). The category enables a compromise on traditional footwear design to meet directed outcomes, which aim to provide benefits to the wearers. Provided the potential advantages can be demonstrated and conveyed to wearers there appears to be an acceptance and a willingness to wear footwear that does not meet traditional styles or functions e.g. Masai Barefoot Technology™. This willingness of the wearer to compromise on traditional footwear aesthetics and technology provides the opportunity for the footwear designer and technologist to manipulate specific features of the footwear to meet specific customer demands. Despite this opportunity, this field currently lacks a thorough exploration of the biomechanical influence of such footwear on wearers. Relevant footwear biomechanics concepts are “Shock Absorption”, “Instability”, “Gait Modifications” and “Comfort”, which should be thoroughly explored within this footwear category to quantify the influence of this footwear on wearers.

2.2 “Shock Absorption”

Shock absorption has traditionally been perceived as a beneficial property of running footwear in order to protect the wearer from loading at initial contact and reduce injury risk. The validity of this premise and relevance to walking footwear is to be established. This literature review was undertaken in July 2010 with the aim of defining the literature pertaining to shock absorption in footwear. This aim enabled the current literature relating to the principle of absorbing shock in footwear to be discussed and reviewed.

2.2.1 Introduction

The footbed of the shoe is a site between the heel and floor where the impacts from collisions with the ground at touchdown can be attenuated. This impact may be attenuated by altering properties of the footwear to increase shock absorption. Aspects of running shoes that have been altered to adapt the heel-strike impact include heel flare (Frederick et al., 1984), footbed longitudinal and torsional stiffness (Park et al., 2007), footbed material properties (Gillespie and Dickey, 2003) and foot motion (Perry and Lafortune, 1995). The thickness and hardness of midsoles has been studied, particularly Ethylene Vinyl Acetate (EVA) midsoles in running footwear (Hamill et al., 2011; Milani et al., 1997; Nigg et al., 1987).

Running has become increasingly popular as a recreational activity and the footwear is designed to attenuate high impact forces, therefore research has primarily focused on quantifying loading characteristics in running due to increased forces and injury potential compared to walking (Hamill et al., 2011; Nigg et al., 1987). Impact forces and tibial acceleration in running have been associated with the development of musculo-skeletal injuries; therefore traditionally the shock absorption properties of running footwear have been investigated and enhanced with the expectation and hypothesis that developments could reduce injury rates in runners (Milner et al., 2006; Nigg et al., 1984; Pohl et al., 2008). Research considering military footwear also establishes the value of shock absorbing insoles and footwear to reduce rates of injuries such as metatarsal fractures (Milgrom et al., 1992; Rome et al., 1996). However, quantifying forces experienced at impact in walking is more relevant for clinical groups and the general population as walking is a daily activity while running is less common. A recent survey suggests only 2.0 million adults in the UK take part in athletic activities (including running and jogging) for at least 30 minutes at least once a week (Sport England, 2012). This is a small proportion of the 50 million adults in the country (Office for National Statistics, 2010), most of whom would require to be active and mobile in their daily lives through walking. In walking the characteristics of the heel-strike transient has been identified as related to the symptom of lower back pain (Voloshin and Wosk, 1982); with the provision of viscoelastic insoles related to the relief of such pain (Wosk and Voloshin, 1985). Transient forces are also implicated in pathological conditions such as Achilles tendonitis and plantar fasciitis (Collins and Whittle, 1989). Specific studies have identified prolonged walking on hard surfaces to result in significant changes in both cartilage and bone in the knees of sheep (Radin et al., 1982) and higher heel-strike transient and peak tibial acceleration magnitudes in patients with knee osteoarthritis (Radin et al., 1991). Additionally, subjective observations point to more elastic surfaces, which produce lower peak acceleration values in a drop-test, being more comfortable to walk on (Whittle et al., 1994).

The nature of the impact of the foot with the floor is evident in the vertical ground reaction force as a heel-strike transient. This may be 0.5-1.25 times body weight, and last between 5-25ms in walking (Collins and Whittle, 1989; Henriksen et al., 2006; Lafortune and Hennig, 1989; Perry and Lafortune, 1995). In running the heel-strike transient can increase in magnitude to as much as 3 times body weight (Cavanagh and Lafortune, 1980; Munro et al., 1987) (Figure 2.1). Heel-strike transients are also evident with accelerometers mounted on bony sites on the body such as with a bite-bar (Light et al., 1980), at the sacrum (Wosk and Voloshin, 1981) and mounted on the tibia (Light et al., 1980). In walking peaks are quantified as 2-8 g at the tibia. In running peak acceleration values can be as high as 15 g at the tibia (Hennig and Lafortune, 1989). The nature of the reaction forces and resulting shock wave is dependent on gait velocity (Voloshin, 2000) and impact characteristics (Lafortune et al., 1996). The quantification of heel-strike is undertaken primarily with force plates and accelerometers in vivo (Light et al., 1980; Milani et al., 1997). In walking the analysis of the heel-strike transient is more complex, with the inherent lower forces and loading rates. A transient is not always evident or as easily identified in the ground reaction force, particularly if participants walk in footwear that includes shock absorbing material (Figure 2.1).

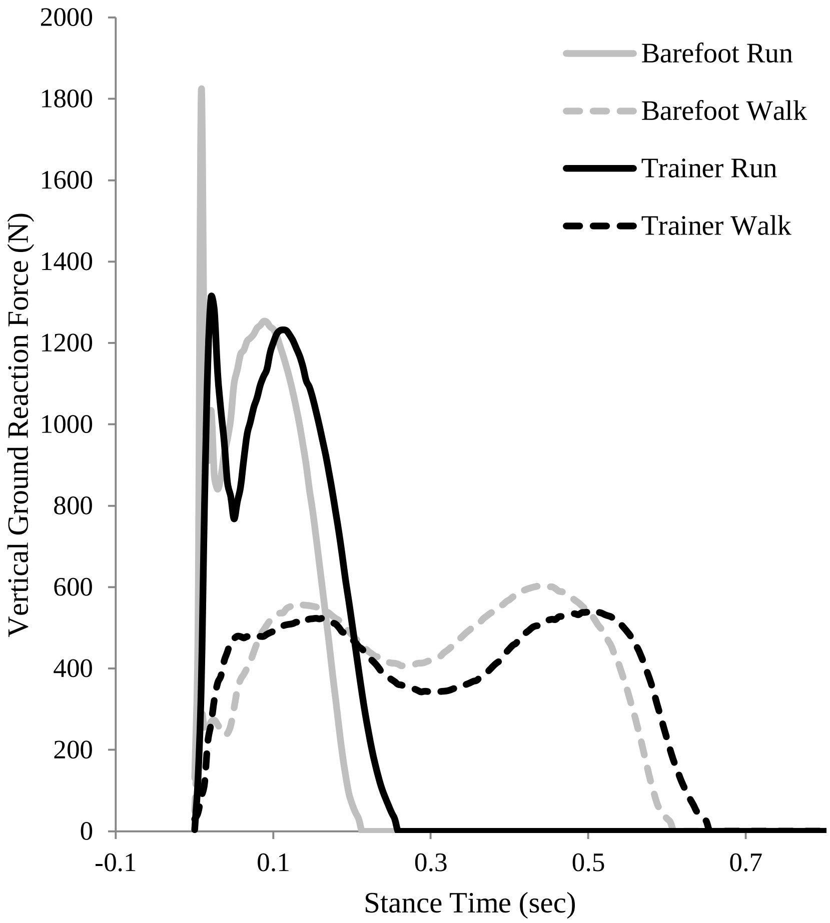


Figure 2.1 Comparison of raw vertical ground reaction force in walking barefoot, walking in a trainer and jogging in a trainer of a 53 kg participant at self-selected velocities.

Additional testing of footwear is undertaken with mechanical impact testing attempting to replicate the characteristics of heel-strike in a controlled environment (Frederick et al., 1984; Hamill and Bensel, 1992). Human testing has the advantage of including the interaction of the human system with the footwear. This quantifies any secondary influence of the footwear such as any effect that the footwear may have on heel pad confinement (Jørgensen and Bojsen-Møller, 1989) and muscle activation (Nigg and Gérin-Lajoie, 2011) as opposed to the footwear as an independent material intervention underneath the heel. Mechanical testing is cheaper and more time effective and does not require a gait laboratory or participants to be undertaken it is therefore more accessible to footwear companies for routine assessments. If a valid replication of the impact in humans can be re-produced with a device then shock absorption of footwear can be quantified quickly and cheaply by a technician in isolation in a manufacturers' office or factory. Both human and mechanical methods will be considered in turn, in addition to some research methods which aim to bridge the gap.

2.2.2 Human Testing

Running footwear has been investigated extensively for shock absorption properties testing participants running in footwear and, more specifically, with differing midsole properties. Peak acceleration values have been recorded by a range of authors for differing hardness or stiffness of footwear midsoles (Hardin et al., 2004; Milani et al., 1997). Milani et al. (1997) identified higher peak tibial acceleration in stiffer footwear; values ranged from 7.58 to 8.49 *g*, although were not linearly related to stiffness of the footwear. Consistent with these findings, Hardin and Hamill (1998) analysed an interaction between footwear midsole and surface hardness when running in Soft (40 Shore A) and Hard (70 Shore A) shoe conditions on three different surface hardness at a running velocity of 3.4 $m \cdot s^{-1}$. Peak impact frequencies on the hard surface and hard midsole combinations were five times the magnitude of those experienced in the soft combinations.

Comparison of ground reaction force data parameters does not reflect the same conclusions as accelerometer data from research regarding the influence of midsole hardness. Nigg et al. (1987) compared maximum impact force and time of occurrence, maximum loading rate to maximum impact force and time of occurrence and shank and rearfoot angle as participants ran in three midsole hardnesses (25, 35, 45 Shore A) at three running velocities. The results indicated a non-significant 10% decrease in maximum impact force with increased hardness.

The time of the impact force peak and time of the maximum loading rate did not differ between 25 and 35 Shore A. However, these decreased for the 45 Shore A condition. A change in foot motion was evident in the harder midsoles, which was proposed as an explanation of the lack of difference in maximum forces between the harder and softer conditions. There was a more lateral impact in the harder midsole, creating a larger pronation velocity (25 Shore A $9.7°\cdot sec^{-1}$, 45 Shore A $22.2°\cdot sec^{-1}$). Therefore the initial movement pattern decreases force as the hardness increases, which was justified and explained as a modification strategy to keep external impact forces constant. This study highlights the requirement for the concurrent collection of kinematic and kinetic data in impact quantification at heel-strike to explain any changes in kinematics affecting recorded peak acceleration or force variables. Similarly, Hennig et al. (1996) identified adaptive changes in loading in harder footwear through differences in impulse related pressure variables; alluding to reduced loading times in the hardest running shoe studied. The results from a Likert scale also identified that runners can perceive differences between hard (3.3±0.7) and soft (10.9±3.4) shoes. Reasons for this contrasting conclusion from force plate and accelerometer measures relate to the variable being quantified. Measurement of the ground reaction force quantifies the load acting on the whole body, not just the heel impact (Mientjes and Shorten, 2011). However tibial accelerometer measures are specific to the tibia and consequently may be more sensitive to interventions applied to the foot such as footwear.

The focus of the majority of footwear testing is running footwear, however other footwear styles have been compared for their influence on shock in gait, for example high heels (Voloshin and Loy, 1994), crepe soled shoes (Lafortune and Hennig, 1992), leather soled shoes (Light et al., 1980) and footwear insoles (Perry and Lafortune, 1995). Some studies do not quantify the footwear characteristics, such as thickness of heel section, as a result making comparison between styles and any implications to footwear design difficult (Lafortune and Hennig, 1992; Light et al., 1980). These studies however remain relevant for practitioners who need to relate research on shock to commercially available footwear that has not been quantified for stiffness or hardness of material composition prior to testing. Light et al. (1980) compared barefoot to shod walking in a shoe with a Sorbothane insert positioned in the sole section cut into the heel and to a crepe soled shoe and a hard leather shoe. Results were reported graphically and identified higher peak accelerations in the hard leather and barefoot conditions and reduced magnitudes and longer loading times in the crepe soled and adapted heel-section footwear. Lafortune and Hennig (1992) studied a similar range of shoes, including an athletic shoe in their protocol. A more robust and quantifiable analysis of data was undertaken and force plate data was concurrently collected. The athletic shoes reduced tibial acceleration peaks significantly from 4.68 g in barefoot to 2.52 g. Force loading rates concurrently reduced from 177.8 $BW{\cdot}s^{-1}$ in barefoot to 35.2 $BW{\cdot}s^{-1}$ in street shoes and 31.1 $BW{\cdot}s^{-1}$ in athletic shoes. It would be expected that as a viscoelastic material (in the form of an insole or shoe midsole) is placed beneath the foot, shock is attenuated. Despite the difficulty of generalising results to footwear, similar to the running footwear research, these studies allude to thicker, softer footwear reducing shock at impact in walking. These studies highlight the importance of quantifying loading in walking for the general footwear user in order to modify walking footwear to increase shock absorption and be more comfortable. However inherent weaknesses mean that further research is required to define and replicate the nature of the impact with the floor when walking in different footwear styles.

In addition to research relating to footwear, various studies the effects of insole design considering shock absorption in walking have been undertaken (Healy et al., 2010). Suitable materials for footwear insole have been assessed for their ability to absorb shock in human testing (Pratt et al., 1986). The insoles were made of Spenco, Sorbothane, Poron and Viscolas and were compared to medium density Plastazote (45 $kg{\cdot}m^{-3}$) which was traditionally being used by this group as an insole material. The thicknesses of the insoles were between 5 and 6 mm and they were assessed during walking using an accelerometer attached to a bite-bar and a force plate integrated into the floor. The results from the gait assessment indicate that values were relatively low when compared to other literature (between 0.75 and 1.08 g) and all insoles reduced impact shock when compared to the footwear only condition. The Viscolas material recorded the lowest acceleration values and force values in a test where a ball bearing was dropped onto the material situated on a force plate (Pratt et al., 1986). The protocol did not report any control of walking velocity and utilised ensemble average data for comparison and as a result conclusions must be interpreted with caution. Similarly, Voloshin and Loy (1994) studied the influence of an insole intervention in high-heeled walking to reduce the impact shock at heel-strike in this footwear style. A lightweight accelerometer was mounted above the tibial tuberosity of three female subjects. The insole intervention was a viscoelastic elastomer with a Shore A hardness of 29 and a thickness of 4 mm in the heel. The inclusion of an insole reduced peak acceleration values by 29% on average in the heeled conditions, suggesting that if heels are to be worn a simple insole offers a worthwhile protection. The study was limited by the lack of kinematic data collection and further analysis of the acceleration signals to look at loading rates or further variables. The studies highlight that footwear can play a protective role at impact with the floor and that different types of footwear (with different constructions) can alter loading magnitudes and rates. The influence of this study on footwear manufacture and design appears to be limited, despite this being a simple solution to potentially improve the comfort of wearing high heels it is not evident in high-street footwear. It is essential that future biomechanics research in walking footwear impacts directly on walking footwear design and what is available to the wearer. This process is evident in running where a substantial volume of the research undertaken is within, or commissioned by, footwear companies themselves to directly drive their design process.

The testing of shock absorption properties of walking footwear and insoles would demonstrate greater external validity and be more accessible to footwear companies if a mechanical test protocol was available which replicated impact situations in walking footwear. Fully quantifying the kinematics of walking in these studies and footwear styles would enable a footwear testing protocol to be developed that is specific to walking gait. This would enable the design process to be populated with relevant shock absorption data and marketing to end users/wearers more worthwhile.

2.2.3 Mechanical Impact Testing

Mechanical impact testing aims to replicate the heel-strike of a human foot on the ground. This has the inherent advantages of being controlled, repeatable, quick and less variable than human subjects. Generally the materials studied in mechanical testing of footwear have viscoelastic and non-linear properties (e.g. EVA) so their rigidity and ability to dissipate energy (so their shock absorbing capability) is dependent on the frequency and magnitude of loading. This means that mechanical test conditions must mirror the characteristics the material will experience in vivo in the footwear if the real-loads on a human are to be compared (Schwanitz et al., 2010). Consideration must therefore be given to the:

- Energy of the impact,
- Direction of and location of the impact,
- Area, mass and shape of missile,
- Repetition rate: material recovery will vary on cyclic loading rates consequently loading and recovery times should both match stride patterns they aim to replicate,
- Footwear construction: if the material is to be bonded to footwear in a specific manner and coated etc. then this fixation and building into actual footwear will affect the material properties and surfaces. The footwear thickness, sole geometry, outsole grid

and other cushioning properties will affect the finite performance of the footwear at impact.

Standardised protocols are utilised in the footwear industry within research and development processes and to test footwear prior to sale for safety. Shock absorption properties of footwear are quantified using the ASTM (American Society for Testing and Materials) F1614-06 standard to test impact response properties of athletic shoes using an impact tester and the SATRA PM 142 Shock Absorption Test Method (Figure 2.2). The ASTM test involves a fixed anvil which is a flat tub with rounded edges weighing 8.5 kg. The SATRA protocol differs slightly with the 8.5 kg mass including a 200 g detachable spherical tub (45 mm diameter). In both methodologies the mass is dropped from a height of 50 mm in order to replicate the 5J of energy at impact with the ground for a man running (Cavanagh et al., 1984). Variables recorded in theses methodologies are:

- Maximum deceleration of the mass on impacting the sample,
- Energy return after impact from the height of the mass,
- The maximum dynamic compression of the sample (penetration).

Figure 2.2 SATRA STM 479 Dynamic shock absorption test machine

Drop tests of varying methodologies are well documented in footwear assessment where a projectile of known mass is released from a given height (Frederick et al., 1984; Pratt et al., 1986). This has the advantages of being the simplest method to quantify shock absorption and requires no additional equipment than would be found in a basic lab (a force plate). These studies, however, over-simplify the impact and may not produce results which can be replicated in vivo. Pratt et al. (1986), for example, dropped a ball bearing onto insoles, a missile, which did not replicate the shape, area, or mass of the human heel.

Frederick et al. (1984) utilised mechanical methods to quantify differences in hardness, heel heights and angles of midsole flare on impact. The impacter was a 7.3 kg cylinder with a force and velocity transducer and a loading rate that was too high compared to human gait impacts. The three hardness values utilised were 25, 35 and 45 Shore A with 10 and 30 mm midsoles and the variable tested was peak acceleration. Peak acceleration values in the testing ranged from approximately 10 g (30 mm thickness/25 Shore A hardness) to 21 g (10 mm thickness/45 Shore A hardness). The results identified that in the 10 mm thickness footbed the 35 Shore A hardness was 27% higher than the 25 and the 45 Shore A 38% higher than the 35 Shore A. The recommendations from the study was that a thicker soled shoe should be chosen ahead of a soft sole as this has no effect on maximum pronation and can increase cushioning. The study was not, however, supported with kinematic data collection, thus the application to pronation is postulated. Milani et al. (1997) identified similar peak acceleration values using the ASTM methodology in footwear with modified heel sections (range 9.60-13.66 g). Mechanical testing protocols have been implemented to compare peak impact in different footwear styles. Hamill and Bensel (1992) compared peak impact in mechanical testing of different footwear styles used by the U.S. military (combat boots, trainers and hiking shoes). The protocol was able to differentiate between footwear conditions as the combat boot recorded peak accelerations of 29.8±1.21 g and the Nike Air recorded lower values of 14.3±0.88 g. Similarly, Hennig et al. (1993) compared 19 different sports shoe constructions, top model and highest price point shoes from the highest selling footwear retailers. The peak acceleration range also demonstrated differences across footwear conditions and was ≈10.9-15.4 g.

Inconsistencies between mechanical methods and the human data, which they aim to replicate, are evident by the variation in reported correlations between human and machine results. Pratt et al. (1986) demonstrated good agreement between dropping a ball bearing on the insole and human testing. The method did not produce comparable values; however ranking the insoles in order of effectiveness produced the same results from both testing procedures. The study utilised only one subject and therefore results may be more similar than if a range of participants were tested. Shiba et al. (1995) identified an 11% difference in material properties in humans and in a laboratory test involving dropping a golf ball onto the material sitting on a force plate. Hennig et al. (1993) utilised an impact tester (Exeter Research Inc.) to characterise the material properties of heel sections of different shoe soles. The mechanical test results were compared to peak tibial accelerations during running, and only a low non-significant correlation was identified ($r = .26$, $p > .05$). The impacter scores suggested large differences in stiffness between the shoes (11-14 g), however the material properties had only small influences on the shock attenuation behaviour at initial-contact in running. The authors attributed this to an unrealistic simulation of the foot-ground contact, likely due to kinematic changes at heel-strike. This is consistent with other authors' findings reporting alterations in eversion at initial contact in running shoes with different structures (Nigg et al., 1987). The accuracy/validity required from mechanical tests depends on how data will be integrated into the process. For some uses simply being able to rank materials effect may be enough, for others estimated absolute values of shock absorption may be required.

Various authors have compared mechanical testing protocols, with consistency between protocols being dependent on the materials tested (Schwanitz et al., 2010). Chi and Schmitt (2005) proposed a variation to standard protocols with the moderation of the mass and/or height of the mechanical impact situation in order to manipulate the impact energy to better simulate varied impacts in running. Testing of insoles with a given impacter dropped from a range of heights allowed the calculation of energy return for six impacts of varying energy (2 to 6 Joules) with the aim of replicating variations in impact between and within participants/wearers. Thus, impact velocities have varied in the literature, with vertical velocities of approximately 1 $m{\cdot}s^{-1}$ used for running shoe testing (Cavanagh et al., 1984; Frederick et al., 1984) and 1.4 $m{\cdot}s^{-1}$ for quantifying the shock absorption properties of the heel pad (Jørgensen and Bojsen-Møller, 1989). Concurrently, the latter authors utilised a dropped mass of 1.6 kg from his height to produce a reportedly similar force and collision time to gait (Jørgensen and Bojsen-Møller, 1989). This modification of overall impact energy is necessary to ensure that the energy of the impact represents that which may be experienced when wearing the shoe for its intended use.

Currently the standard mechanical testing protocol replicates an adult male running. This test method is utilised as the testing standard for all types of footwear by SATRA and is reported as standard when authors are defining footwear in methods of papers, in studies relating to military footwear (Hamill and Bensel, 1992; Windle et al., 1999), walking footwear (Allen, n.d.; Silva et al., 2009) and tennis footwear (Morey-Klapsing et al., 1997). The relevance of a standard protocol is undetermined, as the energy of the impact, the mass and shape of the missile and the direction of the impact may not closely replicate that evident in the wearer of the footwear. For example, walking footwear is exposed to substantially different impact energies at slower rates than running footwear, and footwear for children is contacted with a calcaneus of smaller area than footwear for adults. These specific features and characteristics of the nature of the impact of the heel and footwear with the ground mean that wearer, activity and footwear style specific protocols are required to effectively and validly test the broad array of footwear that is currently commercially available.

2.2.4 Alternative Methods

Other test methods attempt to test on human participants, but in a more controlled environment than running in the laboratory For example, impact testing using a pendulum method was used by Aerts and De Clercq (1993) to quantify the effect of varying midsole hardness on the heel region of the foot. They proposed that midsole hardness influences the confinement and loading rate of the heel pad and consequently the inherent effectiveness to attenuate shock. The footwear (trainers with midsoles EVA 65 Asker C and EVA 40 Asker C) was loaded at impact velocities ranging from 0.37-1.06 $m \cdot s^{-1}$ in a pendulum system. Maximal force increased with input energy and was higher in the hard than the soft shoe, this pattern was mirrored by the loading rate, but differences were higher between shoes. A ballistic pendulum method has been utilised by Lafortune and Lake (1995) to quantify both tibial acceleration and discomfort associated with the impact. The adjustment of impact velocities demonstrated by Aerts and De Clercq (1993) highlights that some authors have considered the relevance of higher impact velocities to footwear testing, however the widespread adaptation of impact velocity for walking footwear testing is yet to emerge.

Nishiwaki (2003) devised a drop jump protocol onto foam sheets to attempt to replicate running impacts to test running footwear without having to manufacture midsoles. He tested a range of different thickness and hardness EVA with one subject and a tibial mounted accelerometer. SRIS-C hardness 40 foam at 15 mm thick produced similar accelerations as 70 hardness and 30 mm thickness, despite the subjects' perceptions being different and the 40-15mm combination being preferred. These methods represent attempts to replicate running impacts in environments where more variables are controlled. Ensuring that loading magnitudes and rates are valid, these methods represent time and money saving opportunities for footwear testing in vivo. They are advantageous when compared to mechanical testing as participant-specific mass is present for the impact as opposed to selecting a generalised mass to impact the footwear and represent the population. Manipulation of such methods to match walking impacts would be feasible by reducing the drop height of the participant onto the material.

2.2.5 Literature Summary

Due to the increase in the footwear market encompassing 'health and well-being' footwear, the development of methodologies to replicate the characteristics of walking is essential to quantify the shock absorption properties. Currently running footwear protocols are used to test all footwear styles in commercial and research situations. Wearers expect 'health and well-being' footwear to absorb shock; however no data currently exists to quantify this, which necessitates realistic and valid testing. Running methods over-apply energy at impact and as a result may overestimate the material volume or density required for effective shock absorption in walking footwear. This may unnecessarily lead to increased cost for manufacture and distribution. In order to define an appropriate drop-test, the characteristics of walking need to be defined, such as the effective mass of the lower limb, the limb configuration and velocity and the footwear style.

2.2.6 "Shock Absorption": Key Points

- 'Health and well-being' footwear may have the potential to absorb shock in walking and methods exist to quantify this biomechanical concept for research and development purposes.
- Current methods are mechanical or human in nature. Human testing of shock absorption has the advantage that it includes any interaction effect that the footwear has on gait, however mechanical testing can be completed more quickly and cheaply
- The current widely used mechanical method is designed to replicate the impact evident in running and as a result the magnitude of the impact is too high to mimic walking. Therefore mechanical methods to replicate impact between the shoe and floor in walking are required.
- Modified mechanical protocols would allow valid, generalizable and quick testing of walking footwear.

- The ‘health and well-being’ footwear category thus provides the opportunity to produce appropriate thickness and hardness of footwear to cost-effectively modify shock at initial contact in walking.

2.3 “Instability”

Some ‘health and well-being’ footwear is designed to induce instability in the wearer during walking and standing. The research backing of the footwear styles in the category of unstable footwear varies greatly with varying volumes and quality of research data investigating this biomechanical concept. Some footwear companies and styles have substantial research and others have no published research support evident in peer-reviewed literature at this time demonstrating reduced stability in the footwear (August 2010). This literature review aims to summarise existing literature on unstable ‘health and well-being’ footwear and methods evident in literature to quantify instability which could be applied to quantify the biomechanical effect of such footwear.

2.3.1 Introduction

Instability or “toning” footwear has been a fashionable trend with commercial sales increasing and many longstanding leisure footwear companies taking up the challenge of developing and marketing a shoe claiming to increase muscle activation (e.g. New Balance^{TM}, Skechers^{TM}, Reebok^{TM}; Figure 2.3). As the footwear category continues to grow, the emphasis must move from developing the products with ideology to validating the products with biomechanical and physiological analysis to demonstrate this instability and therefore function. In wider literature, instability during standing and walking is typically quantified via postural sway analysis, and more often is undertaken on vulnerable populations such as the elderly (Hijmans et al., 2007; Rugelj and Sevšek, 2007). There are a plethora of variables to quantify postural sway reported in research literature, including centre of pressure, electromyography and kinematic measures (Murray et al., 1975; Raymakers et al., 2005).

2.3.2 Methodologies in Instability Assessment

Postural sway testing is undertaken to quantify instability during stance in gait and to assess the interaction of effects to improve stability such as footwear (Ramstrand et al., 2010) and fatigue (Suponitsky et al., 2008) on measured variables. Variables to quantify postural sway vary and those previously implemented in the vast literature base are later discussed. Examples of centre of pressure and electromyography variables used to denote stability are defined in Tables 2.1 and 2.2 respectively. Literature focuses on these variables for quantification of instability, while other variables are less commonly reported within the footwear domain and are later briefly discussed.

Figure 2.3 Instability footwear examples

a) New Balance True Balance™, b) Reebok EasyTone™, c) Masai Barefoot Technology Kesho™, d) FitFlop Pietra™, e) Skechers Tone-Ups™ and f) Marks and Spencer Step-Tone™.

Table 2.1 Centre of pressure variables from protocols to quantify instability.

CoP Variable/ Terminology	Description	Static or dynamic	References
AP range	Absolute value of the path covered by the CoP: Max AP value – Min AP value.	S	Hasan et al., 1990; Landry et al., 2010; Murray et al., 1975; Raymakers et al., 2005.
ML range	Absolute value of the path covered by the CoP: Max ML value – Min ML value.	S	Hasan et al., 1990; Landry et al., 2010; Murray et al., 1975; Raymakers et al., 2005.
CoP range as % of foot length	Calculated path as % of foot length in ML and AP directions	D	Schmid et al., 2005.
Total path length/ mean distance/ total excursion	Total length of the CoP path, summation of consecutive points on CoP path in mms. Calculated for AP and ML separately or combined.	S	Prieto et al., 1996; Murray et al., 1975; Heller et al., 2009.
Planar deviation (mm)	The square root of the sum of the variances of displacements in AP and ML direction.	S	Raymakers et al., 2005.
Ellipse area (mm^2)	Fitting an ellipse to the area covered by CoP motion.	S	Davis et al., 2008; Heller et al., 2009; Hasan et al., 1990.
Sway area	Estimates the area enclosed by the CoP path per unit of time. Approximated by summing the area of triangles formed by 2	S	Prieto et al., 1996; Davis et al., 2008.

	consecutive points of CoP path and the mean CoP position.		
95% confidence circle area	Model of the area circle including approximately 95% of the distances from the mean CoP point assuming distances are normally distributed.	S	Prieto et al., 1996.
Instantaneous velocity of CoP	First derivative of displacement data.	D	Schmid et al., 2005.
Velocity CoP / mean velocity / mean sway velocity	Velocity of CoP path AP and ML.	S and D	Heller et al., 2009; Raymakers et al., 2005. Schmid et al., 2005;
95% power frequency (Hz)	Frequency domain measure taken in the ML and AP directions separately. Calculated from resultant distances from the mean CoP point.	S	Prieto et al., 1996; Santos et al. 2008.
Lateral-medial area index/ CoP index	Ratio between the area lateral of the CoP path and the area medial of CoP path: [(lateral area – medial area) / (lateral area + medial area)] × 100	D	Scherer and Sobiesk, 1994; Cornwall and McPoil, 2003.
Lateral-medial force index	Ratio between the force lateral to and the force medial to the CoP line: [(lateral force – medial force) / (lateral force + medial force)] x 100	D	Cornwall and McPoil, 2003.

Where S= static, D= dynamic, ML= medial-lateral, AP= anterior-posterior, CoP= centre of pressure.

As evident, various centre of pressure variables have been used in previous literature to quantify dynamic and static stability characteristics of subjects. The centre of pressure variables utilised in research quantify the magnitude, duration or frequency of motion. Variables that separate medial-lateral and anterior-posterior sway, such as anterior-posterior range (Landry et al., 2010), may better allow the features of the footwear to be assessed and compared in walking and standing activities. Some variables assess just ranges, allowing large differences to be based on one large deviation from the average point, other variables combine temporal and distance measures from the average parameters, describing not only how far the participant deviated from their mean, and the duration for which they deviated. This may be more worthwhile when attempting to compare footwear styles. Numerous variables utilise subtle calculation differences to present information. For example, papers quantify the medial-lateral positioning of the centre of pressure by calculating the area under the medial-lateral curve, a measure very similar to the more commonly reported centre of pressure index (Cornwall and McPoil, 2003). Treatment of data differs with some variables being normalised to a subject feature in studies. For example, centre of pressure or centre of mass trajectory may be normalised to leg length to compare between subjects. Normalisation of variables is not required in the present work; comparisons between footwear will utilise subjects as their own control (Schmid et al., 2005). It is clear that a combination of measures will give the clearest picture of the instability being imposed on a subject by footwear interventions in static and dynamic situations, with variables that can be directly attributed to specific influencing footwear design features most noteworthy.

The collection of electromyography data in footwear research tends to focus on the muscles of the lower limb that are related to the main sagittal plane motions of gait, for example the hamstrings, quadriceps and calf muscle groups (e.g. Roy and Stefanyshyn, 2006). The tibialis anterior and peroneus longus activity are also commonly quantified (Romkes et al., 2006; Suponitsky et al., 2008). Quantification of the muscle activity of the more distal muscles is more complex and isolating specific signals from muscles requires modifications to general methodologies. A circumferential linear array of electromyography sensors has been wrapped around the shank to quantify activity of the flexor digitorum longus, soleus, peroneus longus and anterior compartment group (Landry et al., 2010). This protocol enabled the quantification of wavelet intensities from the muscles (Table 2.2), which is not achievable without being able to isolate signals from specific muscles, although this allocation of signals is estimated. The method, however requires an electromyography system with numerous channels to position electrodes around the limb, which may not be available to all researchers. Similarly, intramuscular electrodes have been utilised to quantify activation of deeper muscles such as peroneus longus and tibialis posterior in walking (Murley et al., 2010). This technique is more time consuming and may not be available to all researchers due to the ethical and practical constraints.

Table 2.2 Electromyography variables from protocols to quantify instability.

EMG Variable	Description	Static (S) or dynamic (D) testing	References
RMS	Root mean square of surface EMG.	S and D	Suponitsky et al., 2008.
EMG Average rectified value	Average rectified value: dividing EMG integral by the time interval it was recorded over.	S	Hatton et al., 2009.
EMG wavelet intensity	Total intensity was calculated as the sum of EMG intensities attributable to each muscle.	S and D	Nigg et al., 2006b.
Duration	Time that muscle is considered active, defined by activation level being over a specific threshold.	D	Tomaro and Burdett, 1993; Li and Hong, 2007.
IEMG	Integrated EMG- the integral of the linear envelope of EMG sample over a specified time period.	S	Yanagiya and Koyama, 2009.
Mean phases	Mean muscle activity during phases of stance.	D	Schmitz et al., 2009.

Where EMG = electromyography

Other variables pertaining to stability in walking footwear include temporal-spatial and kinematic measures. Increased stance width, double-support time and step width have been used as markers of reduced stability, in addition to a reduced walking velocity (Lemaire et al., 2006; Menant et al., 2008). Additionally, the attenuation of kinematic variability and bi-lateral asymmetry are utilised to estimate stability (England and Granata, 2007; Hurmuzlu et al., 1996). Head, trunk and pelvis acceleration measures are quantified in clinical situations to measure stability of specific patients, such as adults who have diabetes (Menz et al., 2004). MacKinnon and Winter (1993) highlighted specific frontal plane indicators of stability such as pelvis motion and moments acting about the subtalar joint. These variables have been considered in existing research literature alongside the aforementioned electromyography and centre of pressure variables to indicate instability or strategies to maintain or improve stability in walking and standing.

2.3.3 Literature Overview

The research backing of the variety of footwear designs and technologies in the category of unstable footwear varies greatly. Some styles have substantial research categorising wearer's kinematics and kinetics in the footwear and others have no published research support evident in peer-reviewed literature at this time (August, 2010).

Masai Barefoot Technology (MBT™) footwear has thoroughly documented research support, undertaken by both independent and associated or funded researchers. This validates an increase in muscle activity, primarily in the tibialis anterior in swing and gastrocnemius during stance (Nigg et al., 2006b; Romkes et al., 2006). Increased tibialis anterior activation in swing is likely due to the increased mass of the shoe. However, authors do not habitually report the features of the control shoe acting as a comparator, such as mass, hence this cannot be confirmed. Control shoes are generally trainers (Nigg et al., 2006b) or street shoes (Romkes et al., 2006), which likely have a lower mass than the MBT™ shoe due to the thick rubber sole. Nigg et al. (2006) report the mass of their Adidas SuperNova control shoe to be 358g compared to the 650g reported for the MBT™ test shoe utilised.

Nigg et al. (2006) proposed the mechanism by which MBT™ functions, being that the shoe strengthens the smaller lower limb musculature with insertions closer to the axis of rotation. Activation of these muscles therefore reduces joint loading compared to that of larger muscles which insert further from the axis and this potentially reduces joint pain in wearers. Testing this hypothesis, Landry et al. (2010) utilised a "circumferential linear array" of EMG electrodes positioned with magnetic resonance imaging to study the smaller muscles of the lower limb during bi-lateral standing and walking pre- and post- a 6 week intervention. Pre-intervention the centre of pressure ranges in the anterior-posterior ($p < .001$) and medial-lateral ($p < .001$) directions were significantly greater in MBT™ (anterior-posterior 26.24±10.31 mm, medial-lateral 11.36±9.67 mm) than the control (anterior-posterior 44.69±17.33 mm, medial-lateral 18.78±15.09 mm). This was concurrent with significantly increased muscle wavelet intensity in the flexor digitorum longus and the anterior compartment muscles compared to the control shoe. This mirrors findings by Nigg et al (2006) in standing where there was a trend for EMG to increase across the muscles tested by 11-70% in an MBT™ condition compared to a trainer. The highest differences between MBT™ and control were evident in the tibialis anterior (70±85%, $p < .05$) and gastrocnemius (37±46%, $p > .05$). This was concurrent with an increase in centre of pressure range in the anterior-posterior (17%) and medial-lateral (6%) directions. The studies comparing standing in MBT™ consider muscular control in a more static environment than walking, but also one that is valid and relevant to a wearer in daily situations. The large anterior-posterior rocker sole and collapsing heel "element" of the MBT™ shoe are potentially instrumental features resulting in the decrease in anterior-posterior and medial-lateral stability of the wearer respectively.

Kinematic adaptations to walking have been assessed in MBT™ footwear and include increased ankle dorsiflexion in the first half of stance compared to a flat trainer (Nigg et al., 2006b; Romkes et al., 2006). Contrasting this, no significant differences were evident in the ankle frontal plane angle (Nigg et al., 2006), despite this previously being identified as a potential variable to denote frontal plane stability challenges in walking (MacKinnon and Winter, 1993). It may be that, due to the large anterior-posterior rocker shoe and minimal medial-lateral design features, the instability induced by MBT™ is greater in the sagittal plane and that frontal plane compensations are less severe. Vernon et al. (2004) report a more upright posture in MBT footwear, with a significant reduction in trunk angle from 5-10° to -4-0° in the 22 participants that they tested, concluding that MBT™ alters gait beneficially by shifting the centre of gravity backward. Furthermore, New and Pearce (2007) found that MBT™ footwear influences trunk flexion and anterior pelvic tilt at heel-strike and toe-off compared to a control shoe. The interpretation and meaning of these findings, in terms of clinical effect, however would require further investigation. The influence of the period of wear, or familiarity with the footwear, is likely to produce differing responses in wearers and this is of interest to quantify the long-term function or efficacy of the footwear. Stöggl et al. (2010) identified that the variability in gait kinetics and kinematics evident at the first time of wearing (+35%) becomes negligible compared to the variability apparent in a standard trainer after a 10 week intervention period. This equal variability was later demonstrated to be a combination of reduced variability in MBT™ in some variables (e.g. peroneus longus muscle activation) after the intervention and an increased variability in the control shoe in some variables (e.g. vertical centre of mass position). Additionally, relevant specifically to MBT™, is how wearers familiarise with the shoe with and without specific training. The footwear is sold by trained retailers only in the UK and supplied with an instructional DVD. The influence of this training on kinematics when wearing the shoe is yet to be quantified.

Plantar pressure assessment has been utilised by some authors to establish the mechanism by which the MBT™ footwear interacts with the foot and to establish whether the footwear might be appropriate for use in clinical groups, such as adults who have diabetes. Kalin and Segesser (2004) tested 15 healthy subjects in an MBT™ shoe compared to "conventional shoes". The study identified that the MBT™ footwear shifts plantar pressures from the heel to the forefoot. Maximum load was reported to reduce in the heel and increase in the metatarsals by 400-500%. Stewart et al., (2007) reported the same pattern, but a lower magnitude with a 76% increase in peak pressure in the forefoot when standing for 30 seconds and 7% in walking. Furthermore, an increase in contact area in standing from 81.7 to 91.4 cm^2 was identified and attributed to the mass displacing to the forefoot and the increase in instability. Stewart et al. (2007) compared the MBT™ to the subjects own flat-soled sports shoe, a comparative shoe in function and one that you would expect to have high contact areas and uniform pressure distribution properties. One issue with the validity of the study is that the protocol did not control walking velocity, which could have enabled participants to walk more slowly in the MBT™ condition (Romkes et al., 2006), consequently reducing plantar pressures independent of footwear design features.

FitFlop™ and Reebok™ both went some way to validating their products with small subject number studies commissioned at Universities, which identified increased muscle activity, however are not all accessible in the public-domain e.g. Gautreau et al. (2009). Reebok EasyTone™ consists of small air pods under the heel and forefoot that compress on weight bearing and aim to promote instability. The company promotes stability training with the footwear highlighting results from unpublished University studies reporting:

-28% more gluteus maximus activation

-11% greater hamstring activation

-11% increased calf activation compared to a foam-based shoe

As air is able to travel between the forefoot and heel in the Reebok EasyTone^{TM} are promoted as three times softer than foam based shoes, however no further explanation is evident. The FitFlop^{TM} shoe sole technology is based on micro-instability and increasing muscle activation to stabilise the body during mid-stance. The footbed is made of multi-density EVA positioned throughout different regions of the sole. Marketing claims are based on both prolonging and increasing the magnitude of muscle activation, particularly in the gastrocnemius muscle, with percentages ranging from 18 to 40%. Studies by Reebok^{TM} and FitFlop^{TM} at Universities report increases in muscle activity in the lower limb when wearing the footwear compared to a control shoe or trainer, although studies are not published in peer-reviewed journals and the protocols are not currently evident in the public domain. Future research studies quantifying instability in these footwear styles with larger subject numbers are therefore warranted. The studies considering MBT^{TM} include relatively large subject numbers and span clinical groups, adults and children (Nigg et al., 2006a; Ramstrand et al., 2010, 2008). The consideration of clinical, older, or symptomatic groups would be beneficial as these may be the realistic wearers for this footwear category and companies or manufacturers may consider this a future research priority.

Skechers^{TM} report results from a range of applied studies looking at increases in the energy expenditure during walking in the footwear as well as muscle activity, however it could be argued that these studies lack experimentally robust methods. For example, results reported by Gautreau et al. (2009) use a control group that is not weight matched in a longitudinal study. The research reports greater weight loss and increased body fat reduction in exercise prescription wearing Skechers Shape-Ups^{TM} footwear than without. A study from Juntendo University identified increased muscle activation when compared to a trainer in walking at various velocities with Skechers^{TM} increasing the integrated electromyography values in the thigh, calf, buttocks and back for most walking velocities tested (Yanagiya and Koyama, 2009). The studies are proposed to support two entirely different footwear constructions (the multi-density flip-flop range and the rocker soled shoe range), thus highlighting a lack of clarity or understanding of the mechanism and function of the product within the marketing. This highlights a requirement to increase knowledge and understanding of biomechanics and the implications of footwear design within footwear companies.

The studies commissioned by these companies all act at as a starting point for research with the standard of peer-reviewed published data being a must for this footwear category going forward to secure support from the scientific community and wearers alike. As is evident in Figure 2.3, a range of technologies incorporated in shoe midsoles are included in the footwear construction and promoted as unstable. Clarifying differences in technologies and effects will correctly direct wearers to the most relevant footwear for them. Genuinely independent studies on fitness or toning shoes are sparse, with a lack of thorough investigation into reported claims and actual function. Comparative studies act to inform a clinician wishing to prescribe specific instability to a specific wearer and some are evident in the literature.

A study by the Ace Fitness Group, associated with the University of Wisconsin, reviewed the claims of MBT™, Reebok EasyTone™ and Skechers Shape Ups™ (the rocker soled version of their fitness shoe) and concluded that the shoes performed no better than a pair of standard trainers (Porcari et al., 2010). The assessment of the footwear was against a normal running shoe and utilised a two stage assessment. Stage one quantified oxygen consumption measures and stage two electromyography variables of six muscles. The oxygen consumption study included rate of perceived exertion measures, and did not identify any significant differences between the shoes. This may have been down to the study being 5 minute tests on a treadmill with controlled walking velocity, it is unlikely that prolonged walking in the MBT™ shoe would not increase perceptions of exertion due to the mass of the shoe being double that of the other footwear. Additionally, if left to walk naturally in the MBT™ shoe some subjects walk slower (Romkes et al., 2006) and this would result in altered oxygen consumption in a ‘real life’ situation if everything else remained constant. The validity of a five minute walking test and the use of a treadmill potentially altering gait changes reduce the external validity of the research. The electromyography assessment quantified activity of the erector spinae, gluteus maximus, biceps femoris, rectus femoris, gastrocnemius and rectus abdominus. The methodology is not described, and the results are presented as a percentage of maximum voluntary contraction. From graphical representation there appear to be increases in electromyography amplitude, particularly with the gastrocnemius in MBT™ and gluteus maximus and biceps femoris in the Reebok EasyTone™. However the results are reported as not significantly different. No measure relating to muscle activation duration is addressed in this study, which is unfortunate as this is a marketing claim and reported finding for many instability shoes as opposed to looking at the magnitude of activation at one instance in time. Using the integral of muscle activation recordings would have captured aspects of the time domain, which is a variable used in previous footwear studies (e.g. Nigg et al., 2006). The external validity of future research studies in unstable footwear can be increased by including more varied participants. Porcari et al. (2010) recruited 12 physically active females (21-27 years), who may not have found the shoes unstable enough to elicit a response and who may not be relevant representatives for users of the footwear in this category.

2.3.4 Literature Summary

Previous research on instability footwear identifies that the footwear does alter gait when compared to a control shoe, however exact relationships and determinants are not clear. The focus of the literature currently is that MBT^{TM} footwear and other styles lack thorough analysis as to their effect on stability in wearers. It is evident from consideration of literature pertaining to unstable footwear that the study outcomes and relevance are highly dependent on methodological choices made during the study design. More specifically, the consideration of the relevance of the participants to be recruited, task they undertake and footwear conditions to be tested impact on the internal and external validity of the research and findings. These variables determine the sensitivity of the protocol at identifying instability in the footwear conditions tested. It is evident from the work undertaken on unstable shoes that thorough protocols with specifically chosen variables should be utilised to quantify the specific effects of these footwear styles on a user and to relate these outcomes back to specific features of the shoes being tested.

2.3.5 "Instability": Key Points

- Variables pertaining to centre of pressure trajectory, electromyography integrals are sensitive to differences in stability in participants and therefore may be sensitive to any biomechanical influences of unstable footwear.
- Unstable footwear has been shown to alter muscle activation and centre of pressure trajectory in both standing and walking.
- The population quantified wearing unstable footwear in research must be relevant to the wearer of the shoe due to the influence of age and activity on stability.
- Tasks must be relevant to 'real life' situations or previously correlated with other measures of stability. The task must be challenging enough to illicit a response in the tested population.
- The current focus of unstable footwear research is MBT^{TM} centred and the investigation of other footwear designs is warranted. Current data directly comparing different unstable

footwear constructions of instability/toning footwear is not available or of high enough quality to determine specific biomechanical consequences of the footwear and thus recommendations or prescriptions.

2.4 “Gait Modifications”

Styles of fashion footwear such as high-heels and flip-flops are often anecdotally reported to be detrimental to a normal gait pattern and result in biomechanical modifications to gait. However there exists minimal extensive biomechanical assessment of these footwear styles of relevant alternatives. The aim of this literature review was to consider the influence of toe-post footwear on gait and discuss the literature that currently exists on this topic (June, 2009).

2.4.1 Introduction

Shoes are implicated for many foot deformities and symptoms, however establishing causality in these relationships is not possible (Kilmartin and Wallace, 1993). Gait modifications to accommodate wearing fashion footwear styles have been reported in footwear including high-heels (Cowley et al., 2009; Lee et al., 2001), flip-flops (Shroyer, 2009) and rocker shoes (Brown et al., 2004; Myers et al., 2006). These changes range from observational contrasts, such as decreased walking velocity (high-heels and flip-flops), to altered contact position at touchdown (rocker-soled footwear) and changes to lower limb joint kinematics (flip-fops, rocker-soled footwear and high-heels) (Albright and Woodhull-Smith, 2009; Cowley et al., 2009; Myers et al., 2006; Shroyer, 2009). Due to these alterations to ‘normal’ biomechanics and resulting symptoms, healthcare professionals generally criticise fashion footwear styles, such as flip-flops and high-heels, and recommend against their regular use (American College of Foot and Ankle Surgeons, 2007; Shroyer, 2010; Thompson and Coughlin, 1994). The modifications to “normal” or “barefoot” gait are generally regarded as being detrimental or negative to a healthy walking gait in the long-term. However minimal longitudinal research into the specific adaptations and subsequent effects exists. Additionally, despite the abundance of criticism of these footwear styles, peer-reviewed and published research denoting specific gait changes is sparse. The footwear holds a place in society and it is now unlikely that the use of the footwear will reduce, for example due to their affordability, convenience and thermal benefits flip-flops will always be commonplace in countries such as India and Australia (D’AoÛt et al., 2009). Thus, once specific gait characteristics and symptoms have been researched and highlighted in the fashion footwear then viable footwear alternatives or design adaptations are required. These alternatives must meet the fashion demand while concurrently making specific and directed design-feature changes to reduce any potentially damaging or detrimental behaviours that were evident in the original fashion style. If a similar shoe (which meets the same purpose) can be provided without the associated detrimental modifications to gait then this poses healthcare benefits for specific wearers and may reduce the appearance of symptoms in the general population. The following review will focus on the influence of toe-post footwear on gait.

2.4.2 Definition of Toe-Post Footwear

Toe-post footwear is defined by having one strap across the dorsal fore-foot which attaches to the footbed between the hallux and lesser-toes (Figure 2.4). The footwear has an open upper and no heel-strap or support to the rearfoot. Standard midsoles have a flat profile with no medial arch, heel cup or similar features.

Figure 2.4 Havaiana™ flip-flop

Localised heel pain and other conditions are specifically implicated for the wearers of flip-flops by podiatrists (American College of Foot and Ankle Surgeons, 2007). Flip-flops defy general recommendations for footwear by having a thin midsole, not supporting the medial arch, by not protecting the toes or the dorsal surface of the foot, by the upper being loose on the foot and being flat with no pitch from heel to toe (Barton et al., 2009; Stewart et al., 2007) (Figure 2.4). It is proposed that these design issues result in adaptations to gait when walking in this footwear style, even in individuals who are otherwise asymptomatic (Shroyer and Weimar, 2010). Additional links with this footwear and clinical conditions relate to simple issues such as blisters and puncture wounds, to strain of the ankle dorsiflexors (Shroyer et al., 2010). However, in contrast, it has been highlighted that flat, flexible footwear may provide some advantages to patients from specific clinical groups, such as those with knee osteoarthritis (Shakoor et al., 2010). Joint loading has been investigated in flip-flops in a knee osteoarthritic population of 31 participants (Shakoor et al., 2010). The results indicated that when walking in the flip-flop, patients had the lowest peak frontal and sagittal plane knee moment of all the footwear and this value did not differ from the barefoot values. This finding was attributed to the flat, flexible footwear and the lack of medial arch on the plantar surface of the shoe (Shakoor et al., 2010).

Despite the general criticisms of flip-flop footwear styles, a thorough search of peer-review published literature identifies limited scientific investigation into the biomechanics of walking in these shoes. The focus of literature in this field is on the effect of footwear on arch development in children's feet, where studies have identified a decrease in prevalence of flat foot in children who wear shoes as opposed to walking barefoot or wearing open toed shoes as opposed to closed (Rao and Joseph, 1992). Rao and Joseph (1992) identified 13.2% of 2300 Indian children who wear shoes had flat feet, compared to 6.0% who wore sandals and 8.2% who wore slippers. Similar to the findings from knee osteoarthritic population, these studies suggest that flip-flops may pose some advantages. It has also been demonstrated that the use of footwear has been linked to valgus deformities in adults and children (Kilmartin and Wallace, 1993; Stewart et al., 2007). Oeffinger et al. (1999) identified the specific biomechanical changes occur in shod compared to barefoot walking in children, which may explain Rao and Joseph (1992) indentifying varying foot structure in children with different footwear habits. Fourteen children were assessed in barefoot and athletic shoes for their gait kinematics and kinetics. Whilst walking in footwear the authors identified a decrease in participants' external foot rotation, a decreased knee flexion in early stance, decreased plantarflexion throughout and an increase in stride length compared to barefoot (Oeffinger et al., 1999). These findings are consistent with other shod/barefoot comparisons in walking and running (De Wit et al., 2000). In an observational study, Finnis and Walton (2008) utilised video data to study pedestrians walking and identified a reduced average walking velocity in the people walking in flip-flops. This was attributed to a reduction in stride length compared to other footwear, which has correspondingly been identified in a laboratory environment when compared to walking in trainers (Shroyer and Weimar, 2010). However, it is likely that those walking in flip-flops are undertaking a leisure activity, in contrast to shoes such as brogues, which may be indicative of someone walking faster to work. In future research the consideration of a persons' activity should be included in observational studies, or gait velocities should be compared within controlled environments.

2.4.3 Issues Related to Toe-Post Footwear

Numerous specific features of toe-post footwear have resulted in specific criticism of the footwear. These include that it does not protect the sole, does not control frontal plane foot motion and causes the wearer to grip the shoe to hold it in place. Further exploration of these features and their associated symptoms or conditions would enable design improvements to be inferred if required.

2.4.3.1 Lack of Protection of Sole

Due to the thin sole and lack of upper inherent in the design of flip-flops, the style has been criticised for not protecting the sole of the foot during walking. This implies both not protecting the skin surface from damage from puncture wounds, and not absorbing shock, or alleviating pressure on the sole of the foot. Shroyer and Weimer (2010) undertook a 2D laboratory study on college students considering the ankle angle and kinetics at heel-strike.

The study assessed 56 college students (37 women) wearing their own flip-flops and trainers. Data was collected using a video camera (kinematic, operating at 30 Hz) and a force plate (kinetic, operating at 1000 Hz). Additionally, the study reported an interaction effect for gender and footwear on “attack angle”; a variable determined by the angle between the maximum vertical ground reaction force at heel-strike and the corresponding anterior-posterior value. This variable represents an interaction of walking velocity and touchdown position and energy. The paper reported a significant interaction effect; women had a greater mean attack angle in flip-flops (82.19°) versus trainers (81.39°), in comparison men had a greater mean value in trainers (81.90°) compared to flip-flops (81.66°). This small difference, despite being statistically significant would translate to approximately 4 N difference in the horizontal or vertical force. Without the authors presenting repeatability information or measurement error, it is difficult to establish whether this magnitudes exceeds the error of the protocol and whether this difference is meaningful in real terms should be established. The study also did not normalise forces to body weight then suggested significant gender difference in kinetics, with a variable that would be largely influenced by body mass (Shroyer and Weimar, 2010). The presentation of variables relating to loading rate and forces normalised to body weight would have increased the value of this paper in establishing the forces applied to the body at touchdown when walking in flip-flops.

Carl and Barrett (2008) praised flip-flops for providing protection to the plantar surface of the foot, after their results demonstrated that the flip-flop footwear reduced plantar pressures compared to barefoot walking. The study compared plantar pressures in Nike™ trainers, standard flip flops and barefoot (the insole affixed with double sided tape and a thin cotton sock). Three walking trials were undertaken at the participant's self-selected velocity, along a 9.1 m walkway for each shoe. Peak pressures for the hallux, metatarsal heads and calcaneus were recorded from specific pre-defined regions using the F-Scan system. Unfortunately, the study did not report the pressure values recorded for comparison with other studies; only the p values were included in the paper, so no plantar pressure data is currently available in the literature for walking in flip-flops. Significant differences were reported between peak pressure in the calcaneus and metatarsals regions, with barefoot the highest, followed by flip-flop then the trainer condition. This does not support the hypothesis of the authors, nevertheless flip-flops should reduce plantar pressure compared to barefoot as there is a viscoelastic material being placed between the foot and the floor. Consequently pressure would be alleviated in these areas due to the material absorbing energy. The study would have benefited from a controlled walking velocity as, it is likely that the flip-flop condition produced a slower walking velocity than the trainer (Finnis and Walton, 2008; Shroyer and Weimar, 2010). Therefore, comparison of pressures between the two is misleading as plantar pressure increases with increased walking velocity within-subject (Burnfield et al., 2004). This protocol, however demonstrates greater external validity depending on conclusions and research aims. In a further plantar pressure study on sandals/flip-flops, Song et al. (2005) compared reported comfort, peak plantar pressures and pressure time integrals in five different Birkenstock™ sandals with differing arch heights. Findings suggested that mid-range arch heights produced the lowest pressure time integral beneath the first metatarsal-phalangeal joint and were reported to be the most comfortable, likely due to the increased contact area. These findings allude to a profiled footbed in an open shoe posing comfort benefits, likely due to a redistribution of pressure and increased contact area (Che et al., 1994). The aforementioned studies highlight that the footbed of a flip-flop protects the sole of the foot when compared to barefoot and that modifying the profile of the footbed may be beneficial. Future studies should consider loading variables relating to heel-strike such as peak tibial acceleration and loading rate of the ground reaction force.

2.4.3.2 Does Not Control Frontal Plane Foot Motion

Specific studies on flip-flops are more limited than general barefoot/shod comparisons; however research on both children and adults exits in this footwear. Shroyer (2009), as part of his thesis, undertook data collection on 79 females (age: 21.54±1.53 years, height: 1.65±0.58 cm, mass 63.53±10.61 kg) walking in barefoot and three flip-flop styles. The work identified a reduced eversion during midstance when walking in a flip-flop (-3.39±3.10°) compared to barefoot (-4.00±1.78°) in the participants with a normal arch height (N = 53). Despite reporting significant differences, the absolute difference in eversion experienced ranged by around only half a degree between conditions in Shroyer's work (Shroyer, 2009). This data was collected using 3D motion capture and the Oxford Foot Model (with 14 mm markers) setup in Vicon collecting with 6 cameras operating at 100 Hz. Unfortunately, the author does not report any specific repeatability and validity data and the meaningfulness of half a degree difference lacks context in this situation. Accurate and precise 3D motion data is essential for the quantification of such minute changes in foot and ankle motion. Three-dimensional motion capture is reported in a range of footwear biomechanics research from studies on unstable footwear (Nigg et al., 2006; Romkes et al., 2006), high-heels (Voloshin and Loy, 1994), children's footwear (Wolf et al., 2008) and running footwear design (Hardin et al., 2004). Electromyography is commonplace in literature relevant to the biomechanics field, with research considering orthotics (Tomaro and Burdett, 1993) and insole texture (Nurse et al., 2005). These techniques allow insight into the interaction of the wearer with the footwear in a walking or running environment; how footwear affects both their muscle activation and resultant kinematics. These techniques could be implemented in an assessment of adults walking in flip-flops, to fully quantify any gait modifications walking in this footwear style.

2.4.3.3 Requires Gripping in Swing

The motion of the hallux has been proposed as a method to "grip" toe-post style footwear to hold the footwear onto the foot during swing and position the sole under the heel in preparation for initial-contact (Carl and Barrett, 2008; Shroyer, 2009). Carl and Barrett (2008) concluded that differences in peak pressure under the hallux were expected between conditions, due to the expectation that subjects grip flip-flops. The study however could have included more diverse and valid variable than solely peak pressure under the hallux to establish this and validly arrive at this conclusion. The peak pressure underneath the hallux in asymptomatic gait would occur close to toe-off under the influence of maximum placement of force through this region. The "gripping phenomenon" would not necessarily the most prevalent at this time and may not be effectively differentiated from standard plantar pressures. Thus, establishing variables that were observable in swing in the small time frame pre- or post-toe-off may have been more informative and both validly supported their conclusions and demonstrate a gripping mechanism in flip-flops. Shroyer (2009) also considered gripping with the hallux and quantified hallux motion in swing using a multi-segment foot in normal arch participants in a flip-flop compared to barefoot. The variable chosen was peak hallux extension after toe-off, which did not differ between the two conditions. Again, the variable selected may have not identified the true motion, due to the wearers gripping the footwear by flexing the hallux, as opposed to extending the hallux to hold the upper on the toe. Further exploration of the data and examining variables such as peak hallux flexion in swing may have further clarified this relationship. Despite this, the authors identified an increased plantarflexion during swing in the barefoot conditions, which they attributed to the flexor digitorum longus and flexor hallucis longus being contracted to keep the flip-flop on the foot (Shroyer and Weimar, 2010). These mixed conclusions and comments suggest that further research should assess electromyography in this footwear style. It is evident that quantifying the pressure beneath the hallux would enable any method of gripping the sole of the flip-flops to be quantified.

2.4.4 Literature Summary

The literature currently existing relating to toe-post footwear is minimal and some of the variables selected by authors may have not specifically addressed the hypothesis or supported the conclusions that were drawn. Future work is required to investigate these potential symptoms and detrimental outcomes for wearers of flip-flop footwear using methodologies and variables which specifically quantify variables of interest to measure the biomechanical influence of such footwear on users.

2.4.5 "Gait Modifications": Key Points

- The modifications to gait imposed by walking in toe-post footwear are yet to be fully defined in a range of wearers and therefore most criticisms and aetiology of potential injuries remain anecdotal.
- Research on toe-post footwear is minimal, and previous studies that consider motion analysis and pressures have some methodological constraints, which reduce the validity of conclusions.

- Both in-shoe plantar pressure devices and gait analysis methodologies are useful tools to compare the effect of different footwear on walking kinematics and plantar pressures.
- Bespoke insoles are required to ensure all data is captured on the plantar surface of the foot when testing toe-post footwear with no data-loss.
- Variables defined should be specific and isolate characteristics, which are deemed to be detrimental or indicative of modification and design constraints of the footwear.
- Studies considering alternatives to standard flip-flops, in addition to comparing flip-flop walking to barefoot, are advantageous in that they suggest replacement footwear for wearers who may not be willing to modify their shoe-wearing behaviour due to the convenience or practicality of flip-flops.

2.5 "Comfort"

Comfort is a multi-factorial concept, which is widely considered in footwear biomechanics literature with variables such as in-shoe pressure measurement and shock absorption. This literature review aims to identify and critique literature considering comfort in footwear which currently exists (June 2012). 'Health and well-being' footwear is expected by consumers to be comfortable; hence methodologies to quantify comfort in addition to footwear features which influence comfort are relevant to the current investigation of biomechanical concepts in this footwear category.

2.5.1 Introduction

Comfort is an important factor in the decision of a wearer to purchase footwear. It is subjective and is therefore influenced by an interaction of psychological, physiological and mechanical factors (Alcántara et al., 2005a; Au and Goonetilleke, 2007; Mills et al., 2010). Literature has reported perceptions of comfort and, as a consequence, reported comfort of footwear to be affected by:

- Material properties: the upper and footbed materials and constructions.
    - Cushioning (Alexander and Jayes, 1980; Hennig et al., 1996)
    - Stiffness (Miller et al., 2000)
    - Upper characteristics (Jordan et al., 1997)
- Wearer characteristics: anthropometric and biomechanical aspects of the wearer, including foot-shape and skeletal alignment.

    - Foot and leg anthropometric characteristics (Miller et al., 2000).
    - Gender differences (Wunderlich and Cavanagh, 2001)
- The activity: perceptions of comfort and perceptions of desired comfort vary across different activities.
    - Jogging or walking (Mündermann et al., 2002)
    - Standing or walking (Miller et al., 2000)
- Shoe dimensions: the fit of the shoe to the wearer's foot; if too loose will lead to slippage and friction, if too tight will compress tissue.
    - Control, stability and lace tightness (Hagen et al., 2010)
    - Perception in foot regions (Au and Goonetilleke, 2007)
    - Footbed plantar surface shape (Alexander and Jayes, 1980)
    - Shoe heel height or pitch (Alexander and Jayes, 1980)
    - Pressure at specific locations on the foot (Cheng & Hong 2010)
- Shoe climate: the humidity and temperature within the shoe relative to the environment.
    - Sock construction in running (Davis et al., 2008; Hennig et al., 2005)
    - Perspiration and blistering (Barkley et al., 2011; Davis et al., 2008)

Another less frequently reported aspect includes the sound the shoe makes when it impacts with the ground influencing comfort perception (Au and Goonetilleke, 2007). Additionally, external factors such as advertising copy regarding cushioning in running shoes have been reported to modify movement patterns at heel-strike (Robbins and Waked, 1997), altering the wearers perception of comfort in the shoe.

Methods used to quantify measures of comfort are generally subjective scales (Au and Goonetilleke, 2007; Mündermann et al., 2002), or ranking of shoes in order of preference post-wear (Che et al., 1994; Mills et al., 2010). Often these subjective measures are combined with objective measures such as foot plantar or dorsal pressures (Jordan et al., 1997; Wegener et al., 2008). The SATRA Comfort Index, used by footwear manufactures in the United Kingdom to assess the comfort of their styles, utilises a four stage process (SATRA, 2009):

- fit assessment
- aesthetics and handling (softness, flexibility, texture)
- moisture disposal assessment
- treadmill walking with in-shoe pressure assessment

Including aesthetic aspects when quantifying comfort is mirrored during in-house testing by footwear companies as this can sometimes include aesthetic aspects of the footwear as opposed to a pure measure of the feel of the shoe on the foot. This has disadvantages in that the aesthetics of the shoe are not directly associated with the fit or comfort of the shoe; however it is advantageous in that it better reflects users purchasing patterns (Kong and Bagdon, 2010). The measurement of comfort via questionnaires and its relationship to other measures (correlates) are discussed below.

2.5.2 Methodologies in Comfort Measurement

Due to the subjective nature of comfort and its variability across people, questionnaires are often used to obtain measures of the wearers' perception of the comfort of the footwear. This subjective data collection is often coupled with objective measures of pressure and impact.

Research stemming from educational institutions is generally scale-based and normally encourages the assessment of comfort as opposed to the subjects' pre-formed expectation of the comfort of the footwear from aesthetics. To do this some studies blind subjects from the footwear style (Wegener et al., 2008). When using a scale to assess comfort researchers must decide which factors to evaluate then design scales to assess this, and due to the subjective nature of comfort this may not provide useful information from all participants. Due to this restriction some authors have utilised interviews post-wear and psychological theories in order to better quantify comfort of footwear (Alexander and Jayes, 1980) as well as semantic analysis to determine appropriate terminology to obtain accurate reports from study participants (Alcántara et al., 2005a, 2005b). Kong and Bagdon (2010) attempted to better replicate footwear selection in their methodology by simulating selection in a running shoe shop prior to testing footwear, allowing subjects to try the shoe, move around in limited space (walking and running) then select one shoe. However, in the context of the present research and other work led by footwear companies, the specificity of the scale is linked to a footwear design feature or region. Therefore, the scale may not entirely encapsulate "comfort" as a concept, nonetheless it can be designed to capture aspects of the tested footwear that have been, or can be, modified depending on the research design. A range of the questionnaire methodologies utilised in peer-reviewed, published literature are presented in Table 2.3, from which selected notable studies are further described and discussed.

In addition to influencing the participants reporting of comfort, the scales utilised to quantify subjective measures of comfort influence the statistical interpretation of the data

(Mündermann et al., 2002). Likert-type scales where data is categorised limit analysis to ranking of data. Ordinal scales are ordered from least to most, and limit the interpretation of data as there is no indication of the absolute difference between the conditions, or the actual measure. For example four shoes may be ranked 1-4 (1 being most comfortable) and they may all be uncomfortable. An ordinal scale (such as an adapted Borg scale) provides a 7- or 15-step ranking, which does enable the relative difference to be measured, however only to the resolution of the ranking. Also, the use of scale data impacts on the statistical analysis that can be undertaken. Non-parametric statistical techniques must be utilised and any correlations with biomechanical or other variables will include errors due to the resolution of the scale. Numerous authors have undertaken erroneous statistical analysis of comfort questionnaire data, for example Jordan et al. (1997) undertook correlations on ranked data from a Likert scale. VAS are a well validated and widely used tool to assess opinion which produces continuous data for more diverse statistical treatment and interpretation. Other consideration given to comfort scales must include:

- The audience
  - The language and semantics of the scale anchors and instructions must be understandable
  - The footwear must be relatable to their own or they must have time to familiarise to the shoe prior to testing
- Questionnaire fatigue.
  - Number of questions
  - Number of repeats
  - Total duration of testing
- Use of a control condition for referral.
  - Whether used
  - What shoe
  - How many times applied
- Relevant protocol.
  - Walking, incline, incorporate standing, running
  - Appropriate surface e.g. football boot on turf
- Aims of assessment.
  - Desired outcomes
  - Assessed features
  - Data required
  - Interpretation

(Alcántara et al., 2005; Mills et al., 2010; Mündermann et al., 2002).

Mündermanns' original methodology was based on the prediction that the repeatability of an insoles comfort rating can be increased if the conditions are compared to a control, as a subjective comparison is needed by the participant (Mündermann et al., 2002, 2001). The questionnaire was also adapted slightly to include 9 questions, assessing the overall comfort, comfort of forefoot cushioning, heel cushioning, heel cup fit, shoe heel width, shoe forefoot width and shoe length. The 150 mm VAS were anchored with "not comfortable at all" and "most comfortable condition imaginable" and a detailed instruction and descriptions of each aspect on the questionnaire was included, for example "arch height:" was defined as the "medial arch height of the insole". Findings from the results included that the control condition was rated more consistently than the insert conditions and some subjects were very consistent and others demonstrated large fluctuations in their ratings. The average difference between repeated comfort ratings equalled 0.53 comfort points. This was relatively large considering all conditions tested were ranked in the range 4.15-5.78 comfort points. This clustering of comfort scores is likely due to the same footwear being used for each condition with a new orthotic positioned in the shoe. The use of the VAS varied between subjects, with some using the whole range and others clustering scores and using only a fraction of the scale. The intra-class correlation coefficient (ICC) for intra-test repeatability was 0.799 for all subjects and all conditions (range ICC = 0.108-0.952 for individuals). Inter-test variability decreased with increasing session number and for a long-term comfort assessment the authors recommended reporting mean data from session's four to six. Overall the study results identified that subjects preferred soft to hard insoles and medial wedge insoles rated significantly lower than flat control.

Table 2.3 Comfort questionnaires.

Reference	Scale type	Format	Protocol
Che et al., 1994	Ranked	Shoes ranked in order of preference 1-4.	Walked for a self-selected duration and velocity on treadmill.
Hennig et al., 1996	15 point Likert	Anchor: “Very very hard” to “very very soft” Question: Perception of cushioning.	5 min treadmill runs. 8 repeats for each shoe.
Jordan et al., 1997	5 point scale	Anchor: 1 very uncomfortable, 5 very comfortable. Question: For individual areas of shoe (6 regions). Images of foot and shoe.	Subjects tied laces, walked 1.6 steps/second. 25 min walk in 3 shoes.
Miller et al., 2000	Modified Borg 10 point	Anchor: 10 = “very comfortable”, 1 = “very uncomfortable”.	Standing, walking 200 m and running 600 m.
Mündermann et al., 2001	8 questions	Anchor: not comfortable” right “very comfortable”. Questions: Forefoot cushioning, heel cushioning, arch height, heel cup, shoe heel width, shoe forefoot width, shoe length, overall comfort.	Marched 500m across a variety of surfaces then rated. Control condition was military boot.
Mündermann et al., 2002	150 mm VAS	Anchor words: “not comfortable” and “most comfortable condition imaginable”. Scales: overall comfort, heel cushioning, forefoot cushioning, medial-lateral control, arch height, heel cup fit, shoe heel width, shoe forefoot width, shoe length.	45 min session, 10 consecutive days, 2 laps of 450 m track in control to warm up. 1 lap for assessment. Mean score of sessions 4-6 recommended.
Llana et al., 2002	Interview	Interview assessed global comfort, subjective opinions of errors in design and discomfort of parts of the body.	Long-term study.

Hong et al., 2005	100 mm VAS, Mündermann scale	Anchor words: “not comfortable” and “most comfortable condition imaginable”. Scale: overall comfort only.	Treadmill walk 5 min. 1.3 m/s for warm up. Level walk way 5 min rest before.
Au and Goonetilleke, 2007	Likert	Perception: 16 item 1-7 rating, anchor words “strongly agree/strongly disagree”. Question: e.g. “this is my favourite brand”. Fit: fit preference rating 1-7, anchor words “very tight” to “very loose” and preference 1 “strongly agree” to “strongly disagree” Question: e.g. “I like the fit”.	5 min break between. Walked for self-selected duration 5 min (max) treadmill walking for fit preference.
Wegener et al., 2008	150 mm VAS, Mündermann scale	Anchor words: “not comfortable” and “most comfortable condition imaginable”. Scale: overall comfort only.	5 min. No control shoe between conditions.
Mills et al., 2010	100 mm VAS 7 scale Likert	Anchor: “not comfortable” and “most comfortable condition imaginable”. Questions: overall comfort, cushioning of the forefoot, arch and heel and support of the arch and heel.	2 min run or walk on a treadmill at self-selected, comfortable pace.
Barkley et al., 2011	100 mm VAS	Anchor words: “least comfortable imaginable” and “most comfortable imaginable”. Scale: overall comfort only.	10 min constant velocity on treadmill, questionnaire administered alongside a foot temperature scale.
Burke, 2012	150 mm VAS	Anchor: “most uncomfortable” and “most comfortable”. Questions: overall comfort.	Single-leg balance task while rating comfort for two orthoses in a trainer.

Where VAS = visual analogue scale, min = minute.

Literature focuses on the use of the full Mündermann et al. (2003) comfort scale (Wegener et al., 2008), or on sections of the scale used in isolation (Barkley et al., 2011) to quantify comfort in footwear This tool has been validated and shown to be repeatable when testing running shoes on a healthy running population utilising a control shoe as every other condition (Mündermann et al., 2002). Despite this specific validation of the protocol and questionnaire, the scale (or a slightly adapted version) is widely used by a range of footwear studies using a range of protocols:

- Walking tests (Anderson et al., 2005; Davis et al., 2008).
- To compare heeled shoes (Yung-Hui and Wei-Hsien, 2005).
- With no control shoe in use (Davis et al., 2008; Wegener et al., 2008).
- On specific population groups (Wegener et al., 2008).

The application of this scale to shoes that are not running shoes and users who are not runners is questionable. The scale includes aspects of consideration, which are specific to running shoes and terminology that may not be clear to the general wearer (i.e. "medial-lateral control"). It is evident that the moderation and re-validation of such a scale may provide better measures of comfort for a walking shoe from the general population. The population, activity and footwear may all impact significantly on the use and appropriateness of a comfort scale and the reliability and minimal clinically important difference must be established to infer meaningful information from results (Mills et al., 2010).

The reliability of the scales identified in Table 2.3 for footwear comfort assessment has only been established by a few authors (Kinchington et al., 2011; Miller et al., 2000; Mündermann et al., 2002). Mills et al. (2010) highlighted the importance of stability and relevance when designing a comfort assessment tool. They proposed a method for determining a minimal clinically important difference (the smallest change in score that the patient perceives to be a beneficial increase in comfort). The author constructed a data collection and analysis procedure that assessed Likert scale, VAS and ranking scales for footwear influence and the impact of gait velocity on outcome measures, repeating the process over five days. Subjects were tested in their own running shoe (control condition) and a further four insert conditions, blinded from the condition that they were wearing. The VAS scale design utilised six horizontal 100 mm lines anchored with "not comfortable at all" and "most comfortable imaginable", consistent with Mündermann et al. (2002). The seven point Likert scale ranged from "very comfortable" to "very uncomfortable". The scales addressed overall comfort, cushioning of the forefoot, arch and heel, and support of the arch and heel. Similar to

Mündermann et al. (2002), these scales are running-shoe specific and as a result the validity and application outside of assessing a jogging shoe is questionable. An exit questionnaire was also prescribed to assist in the development of the minimal clinically important difference (Mills et al., 2010). The study results identified that the VAS was stable between gait velocities (walking and jogging). However this required a minimum repeat of two sessions to gain stability, with aspects relating to the heel requiring three days. Ranking overall comfort was not significantly different between sessions or gait velocity with the VAS, suggesting that an overall measure as used in numerous studies (Barkley et al., 2011; Burke, 2012) is repeatable with a VAS, but not with the Likert scale (Mills et al., 2010). In this study, the subjects identified that the arch was the most important consideration for comfort (Mills et al., 2010), however it may be that this differs when a more diverse range of footwear is tested, similar to Mündermann et al. (2003, 2002) every condition in this study has the same shoe and consistent upper material, fit etc. The minimal clinically important difference was identified as 9.10 (6.16-12.02) mm in this study, with the authors concluding this difference in VAS scores represents a change in comfort level (Mills et al., 2010). In a further exploration of the data, using the results from subject questioning after the testing, Mills et al. (2010) identified a mean minimal clinically important difference of 10.2 mm, however one subject stated only 5 mm and another 25 mm suggesting that this value is not likely to be consistent between wearers.

2.5.3 Footwear Comfort Findings

Comfort and aspects of comfort have been assessed using subjective and objective measures with the aim of providing a link between the two (Cheng and Hong, 2010; Whittle et al., 1994; Yung-Hui and Wei-Hsien, 2005). This would allow consumer preferences to be categorised and shoes to be adapted appropriately. For example, if wearers consistently rate a shoe as more comfortable while the plantar pressure decreases under the first metatarsal then the footwear technology team can look at incorporating this mechanism in the shoe to reduce this pressure. Llana et al. (2002) demonstrated this principle of application of comfort measures while assessing the comfort of tennis shoes. The authors considered regional and global discomfort and how the first contributes to the latter. The authors took a new approach to the assessment of comfort, ultimately attributing factors raised by interviewees to design errors in the footwear. For example, if the wearer raised concerns relating to discomfort in the heel or lumbar spine, this was attributed to inadequate shock absorption and if the shoe was described as too loose then this was attributed to an incorrect last. Comfort perceptions have been directly related to objective measures of the footwear and wearer, most commonly impact variables and pressure on the dorsal and plantar surfaces of the foot and anthropometry.

2.5.3.1 Impact

Human pendulum apparatus have been utilised to assess the link between perceptions of impact severity and impact variables (Lafortune and Lake, 1995; Lake and Lafortune, 1998). The nine foam hardness conditions were each administered nine times for each subject and were rated based on a control value where an impact was prescribed as a specific number by the researcher and all other impacts were rated by the subject relative to this. This applies the same control principle, but contrasts the control shoe method utilised by Mündermann et al. (2002). However, it may prove more repeatable to provide a standard value when a more abnormal protocol is being utilised, which induces an unfamiliar sensation to participants, such as the human pendulum technique. The outcome variables demonstrated that impact severity perception increased with faster impacts and harder surfaces. Every mechanical variable was significantly correlated with perceived impact severity when mean data was utilised. Of all the variables time to peak impact force had the lowest correlation with perception (r = -0.77) and regression analysis identified transient rate accounted for 64% of variability in perception. This study highlighted that perceptions of running impact severity are closely related to the rate of force loading and these are likely to be associated with perceived comfort in running. Hennig et al. (1996) further confirmed this finding with runners wearing varying stiffness midsole constructions. This time the perceptions of cushioning was quantified using a modified Borg scale with 15 points ranging from “very very hard” to “medium hard”/”medium soft” and “very very soft” with no control value. The perception scores differed significantly between conditions with soft (10.9±3.4), medium (8.6±2.9) and hard (3.3±0.7), identifying how perceptions of cushioning are related to footwear midsole construction. The impact peak of the ground reaction force decreased as the perceptions of hardness increased (r^2 = 0.99). Maximum rate of force loading (r^2 = 0.89) and the median power frequency of the ground reaction force (r^2 = 0.99), and peak pressures in the heel (r^2 = 0.97) increased with perceptions of less cushioning. Despite including only three shoes in this correlation analysis, these results are consistent with those found later in running and stronger correlations than those evident in the human pendulum method (Lake and Lafortune, 1998). This perception of cushioning of a shoe is not necessarily comfort. The cushioning of a shoe does not necessarily make it more comfortable and this is a subjective decision from the wearer. However, cushioning is a variable for consideration with relevance to the holistic concept. The application of these results to walking velocity where impact variables will be less severe and may show reduced variation between conditions may be more difficult, similarly a more realistic hard shoe may further obscure relationships.

2.5.3.2 Foot Pressure

Plantar and dorsal pressure measurement have both been utilised in studies relating to footwear comfort (Jordan et al., 1997; Wegener et al., 2008) and fit (Cheng and Hong, 2010; Olaso et al., 2007).

Plantar pressure research identifies mixed conclusions when attempting to relate plantar pressures to comfort. Jordan et al. (1997) demonstrated that peak pressures were significantly greater in “uncomfortable” shoes for the rear- and fore-foot, but significantly lower for the midfoot region. The relationship identified between perceived comfort and peak pressure was negative across the whole surface of foot. Wegener et al. (2008) collected plantar pressure data alongside comfort in runners with cavus feet with the premise that it is clinically necessary to reduce and redistribute pressure in cavus feet. Overall in the study there were no statistically or clinically significant relationships between footwear comfort and plantar pressure variables in the whole foot, rearfoot, midfoot or forefoot in the three shoes tested. Contrasting Jordan et al. (1997), the footwear with the lowest plantar pressure was not the most comfortable. Additionally, Clinghan et al. (2007) identified no relationship between perceived comfort and plantar pressure data (r = .081) when investigating athletic shoes of differing price. There were, however, minimal differences in pressure and comfort ratings between the footwear conditions in this study, potentially masking any potential relationships. Che et al. (1994) utilised a contrasting approach and compared the two insoles which participants had ranked as the most and least comfortable. The results demonstrated that the total foot peak pressure, pressure-time integral and contact area were significantly smaller for the most comfortable insoles compared to the least comfortable insole. Regional differences in pressure relating to comfort were also evident, particularly in walking. In walking there were significantly higher pressures in the midfoot (+16.5%) and lower in the medial forefoot (-16%) and hallux (-23%) in the most comfortable compared to the least comfortable insole. In agreement, Lange et al (2009) identified the peak pressure in the medial forefoot (a sensor under the second and third metatarsal heads) was negatively related with comfort and the relative load in the heel region positively in different military boots. These findings mirror those of Jordan et al. (1997) in different footwear and led the authors to conclude that an even distribution of pressure across the plantar surface of the foot was the most comfortable insole condition.

Plantar pressures have been more extensively investigated than dorsal pressures in relation to comfort. This is likely due to dorsal pressures being, in part and in some regions and footwear styles at least, controllable by lace tightness. Jordan et al. (1997) quantified dorsal force at both the flex-line and the lacing, identifying maximum force to be significantly higher and contact area significantly lower in uncomfortable footwear. Hagen et al. (2010) considered dorsal pressures further in their study on lace tightness. They identified that perceived comfort and dorsal pressure were highly correlated (r = 0.84, p < .05). They tested three lacing patterns and one tightened and the data analysis used a masked region encompassing the talus, navicular and extensor tendons. Olaso et al. (2007) utilised dorsal pressures to investigate fit of footwear and how pressure/force is affected at specific anatomical points. Unfortunately the study only tested four subjects and the adjustments to the full upper of the shoe were not clear to accommodate the reported changes in metatarsal-phalangeal joint girth. It is however, clear from the literature that higher dorsal pressures lead to perceptions of discomfort in wearers. These may prove to be more influential in sporting activities where dorsal pressures on the anatomical points on the lateral border of the foot may be particularly high (Greenhalgh and Sinclair, 2012).

From the results discussed, it is evident that the use of plantar and dorsal pressure measurement may be able to differentiate between comfortable and uncomfortable footwear. The results identify that footwear that reduces specific dorsal regional pressures and more evenly distributes plantar pressure would likely be deemed more comfortable by wearers. It is likely that the exact relationship between plantar pressure in specific footwear regions and reported comfort therefore is likely dependent on the participants and the footwear being compared.

2.5.3.3 Anthropometry and Fit

Foot size and anthropometry has been related to perceptions of comfort in running shoes and orthoses (Cheng and Perng, 1999; Miller et al., 2000; Mündermann et al., 2001). Miller et al (2010) identified a significant influence of foot and leg measurements on ratings of perceived comfort, particularly highlighting toe-box depth and width of the heel of the shoe as important for a comfortable fit. Cheng and Hong (2010) considered the difference in dimension between their participants' foot size and the last dimensions as opposed to just an absolute measure of foot dimensions and size as used in other research (Miller et al., 2000; Mündermann et al., 2001). The measure of the instep circumference was significantly related to mid-foot fit of the shoe (r = .120, p < .000) and the overall fit of the shoe (r = .284, p < .000) as identified by a 12 scale visual analogue scale. The regression equation established utilised the measures of heel breadth and ball girth circumference to explain 31% of the variance in reported subjective fit. This study, thus, points to heel breadth being an important factor in determining the subjective comfort of footwear. Similarly, Au and Goonetileeke (2007) highlight the importance of favoured fit at the metatarsal-phalangeal joint and toe for a comfortable shoe. Some studies however, have identified no significant relationship between foot size and reported comfort (Kunde et al., 2009). This may be due to preference, as some participants may prefer a tighter feel to a shoe and others looser. Familiarity may be an additional factor as some wearers may more commonly wear smaller or tighter-fitting shoes and therefore not perceive a tighter fit to be detrimental to comfort whereas others who are not used to this sensation may report discomfort. Furthermore, Au and Goonetileeke (2007) determined that despite aesthetics not discriminating uncomfortable and comfortable footwear, it explained 18.4% of the response variance in comfort.

2.5.4 Literature Summary

It is apparent that numerous methods to quantify both subjective and objective comfort are utilised in footwear literature and could be applied to quantifying the concept of comfort in 'health and well-being' footwear. Future work should establish and validate an accessible comfort scale for walking footwear such as undertaken by Mündermann et al. (2002) for running. This can then be utilised in on-going research to consider the relationship between subjective and objective measures of comfort of the foot in walking shoes.

2.5.5 *“Comfort”: Key Points*

- Comfort is one of the most important factors for a consumer purchasing footwear.
- The repeatedly used comfort questionnaire for footwear analysis includes running-shoe specific terminology.
- To establish an effective questionnaire, the semantics, instruction and format should be clear and coherent to the reader and the scale valid and appropriate to the participants, the footwear tested and research aims.
- Both comfort perception and cushioning perceptions can differentiate between comfortable and uncomfortable footwear and plantar pressure and accelerometer measures can differentiate between these.
- Comfort is highly subjective and coupling questionnaires with objective measures may prove to be the most useful approach for footwear development.
- Tools for footwear companies to bench-mark footwear comfort in existing products to new developments or market leaders may be useful for internal research and development purposes to quantify consumer preference.

Chapter 3 Publications

3.1 Publications and Candidates Work

This section addresses the contribution that each co-author made for each paper. In addition Appendix A includes letters/statements in support of the work undertaken by the candidate and co-authors for each individual publication included within this submission.

3.1.1 A mechanical protocol to replicate impact in walking footwear

Price, C & Cooper, G & Graham-Smith, P & Jones, R

Gait and Posture 40(1), 26-31.

RJ/PGS identified a need for the paper and CP conducted a literature review to develop a protocol and relevant variables. GC/CP worked on a protocol which utilised an accelerometer constructed by GC. GC/CP collected the human test data and GC designed and constructed a drop-device to test a standardised mechanical testing protocol with the technicians listed in the acknowledgements. GC/CP collected the data with the drop-device utilising a standardised protocol identified in the literature review and a new protocol based on the human data collected. CP analysed all data using Visual 3D, conducted the statistical analysis and drafted the paper. Following reading a final draft RJ identified a need for further testing, which GC and CP developed a protocol for and conducted. Paper revisions were undertaken by CP with input and discussion with GC. RJ made structural and grammatical recommendations and approved versions as the paper progressed.

3.1.2 The manipulation of midsole properties to alter impact characteristics in walking footwear

Price, C & Cooper, G & Jones, R

Footwear Science: in review.

CP and GC discussed a paper plan and protocol to continue the work undertaken in paper 4. CP then undertook the testing with the device developed by CP/GC for the previous paper utilising the method defined by GC/CP. CP analysed the data utilising Visual 3D, undertook the statistical analysis and wrote a complete first draft of the paper. GC then met with CP to finalise the paper for submission. Paper revisions were undertaken by CP with GCs assistance. RJ reviewed drafts of the paper.

3.1.3 The effect of unstable sandals on single-leg standing

Price, C & Smith, L & Graham-Smith, P & Jones, R

Footwear Science 5(3), 147-154.

For both papers relating to unstable shoes CP conducted the literature review, constructed the idea and wrote the protocol for the testing. LS contributed to the literature search and revisions of the review. LS and CP collected the data utilising Qualisys, force plates and EMG together. RJ and other research staff had previously demonstrated the use of Qualisys and associated software. LS and CP labelled motion data ready for data processing. CP wrote the analysis script within Visual 3D to process the data and calculate the relevant variables. CP conducted the re-structuring of data to present variables and undertook the statistical analysis. The first draft of the paper was produced by CP and LS, following drafts were worked on jointly by CP and LS. RJ and PGS reviewed drafts throughout the paper-writing process and made recommendations. CP conducted first-stage revisions in discussion with LS, on completion RJ/PGS approved prior to re-submission.

3.1.4 The effect of unstable sandals on instability in gait in healthy female subjects

Price, C & Smith, L & Graham-Smith, P & Jones, R

Gait & Posture, 38(3), 410-415.

Please see above.

3.1.5 A comparison of plantar pressures in a standard flip-flop and a FitFlop using bespoke pressure insoles

Price, C & Graham-Smith, P & Jones, R

Footwear Science 5(2), 111-119.

CP conducted the literature review, wrote the protocol and liaised with Medilogic (acknowledged in the final paper) to instruct them on production of the bespoke insole. CP then recruited the subjects, collected the data for the repeatability and main study. The analysis procedure was then determined by CP and CP constructed the spread sheet to calculate relevant variables. Following processing, the statistical analysis was undertaken by

CP. RJ/PGS involvement was reviewing the paper at the final draft stages. CP undertook revisions of the paper, which were approved by RJ/PGS.

3.1.6 Does flip-flop style footwear modify ankle biomechanics and foot loading patterns?

Price, C & Andrejevas, V, Findlow, A & Graham-Smith, P & Jones, R

Journal of Foot and Ankle Research: accepted.

RJ/PGS identified the need for the study as part of the KTP pre- CPs appointment to the University. The protocol utilised a standard gait protocol for the laboratory developed by RJ. CP then recruited the subjects and collected 10 of them with the assistance of VA as part of his undergraduate dissertation. The analysis procedure was then determined by CP and CP constructed a script in Visual 3D which automatically processed the data. VA labelled the 10 subjects he collected and ran the Visual 3D script, CP labelled and analysed the additional 30 subjects. Following processing, the statistical analysis was undertaken by CP. The paper was drafted by CP with some additional comments and research identified by VA within his dissertation. RJ reviewed drafts throughout the paper-writing process and made recommendations. PGS reviewed a final draft paper. CP undertook various revisions of the paper based on peer-review feedback including re-analysing some data and altering the statistical approach. Changes made by CP during the revisions process were approved by RJ/AF prior to re-submission.

3.1.7 Subjective and objective variables to quantify comfort in walking footwear

Price, C & Jones, R

To submit.

RJ identified a requirement for work relating to footwear comfort. CP conducted a literature review and identified the research question and theme. CP developed the questionnaire and protocol with advice from RJ for commercial testing. CP conducted the testing, analysed all data and conducted the statistical analysis for the research report and paper. CP then wrote a first draft of the paper which RJ reviewed pre-submission. RJ recommended changes to the content, structure and data. CP made these and submitted the paper. Revisions were undertaken by CP and approved by RJ prior to re-submission.

Information about the Journals in which the work is published is included in Appendix B.

3.2 Publications

3.2.1 A mechanical protocol to replicate impact in walking footwear.

Price, C, Cooper, G & Graham-Smith, P & Jones, R (2014).

Gait & Posture.

Abstract

Impact testing is undertaken to quantify the shock absorption characteristics of footwear. The current widely reported mechanical testing method mimics the heel impact in running and therefore applies excessive energy to walking footwear. The purpose of this study was to modify the ASTM protocol F1614 (Procedure A) to better represent walking gait. This was achieved by collecting kinematic and kinetic data while participants walked in four different styles of walking footwear (trainer, oxford shoe, flip-flop and triple-density sandal). The quantified heel-velocity and effective mass at ground-impact were then replicated in a mechanical protocol. The kinematic data identified different impact characteristics in the footwear styles. Significantly faster heel velocity toward the floor was recorded walking in the toe-post sandals (flip-flop and triple-density sandal) compared with other conditions (e.g. flip-flop: $0.36\pm0.05 m{\cdot}s^{-1}$ versus trainer: $0.18\pm0.06 m{\cdot}s^{-1}$). The mechanical protocol was adapted by altering the mass and drop height specific to the data captured for each shoe (e.g. flip-flop: drop height 7 mm, mass 16.2 kg). As expected, the adapted mechanical protocol produced significantly lower peak force and accelerometer values than the ASTM protocol ($p<.001$). The mean difference between the human and adapted protocol was 12.7±17.5% ($p<.001$) for peak acceleration and 25.2±17.7% ($p=.786$) for peak force. This paper demonstrates that altered mechanical test protocols can more closely replicate loading on the lower limb in walking. This therefore suggests that testing of material properties of footbeds not only needs to be gait style specific (e.g. running versus walking), but also footwear style specific.

Key words: shock, footwear, heel-strike transient, accelerometer, impact testing.

1. Introduction

Testing is undertaken by footwear manufacturers to analyse properties of footwear prior to mass-manufacture to make design and component decisions. The testing undertaken depends on the style of footwear and can include sole traction or friction, outsole longitudinal stiffness and impact testing. Impact testing aims to quantify the shock absorbing capability of footwear midsoles by replicating the collision, and resulting transient, between the shod foot and the ground at heel-strike. The nature of this transient has been linked to degenerative changes to tissue such as knee osteoarthritis [1], clinical symptoms like lower back pain [2], as well as subjective comfort in healthy populations [3]. The manipulation of footwear or insole characteristics (thickness, shape and material properties) can attenuate loading from heel-strike, reducing the magnitude of forces and loading rate experienced by soft tissue, bone and joint cartilage in clinical [2] and healthy populations [4].

Some methods for examining heel-strike impacts involve dropping a mass onto the midsole and quantifying force, acceleration, energy dissipation and deformation [5,6]. Mechanical testing has obvious economic and time-saving advantages for footwear companies and allows a larger range of potential midsoles to be tested compared with testing on humans. For example Frederick et al., utilised mechanical testing to quantify a range of heel thickness (10-30 mm), midsole flare (0-30°) and hardness (25 to 45 Shore A) constructions, measuring peak gravity (g) in 36 footwear conditions [5]. Human testing, however, has the advantage of including the interaction of the human system with the footwear, for example any effect that the footwear may have on heel pad confinement [7], gait kinematics [8] and muscle activation [9] and therefore impact characteristics. Comparisons between mechanical and human impact data generally report low correlation with biomechanical tests [10]. For example, Hennig et al. identified a low, non-significant, correlation between peak tibial accelerations during running in 19 different athletic shoes in 27 subjects and the acceleration scores from a mechanical impact tester (r =.26)[11]. Making mechanical testing as representative of the real-life situation as possible therefore has significant benefits for the footwear technician who needs to make decisions based on the outcomes of mechanical testing alone.

The American Society for Testing and Materials (ASTM) stipulate a specific protocol to quantify the shock absorption properties of footwear (F1614 Procedure A, 2006), originally designed to replicate running impacts. This protocol utilises a drop-height (50 mm) and a missile-mass (8.5kg) to replicate the impact velocity and effective mass of the running leg and foot of a male running[12]. Despite the protocol replicating the energy apparent in ground-impact in running, it is used in footwear research considering marching [10], tennis [13] and walking [14]. It is also utilised by the Shoe and Allied Trade Research Association (SATRA) to test shock attenuation in all footwear styles from trainers to sandals [15]. These are conditions where impact energy will typically be significantly lower in a real-life situation. These loads and the duration over which they are applied are not relevant measures of the shock absorption properties of materials and constructions of walking footwear. The assessment of walking is relevant as it is a more frequent activity for the general population and in particular for clinical and aging groups to whom the heel-strike magnitude may be more detrimental [1,2]. It is also more relevant for orthopaedic and walking footwear styles, which are unlikely to be used for running. Therefore quantifying the cushioning properties of different walking footwear is highly relevant. It is likely that the differing uppers in footwear styles also influence the kinematics and therefore the impact experienced [16,17]. Thus adapting this protocol to better replicate the energy apparent in walking and specific styles of walking shoes would be an effective step in footwear biomechanics development for footwear manufactures. Testing protocols on material construction and data analysis and interpretation could then be undertaken more rapidly in footwear style-specific protocols.

The purpose of this study was to modify a mechanical test method (ASTM F1614 Procedure A, 2006) to better replicate walking impacts in a variety of walking shoes. The protocol was adapted using kinematic data from participants walking to produce a more valid method for testing walking footwear styles mechanically. Results from the new protocol were compared with the standard ASTM method in addition to the human results in real-life walking.

2. Methods

Ethical approval for the study was obtained through the University ethics committee and volunteers were recruited from the University staff and student populations.

2.1. Footwear Tested

Four footwear conditions were tested (Table 3.1, Figure 3.1), as well as barefoot using human and mechanical methods. The order of footwear testing was randomised among subjects.

Table 3.1 Characteristics and images of the footwear conditions tested alongside barefoot.

Condition	Image	Style	Heel material/ construction	Heel depth (mm)	Heel hardness (Shore A)
Flip-flop		Havaiana Brazil	EVA	16	33
Trainer		New Balance 539	EVA with microfibre linings	27 footbed 5 insole	52 footbed 26 insole
Shoe		Ecco Unisex (comfort brand)	Rubber outsole, cloth lining and EVA insole	5 outsole 5 insole	65 outsole 30 insole
Triple-density sandal		FitFlop Walkstar I	EVA	41	55

2.2. Human Testing and Processing

Thirteen healthy subjects (2 Male, 11 Female, 27.5±8.8 years, 62.0±10.3 kg, 1.65±0.05 metres) with shoe size U.K. 6 participated in the study. Subjects who reported no lower limb injury were instrumented with a lower limb marker setup for 3-D motion capture and one uni-axial accelerometer resonant at 3.0 kHz.

A 10 camera Qualisys Pro-Reflex system (Qualisys, Sävebalden, Sweden) was used to track 3D motion at 240 Hz. Spherical retro-reflective markers and clusters were positioned to define the lower limbs in accordance with the CAST technique [18]. The foot was defined with markers on the posterior calcaneus and the dorsal aspects of the 1st, 2nd and 5th

metatarsal heads. The shank was defined with anatomical markers on the medial and lateral malleoli and the medial and lateral knee with a rigid plate tracking marker on the anterior tibia. The accelerometer was mounted on the right anterior-medial tibia above the medial malleolus on a small piece of light flexible plastic. It was positioned 5-10 cm above the malleolus, on an area with least adipose tissue, oriented with the tibia axis. The accelerometer was affixed with double-sided tape and secured with an elasticated bandage tightly without causing discomfort. The accelerometer was sampled alongside 2 force plates (AMTI, Advanced Mechanical Technology Incorporated, Watertown, USA) at 2400 Hz collecting ground reaction force data for two consecutive right heel-strikes. Subjects performed 5 trials in each condition following a familiarisation period. Ten data-sets relating to the right limb were therefore analysed for each footwear condition. Walking trials were monitored with timing gates to ensure consistent walking speeds within a range of ±5% of their self-selected speed, trials outside this boundary were re-captured.

Data were processed and analysed using Visual 3D (C-Motion, Inc., Rockville, MD, USA), defining the right limb and pelvis as 4 rigid segments. Kinematic and kinetic data were filtered using low-pass Butterworth filters at 10 Hz and 100 Hz respectively [19]. Force plate contact was defined as the first frame in which the vertical ground reaction force (vGRF) exceeded 4N. Heel-marker vertical velocities at heel-strike were calculated using the mean value from 8ms leading up to heel-strike. This is within ranges found to be reliable in previous research [20]. The accuracy of velocity values calculated from kinematic data were verified by comparing them with velocities calculated from the first-integral of the accelerometer data sampled at 2400 Hz. In vivo heel-strike transient (HST) was defined as a local maximum point between the 4N vGRF threshold and the first vGRF peak. The maximum point was computed using Newton’s difference quotient with a central derivative approximation. The magnitude of vGRF at the HST and time of this variable were quantified.

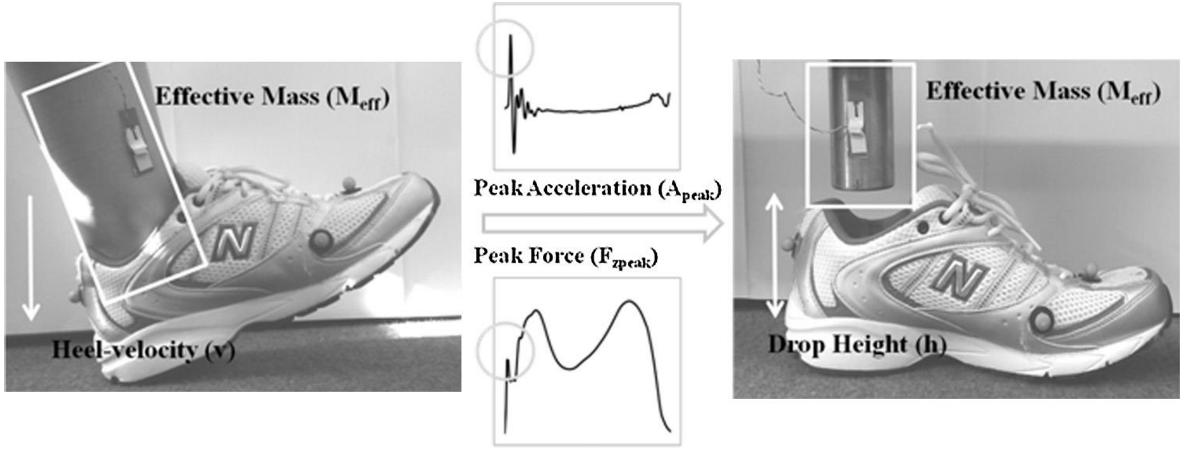
2.3. Mechanical Testing and Processing

A mechanical test device was constructed that enabled ASTM F1614 (Procedure A) to be followed and an adapted methodology to better replicate the energy in walking impacts. The device consisted of a striker of mass 8.5 kg and diameter of 45 mm, which was lifted in a shaft and dropped onto the rearfoot of the footwear approximately every 2 seconds using a metronome to guide. An accelerometer mounted on the striker and a force plate below the footwear (AMTI) collected data at 2400 Hz for drops #26-30, following 25 drops. Variables calculated were peak and time of peak vertical force and peak and time of peak acceleration averaged across the 5 impacts [21,22].

Both the adapted protocol, based on the subject's kinematic data, and the original ASTM protocol were undertaken for comparison. Drop height (Figure 3.1, Equation 1) and effective mass (Figure 3.1, Equation 2) were altered in the adapted protocol to attempt to replicate the specific energy apparent in the foot-ground contact during walking in the footwear styles. This utilised the peak acceleration and peak force to calculate effective mass and the vertical heel velocity to calculate drop height (Figure 3.1). Therefore a missile of mass 10.6-17.3 kg was dropped from a range of heights of 2-7 mm depending on the footwear type. The outcome variables measured were the same as those in the ASTM protocol.

2.4. Statistical Analysis

Statistical comparison was undertaken using SPSS (SPSS Inc., Chicago, IL, USA). Heel velocity was compared among footwear conditions in the human testing using repeated measures ANOVA with Bonferonni correction for multiple comparisons. Comparison between mechanical methods and human data was undertaken with ANOVA across the three data sets. Balanced sample sizes were produced of N= 20 (N= 5 for each of the 4 footwear conditions) through random selection such that the human sample size matched the mechanical testing. Games-Howell post-hoc test was used to identify differences due to the unequal variance in the human and mechanical data.



Equation 1. Velocity equation to match human heel velocity

$$v = \sqrt{2.g.h}$$

where v = the heel-velocity at contact and the matched velocity of the impact missile at contact, $g = 9.81$ and h = the drop height of the impact missile.

For the ASTM protocol, $v = 0.99\ m.s^{-1}$, h = 50mm (Misevich and Cavanagh, 1984)

Equation 2. Effective mass equation (Nigg, 2010).

$$M_{eff} = F_{zpeak} / A_{peak}.$$

Where F_{zpeak} peak and A_{peak} are taken from the human data collection.

For the ASTM protocol, M_{eff} = 8.5 kg (Misevich and Cavanagh, 1984)

Results

	Equation 1		Equation 2		
Condition	Heel Velocity ($m.s^{-1}$)	Drop Height (mm)	Peak A ($m.s^{-2}$)	Peak F (N)	Effective Mass (kg)
SA	-0.31	5	21.8	377	17.3
FF	-0.36	7	22	356	16.2
SH	-0.17	2	22	293	13.3
TR	-0.18	2	18.1	191	10.6

Figure 3.1 Calculation of the effective mass and drop-height from the results of the human data collection to define the methodology of the mechanical test protocol.

Equation 1 defines the drop height of the impact missile, defined by the human heel velocity. Equation 2 defines the mass of the impact missile, determined by the peak acceleration and force in walking. Results for each footwear condition tested are presented.

3. Results

The average walking speed in the current testing was $1.30 \pm 0.12\ m{\cdot}s^{-1}$ with the range spanning commonly reported walking speeds of $1.1\text{-}1.5\ m{\cdot}s^{-1}$ in similar research [4,21]. No differences in joint angles between conditions were evident at heel-strike in the sagittal plane at the ankle, knee or hip. Step length was also not statistically significantly different across conditions, with a maximum range of 2% between the trainer and barefoot conditions.

Table 3.2 Variables for the human and mechanical protocols for testing of impact characteristics (mean±1 S.D)

	Footwear Condition				
	Barefoot (BF)	**Triple-density sandal (SA)**	**Flip-flop (FF)**	**Shoe (SH)**	**Trainer (TR)**
Human					
Peak Tibial Acceleration (m.s^{-2})	40.8±16.1	21.8±8.5	22.0±9.8	22.0± 7.2	18.1±6.5
Time to Peak Tibial Acceleration (ms)	13.6±3.6	24.1±7.1	23.9±7.3	28.0±9.9	21.1± 10.7
Average HST magnitude (N)	379.5±119.7	299.7±126.1	370.6± 122.2	383.9± 133.8	181.4 ±9.2
Time to HST (ms)	12.5±3.1	27.3±0.0	25.6±2.0	27.8±2.0	19.2± 2.0
ASTM					
ASTM Peak Acceleration (m.s^{-2})	-	102.5±1.1	171.0±8.2	324.4±4.6	127.0±1.6
ASTM Time to Peak Acceleration (ms)	-	14.7±0.2	14.4±0.2	14.7±0.2	13.8±0.0
ASTM Peak Force (N)	-	798.2±7.0	1414.1±65.2	2700.4±43.8	1061.5±13.7
ASTM Time to Peak Force (ms)	-	15.4± 0.3	14.7± 0.2	15.1±0.2	13.9±0.2
Adapted					
Adapted Peak Acceleration (m.s^{-2})	-	22.4±0.7	26.1±6.0	20.7±3.6	13.9±2.5
Adapted Time to Peak Acceleration (ms)	-	5.3±1.5	6.0±2.8	10.1±1.1	4.3±0.5
Adapted Peak Force (N)	-	427.3±15.7	437.9±103.9	436.9±75.5	227.0±29.2
Adapted Time to Peak Force (ms)	-	22.2±0.54	25.8±1.8	21.2±1.3	21.2±0.5

Human data from the mean of 13 subjects and 10 data sets per shoe, ASTM and adapted mechanical protocols (mean±1 S.D of trials #26-30). Where HST = heel-strike transient

Vertical heel velocity varied across different footwear styles, despite the controlled walking velocity (Figure 3.2). The heel velocity was significantly greater towards the ground in the two footwear conditions with toe-post uppers (flip-flop and triple-density sandal; Figure 3.2) and all subjects demonstrated this pattern. The flip-flop condition was fastest, an average 0.16 (±0.03) $m \cdot s^{-1}$ faster than barefoot ($p<.001$) and over twice the velocity toward the floor recorded in the shoe (Table 3.2 and Figure 3.2).

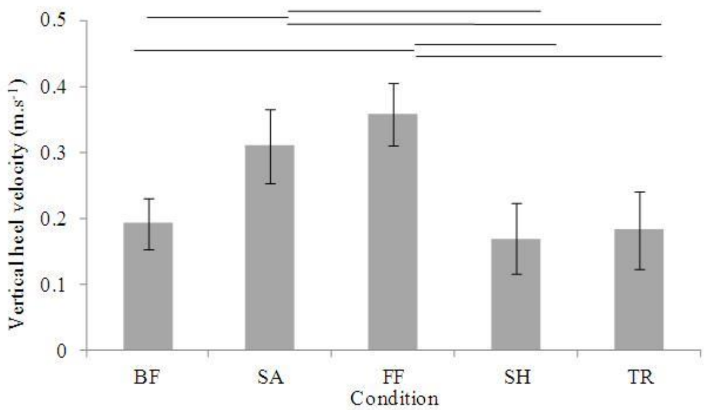
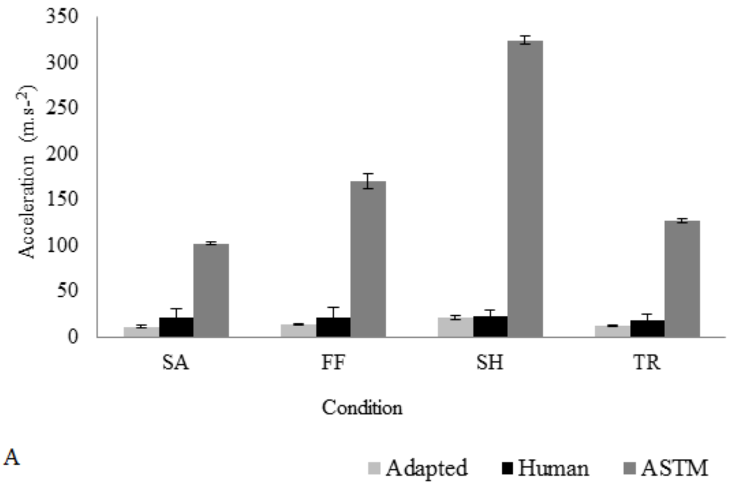


Figure 3.2 Vertical heel velocity towards the floor in the human testing for the four footwear conditions and barefoot.

Where: triple-density sandal = SA, flip-flop = FF, shoe = SH and trainer = TR and barefoot = BF. Error bars denote standard deviation across the 13 subjects tested. Horizontal lines denote statistically significant results (determined by ANOVA where p<.05).

Both the adapted protocol and standard ASTM protocol were compared to the results derived from human testing to determine which better replicated the real-life data. This identified that the ASTM data differed significantly from both the adapted and human protocols for peak acceleration magnitude ($p<.001$) and peak force ($p<.001$). The time of occurrence of these peaks differed only from the human protocol ($p<.001$), the adapted protocol time of peaks did not differ from the ASTM protocol ($p=.116$, $p=.128$). The ASTM protocol overestimated the real-life peak acceleration and forces by 755.9±431.9% and 421.1±5.4% respectively (Figure 3.3). The adapted protocol replicated the human data more closely. A mean difference in the peak acceleration scores of 12.7±17.5% ($p<.001$) and a mean overestimation of 25.2±17.7% ($p=.786$) in the peak vertical force occurred between the human and adapted protocols. The time of occurrence of these peaks was equivalent in the adapted and human protocols for the acceleration ($p=.771$), but not the force values ($p=.001$).



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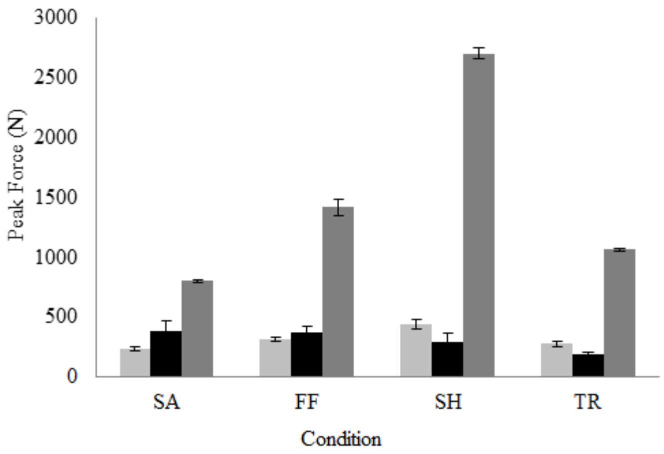

B

Figure 3.3 Comparison of variables between the two mechanical test conditions (adapted and ASTM) and the human results for the four footwear conditions

Where: triple-density sandal = SA, flip-flop = FF, shoe = SH and trainer = TR). A compares peak acceleration magnitude, B compares peak force magnitude. Error bars denote the standard deviation across the five trials (#26-30) used to compute the mean.

4. Discussion

The human testing in the present research quantified impact kinematics and kinetics at initial contact during walking in four different footwear styles and barefoot. The peak tibial acceleration values recorded in this study are consistent with the range of 19.6-78.5 $m.s^{-2}$ identified in previous walking literature [21]. Larger differences were evident between the trainer and other footwear in the present study than previous research [4]. The larger difference may be due to advances in athletic shoes design or materials from 1992 to present.

Joint angles at heel-strike did not differ significantly among conditions in the walking trials. Running literature has long-linked increases in lower limb joint angle at heel strike to increased shock attenuation, alongside decreases in peak tibial acceleration [23]. It is clear that modifying lower limb joint angles at heel strike is not the mechanism for differences in shock attenuation in the current results during walking. Despite there being no significant differences in joint angles at heel-strike, significant differences in vertical heel-velocity just prior to impact were evident among footwear conditions in this study. Vertical heel velocity was significantly faster toward the ground in the toe-post footwear conditions (triple-density sandal and flip-flop) than other shod conditions and barefoot. This finding is consistent with previous literature, which identified that footwear style affects gait kinematics and notably heel-strike velocity in walking both on a treadmill and the laboratory floor [16]. This increased heel velocity toward the floor in toe-post conditions has been previously alluded to and attributed to an adaptation in swing to ensure that the heel contacts the floor on the shoe sole [17]. Explanation of the mechanisms causing heel velocity changes in toe-post footwear would require further analysis. It is however apparent that utilising a consistent impact velocity to mechanically compare the shock absorption properties of materials for walking footwear would be misleading if the materials' intended use is a covered walking shoe as opposed to a flip-flop.

The mechanical impact results from the ASTM protocol in the present study are also consistent with previously reported values utilising the protocol on trainers (98.1-215.8 $m{\cdot}s^{-2}$, 10-22 g) and other footwear styles (147.2- 294.3 $m{\cdot}s^{-2}$, 15-30 g) [5,11,24]. As expected, the peak vertical force values of the ASTM mechanical test are substantially higher than would correspond to the HST evident in walking. The peak acceleration values are also at least 4 times greater than those recorded in vivo and not a systematic overestimation. These differences, as previously identified, relate to the ASTM mechanical methodology being designed to assess running shoe cushioning where the higher impact velocity (≈1 $m{\cdot}s^{-1}$) and energy (≈5 J) are relevant [12]. The modification to the testing protocol encompassed an alteration in impact velocity and mass to modify energy in footwear style-specific testing. The adapted protocol better replicated the walking data, identifying peak force values that did not differ significantly from the values recorded from the participants as they walked. The acceleration values better mirrored those recorded in vivo however, they still differed significantly from the human data. The standard deviation values in both the ASTM and adapted protocol are lower than the walking trials. However, the standard deviation values when compared to the peak acceleration values are high for the adapted protocol. This is due to errors introduced by the manual operation of the prototype testing device. It is anticipated that the standard deviation of peak acceleration would be reduced with further device development to enable automated operation. The calculation of mean vertical heel velocity from the walking trials and estimated lower-limb effective-mass specifically for each footwear condition better replicated the ‘real life’ findings than the general ASTM protocol. Therefore for future use this methodology would enable a more realistic comparison of the shock absorption provided by walking footwear. The nature of the device being a metallic stiff object means that any replication of timing of peak acceleration variables is more difficult due to the attenuation in the structures of the limb in walking. However the times were obviously more similar to the human testing in the adapted protocol compared with the ASTM protocol.

The effective mass calculated in the present study was based on the peak force and peak acceleration values from the gait testing, as recommended for running impacts [25]. The use of this formula for walking data is hindered by the two variables occurring at slightly different times and the double-limb stance influencing force parameters. Effective mass has been reported in existing literature as values ranging from 1.6 kg in walking [26] to 11.6 kg [27] to 20% of body weight [28]. These are lower than the current study (10.6-17.3kg); however most of the literature does report this effective mass as higher in walking than running, due to the more extended limb at initial-contact [29,30]. The effective mass in this proposed adapted methodology could be both footwear-, participant- and gait-style specific to account for specific wearers’ anthropometry and kinematics. The range of walking footwear users demonstrates a variety of footwear upper styles, body masses, heel dimensions, walking velocities, and kinematics. This population spans children to the elderly and also includes symptomatic gaits. The running population may be less diverse in terms of body mass and kinematics making a single, standardised protocol more relevant, however differences will still exists dependent on touchdown kinematics, midsole hardness and running velocity [31].

Further data collection could establish specific estimations of the relevant effective mass and heel velocity for a range of footwear styles, such that the footwear technician can test the footwear using standardised/published drop heights and impacter mass and contact area dependent on the purpose of the footwear e.g. walking/running, the style of the footwear e.g. toe-post/covered and the characteristics of the wearer e.g. obese/children. This would provide useful information for a footwear technician to gain relevant shock absorption values from modifying bench-top tests without access to a gait laboratory. Testing a range of footwear sizes and styles would mean that the results are not dependent on one shoe sample from the production line as they are in the current study.

4.1. Conclusions

The success of this protocol at more closely replicating the human data collected in this research identifies that footwear testing bodies such as SATRA should manipulate current protocols to increase the relevance to the real-life use of the footwear that they are testing. Applying more realistic loads over relevant contact areas and time periods, by using heel velocities and effective masses relevant for the gait mode and footwear style, offers the opportunity for more relevant assessments of shock absorption properties. Heel velocity is altered when walking in different footwear styles so mechanical impact measures cannot be kept constant and considered accurate representations of shock during walking gait. It is recognised that any future work needs to account for the double-support period in walking as opposed to the single-support in running in previous impact research.

5. Conflict of Interest Statement

The research has been funded by a research contract between University of Salford and FitFlop ltd. The primary author works on the project, supervised by Dr Jones and Dr Graham-Smith. Dr G. Cooper is not associated with the research project. Work was undertaken with scientific diligence, data was collected, analysed and the paper written with no influence from the funding company.

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3.2.2 The manipulation of midsole properties to alter impact characteristics in walking.

Price, C, Cooper, G & Jones, R. (2014).

Footwear Science.In Review.

Abstract

The midsole of footwear can provide an opportunity to attenuate the impact at the foot-ground interface. The present study was undertaken to quantify impact in walking in different footwear midsoles, comparing footwear thickness and hardness variations. *Methods:* Footbed thickness (28-41 mm) and hardness (30-55 Shore A) were varied independently in 7 sandals. Thirteen subjects walked in the footwear variations on a level walkway in the gait laboratory as lower limb kinematics, vertical ground reaction force and peak positive axial tibial acceleration were quantified. Peak magnitude and time of the acceleration were quantified and the heel-strike transient was characterised for comparison between conditions with a repeated-measures ANOVA. Thickness and hardness variations were also compared using a drop-test protocol to replicate walking. *Results:* Lower limb joint angles did not vary at heel-strike, however, a faster vertical heel-velocity was recorded in the softer midsoles (e.g. 55 Shore A = -0.294±0.055, 30 Shore A= -0.328±0.052, p<.001). Varying the hardness of the midsoles also significantly altered tibial acceleration and force variables, however limited significant differences existed between the thickness variations in walking. Increasing the hardness of the heel section of the footwear increased the peak positive axial tibial acceleration values, for example increasing Shore A from 30 to 40 resulted in a 35% increase in this variable. Concurrently, the occurrence of heel-strike transients increased from 5.8% in the 30 Shore A condition to 22.5%, 46.7% and 71.7% of all trials in the 40, 47 and 55 Shore A conditions respectively. The drop-test protocol replicated the differences evident in the walking protocol despite magnitudes being elevated. *Conclusion:* Modifying midsole properties of footwear, particularly hardness, alters the gait kinematics and the shock experienced by the wearer in walking. This may pose benefits in terms of comfort and reduction in loading to the lower limb, however the influence on foot motion at initial contact and footwear longevity should be further quantified.

Key words: shock, footwear, heel-strike transient, accelerometer, material properties.

1. Introduction

Lower limb musculo-skeletal loading in gait begins with the transmission of stress waves at heel-strike. Part of this loading process produces a heel-strike transient (HST), which has been linked to degenerative changes to tissue (Radin et al., 1991), clinical symptoms (Voloshin and Wosk, 1982) and comfort in walking (Whittle et al., 1994). The midsole of a shoe provides an opportunity to apply a visco-elastic material between the foot-ground interface to reduce the energy transferred at heel contact and the transient (Pratt et al., 1986; Whittle, 1999). Choices of material (including hardness) and shape (including thickness) are constrained by the design specification of the footwear. Design specification restrictions include purpose/activity type, the target market, manufacturing considerations and cost. For decades, athletic footwear companies have manipulated midsole-heel properties in order to assess the effect of hardness and thickness of Ethylene Vinyl Acetate (EVA) constructions in footbeds to provide an effective combination for the comfort and protection of the runner (Frederick et al., 1984; Hamill et al., 2011; Milani et al., 1997).

Research has been undertaken to quantify changes in midsole hardness and the effect on variables quantify impact in running and mechanical protocols which replicate running (Frederick et al., 1984; Nigg et al., 1987). Researchers report increased positive peak positive axial tibial acceleration values in impact assemblies and maximum loading rate of the impact peak in running with increased hardness of footwear (DeWit et al., 1995; Sterzing et al., 2013). Other authors identify that there are no differences in the magnitude of the impact peak of the vertical ground reaction force or the maximum loading rate in running with alterations of hardness, which they attribute to adaptations to eversion at initial contact (Nigg et al., 1987). Similarly, increasing the thickness of the heel section of a running shoe has been demonstrated to reduce peak positive axial tibial acceleration and maximum loading rate of the vertical ground reaction force in human (Heidenfelder et al., 2010; TenBroek et al., 2013) and mechanical (Frederick et al., 1984) protocols. It is therefore apparent that manipulating midsole thickness and hardness can alter impact characteristics in both human test protocols and mechanical protocols which aim to replicate running. These alterations can include potentially positive outcomes for wearers such as reduced lower limb loading (Hamill et al., 2011; TenBroek et al., 2013) and reduced sensations of impact severity (Lake and Lafortune 1998).

Athletic footwear has provided the basis for most recent work into footbed construction, with walking studies limited to orthotic interventions as opposed to modifications to footwear itself (Healy et al., 2010; Pratt et al., 1986). The impact with the floor in running is defined by a heel velocity of approximately 1 $m{\cdot}s^{-1}$ and effective mass of 8.5 kg (Misevich and Cavanagh, 1984). In walking the comparable variables are 0.17-0.36 $m{\cdot}s^{-1}$ and 1.6-17.0 kg identifying different kinematics and loading magnitudes and rates of the lower limb (Jørgensen and Bojsen-Møller, 1989; Jefferson et al., 1990; Price et al., 2014). Consistently, recent data has identified that the mechanical protocol utilised to quantify the shock absorption properties in athletic footwear over-estimates the peak acceleration and HST magnitude and underestimates the timing of these variables in walking footwear (Price et al., 2014). The importance of this discrepancy is enhanced by the shock absorption characteristics of viscoelastic materials being rate dependent (Whittle, 1999). Thus gait specific testing is required to establish the suitability of walking footwear in protocols specific to their ‘real-world’ wear.

The different design of running footwear compared to some styles of walking footwear with lace-up, covered uppers, reinforced counters and rubber midsoles and outsoles is evident. It has recently been demonstrated that different styles of footwear upper result in altered heel-velocities and effective masses at touchdown during walking (Price et al., 2014). In addition to the aforementioned loading characteristics, these factors combine to indicate that findings and recommendations from running studies cannot be inferred to research and development of walking footwear. Despite the focus of research literature on running, walking is a more relevant activity to the general and clinical populations. Modifying footwear based on walking gait may enable increased comfort and reduced clinical symptoms in these populations (Voloshin and Wosk, 1982; Whittle et al., 1994). Recently the health footwear market has developed and expanded. This market can feasibly accommodate changes in footbed thickness and materials in designs as long as benefits can be justified to consumers. The study of these thickness and hardness alterations is therefore warranted with test protocols that include walking as opposed to running protocols to infer walking footwear design.

The primary aim of the study was to quantify the effects of differing midsole hardness and thickness on impact variables in footwear tests during walking and in a mechanical protocol to replicate walking. It is expected that increasing footbed thickness and decreasing hardness would reduce peak acceleration and forces in walking protocols due to the provision of a longer time to apply force and a more viscoelastic material to absorb more energy from the touchdown (Whittle, 1999).

2. Methods

Ethical approval for the study was achieved through the University ethics committee; volunteers were recruited from the staff and student populations.

2.1 Footwear Tested

Seven footwear conditions were tested with varying midsole depths and hardness in a flip-flop upper (Table 3.3) using mechanical and human methodologies. The shoes were varied only in the heel characteristics, all other shoe features were consistent (upper/pitch/outsole/profile etc). Due to constraints in manufacture, the upper differed between the hardness and thickness shoes, but was consistent within them. The thickness variations had a toe-post upper, the hardness a sandal upper with no back-strap.

Table 3.3 Footwear characteristics for the seven footwear conditions tested in the study, all of which had a sandal upper and an EVA construction.

Condition		Heel Depth (mm)	Heel Hardness (Shore A)
Thickness Variations	T41	41	40
	T35	35	40
	T28	28	40
Hardness Variations	H55	41	55
	H47	41	47
	H40	41	40
	H30	41	30

Where Shore A hardness was measured within the factory and the University with a durometer and a bespoke device, which is utilised for quality control and implements a larger base to contact the test specimen.

2.2 Protocol

Thirteen healthy subjects (2 males, 11 females, 27.5±8.8 years, 62.0±10.3 kg, 1.65±0.05 metres, mean±1 S.D) with shoe size U.K. 6 gave their consent and participated in the study. Subjects reported no lower limb injury in order to take part in the study and were instrumented with a lower limb marker setup for 3-D motion capture and one uni-axial accelerometer.

A 10 camera Qualisys Pro-Reflex system (Qualisys, Sävebalden, Sweden) was used to track 3D motion at 240 Hz. Spherical retro-reflective markers and clusters were positioned to define the lower limbs in accordance with the CAST technique (Cappozzo et al., 1995). The foot was defined with markers on the posterior calcaneus and the dorsal aspects of the 1st, 2nd and 5th metatarsal heads. The shank was defined with anatomical markers on the medial and lateral malleoli and the medial and lateral knee, with a rigid plate of four tracking markers on the anterior tibia. The accelerometer was mounted on the right anterior-medial tibia above the medial malleolus on a small piece of light flexible plastic. It was positioned 5-10 cm above the malleolus, on an area with least adipose tissue, oriented with the tibia axis. The accelerometer was affixed with double-sided tape and an elasticated bandage secured it tightly without causing discomfort. The accelerometer was sampled alongside 2 force plates (AMTI, Advanced Mechanical Technology, Watertown, USA) at 2400 Hz collecting ground reaction force data for two consecutive right heel-strikes. Subjects performed 5 trials in each condition in a randomised order following a familiarisation period of 4 practice walks, data from the right leg only was utilised. Ten data-sets for each footwear condition were analysed. Participants walked at a self-selected velocity for the first condition which was then monitored with timing gates to ensure consistent walking speeds within a range of ±5%, trials outside this boundary were re-captured.

The drop-test methodology has previously been described and utilised a protocol which replicated the energy of the shoe-ground impact in walking (Price et al., 2014). The footwear conditions were impacted with a mass of 17 kg from a drop height of 5 mm to replicate the impact characteristics evident in this style of footwear during walking (Price et al., 2014). This compares to the 8.5 kg and 50 mm utilised in the standard ASTM protocol F1614 (Procedure A).

2.3 Data Processing

Data was processed and analysed using Visual 3D (C-Motion, Inc., Rockville, MD, USA), defining the right limb and pelvis as 4 rigid segments. 3D motion (10 Hz) and accelerometer (100 Hz) data was filtered using low-pass Butterworth filters. Ground reaction force data was not filtered due to findings from Gillespie and Dickey (2003). Force plate contact was defined as the first frame in which the vertical ground reaction force (vGRF) exceeded 4N. Joint angles at heel-strike for the sagittal plane at the ankle, knee and hip were computed for the concurrent frame for which force plate contact was defined. Heel-marker vertical velocities at heel-strike were calculated using the mean value from 8ms leading up to heel-strike, which is within ranges found to be reliable in previous research (Karst et al., 1999). Heel-strike transient (HST) of the vertical GRF was defined as a local maximum point between the 4 N vGRF threshold and the first vGRF peak. This was computed using Newton's difference quotient with a central derivative approximation, to identify zero gradient of the vGRF. The magnitude of vGRF at the HST and time of this variable were quantified. Maximum instantaneous loading rate of the vGRF was computed for all trials from the difference quotient. Magnitude and timing of peak positive axial tibial acceleration was also calculated and used to compute the rate to peak positive axial tibial acceleration. Temporal-spatial data (including step length and stance time) was calculated automatically and output for comparison.

Statistical comparison was undertaken between hardness and thickness variations in SPSS (SPSS Inc., Chicago), using ANOVA with Bonferonni correction for multiple comparisons (p value<0.05). The number of HST in each condition was compared statistically prior to conversion to percentages of total trials for presentation and HST data was not compared statistically due to inconsistent and small N numbers.

3. Results

The comparison of kinematic variables in walking identified no significant differences between thickness or hardness variations in lower limb sagittal plane joint angles at heel-strike, or temporal-spatial characteristics (Table 3.4). Vertical heel velocities at heel-strike differed between the hardness conditions, decreasing as the hardness of the footwear heel section increased (Table 3.4). No differences were evident in this variable in the thickness variations.

Table 3.4 Kinematic data from walking in different hardness and thickness variations

Variables	Footwear Condition: Thickness			Hardness				Significant *p* values	
	T41	**T35**	**T28**	**H55**	**H47**	**H40**	**H30**	**Thickness**	**Hardness**
Ankle (°)	4.7±4.1	4.0±3.8	4.3±3.1	4.0±3.7	4.2±3.9	3.8±3.6	4.0±4.1	-	-
Knee (°)	-0.5± 4.0	-0.5± 3.8	-1.3± 3.5	-0.4± 5.4	-0.7± 4.9	0.1± 5.7	-1.3± 4.2	-	-
Hip (°)	25.7±6.5	25.7±6.4	25.4±7.0	23.5±6.9	24.1±6.3	23.8±6.2	24.7±7.4	-	-
Vertical Heel Velocity ($m \cdot s^{-1}$)	-0.358±0.055	-0.376±0.065	-0.378±0.057	-0.294±0.055	-0.292±0.055	-0.315±0.049	-0.328±0.052	-	H55<H40 p = .003 H55<H30 ≤ .001 H47<H30 ≤ .001 H47<H40 p = .027 H40<H30 p = .009

Where T41= 41 mm, T34 = 34 mm and T28 = 28 mm of heel depth and H55= 55 Shore A, H47 = 47 Shore A, H40 = 40 Shore A and H30 = 30 Shore A hardness in the heel section. Sagittal plane joint angles for the ankle, knee and hip and vertical heel velocity at heel-stike (mean± standard deviation). Statistically significant (ANOVA $p < 0.05$) p values are presented.

3.1 Thickness

Analysis of HST identified the feature occurred in 46.9% of the thickness variation trials collected and did not significantly vary between conditions (Table 3.5). Analysis of the HST magnitude demonstrated an increase in magnitude of HST with decreasing footbed thickness (Table 3.5). Consistent with the force variable, peak positive axial tibial accelerations displayed a trend to increase with decreasing thickness. However, no significant differences were evident in human acceleration variables between thickness conditions, despite the T28 condition producing a 10.3% increase in peak positive axial tibial acceleration compared to T41. The only significant difference between the thickness conditions in the human data was that loading rate in the thinnest condition (T28) was higher than in T35 (Table 3.5). The drop-test protocol identified significantly lower peak acceleration and force in the thinnest condition (Table 3.5).

3.2 Hardness

Analysis of walking in the hardness conditions, demonstrated the HST feature occurred in 37.5% of the trials (Table 3.6). The H30 condition (the softest EVA tested) produced HST in a total of 8 trials from 4 participants, in contrast walking in the H55 condition resulted in a HST in 71.7% of all trials and only 3 participants did not demonstrate HST in this condition. The magnitude of the HST increased and the feature occurred a shorter duration from heel contact following alterations in footbed hardness, although these variables were not explored statistically (Table 3.6). The maximum instantaneous loading rate also reflected this trend and decreased with reduced hardness. Although, despite a 5.7 $kN \cdot s^{-1}$ change, this variable did not significantly differ between the H40 and H30 conditions. Peak positive axial tibial accelerations increased as hardness of the footbed increased. The magnitude of peak positive axial tibial acceleration reduced by 12.8% in H55 compared to H47 and by 28.1% and 46.8% respectively in H55 in comparison to H40 and H30. The time of peak positive axial tibial acceleration was later with softer EVA, therefore rate to peak positive axial tibial acceleration also significantly increased as hardness decreased (Table 3.6). The drop-test protocol demonstrated significant decreases in both peak force and acceleration with reducing hardness until H30, for which magnitudes increased compared to H40.

Table 3.5 Heel-strike transient and peak positive axial tibial acceleration variables for thickness variations.

Variables		Thickness Condition			
		T41	T35	T28	Significant *p* values
Human Testing	Percentage of all trials with HST (%)	43.3	51.7	45.8	-
	HST magnitude (N)	305.8±113.3	332.4±155.8	366.7±117.4	NA
	HST time (ms)	31.9±.04	32.5±0.4	32.3±0.6	NA
	Maximum Instantaneous Loading Rate ($kN \cdot s^{-1}$)	22.0±7.0	21.8±7.1	23.9±8.6	T28>T35 p = .038
	Peak Positive Axial Tibial Acceleration ($m \cdot s^{-2}$)	17.4±8.4	18.1±8.9	21.0±10.6	-
	Time of Peak Positive Axial Tibial Acceleration (ms)	25.4±6.9	24.0±5.8	25.4±10.7	-
	Rate to Peak Positive Axial Tibial Acceleration ($m \cdot s^{-3}$)	692.2±336.3	745.0±363.3	858.8±500.2	-
Mechanical Impact tester	Peak Acceleration ($m.s^{-2}$)	32.9±2.6	33.7±1.5	34.9±0.7	-
	Peak Force (N)	669.7±29.3	700.1±20.5	687.8±8.3	T41<T35 p = .037

Where T41= 41 mm, T34 = 34 mm and T28 = 28 mm of heel depth. Data is presented as mean± standard deviation. Statistically significant (ANOVA $p < 0.05$) p values are presented. HST magnitude and time are presented for the trials that included a HST only.

Table 3.6 Heel-strike transient and peak positive axial tibial acceleration variables for hardness variations.

Variables		Hardness Condition				
		H55	**H47**	**H40**	**H30**	**Significant *p* values**
Human Testing	Percentage of all trials with HST (%)	71.7	46.7	22.5	6.7	H55>H40 p = .027 H55>H30 p ≤ .001 H47>H40 p = .014 H47>H30 p = .009
	HST magnitude (N)	366.7±126.6	375.1±176.6	395.4±177.3	443.5±189.1	NA
	HST time (ms)	26.5±0.4	32.8±0.9	37.8±0.6	42.6±0.7	NA
	Maximum Instantaneous Loading Rate ($kN \cdot s^{-1}$)	36.7±8.9	30.9±5.2	26.8±3.7	21.1±7.2	H55>H47 p = .042 H55>H40 p = .002 H55>H30 p ≤ .001 H47>H40 p = .002 H47>H30 p = .002
	Peak Positive Axial Tibial Acceleration ($m \cdot s^{-2}$)	23.5±9.2	20.5±7.9	16.9±4.5	12.5±3.2	H55>H40 p = .008 H55>H30 p = .001 H47>H40 p = .039 H47>H30 p = .003
	Time of Peak Positive Axial Tibial Acceleration (ms)	19.4±5.3	21.0±7.7	24.4±8.3	26.7±9.2	H55<H40 p = .046 H47<H40 p = .036
	Rate to Peak Positive Axial Tibial Acceleration ($m \cdot s^{-3}$)	1165.2±436.4	961.8±378.1	697.0±275.9	495.9±198.9	H55>H40 p ≤ .001 H55>H30 p ≤ .001 H47>H40 p ≤ .001 H47>H30 p ≤ .001 H40>H30 p = .002
Mech. Impact tester	Peak Acceleration ($m.s^{-2}$)	33.4±2.2	25.2±0.7	20.9±1.0	24.9±0.6	H55>H47 p = .010 H55>H40 p ≤ .001 H55>H30 p ≤ .001 H47>H40 p = .005 H40<H30 p = .011
	Peak Force (N)	688.2±19.8	549±10.0	470.0±10.0	552.5±6.9	H55>H47 p = .003 H55>H40 p ≤ .001 H55>H30 p ≤ .001 H47>H40 p ≤ .001 H40<H30 p ≤ .001

Where H55= 55 Shore A, H47 = 47 Shore A, H40 = 40 Shore A and H30 = 30 Shore A hardness in the heel section. Data is presented as mean±standard deviation. Statistically significant (ANOVA $p < 0.05$) p values are presented. HST magnitude and time are presented for the trials that included a HST only.

4. Discussion

The aim of the study was to quantify the effects of differing midsole hardness and thickness on shock absorption variables in walking footwear. Therefore other aspects of the footwear including outsole shape and upper characteristics were not varied. The study identified significant differences between thickness and hardness midsole variations when being assessed for shock absorption using both human and mechanical protocols.

The temporal-spatial and kinematic data comparison identified limited significant differences within the thickness and hardness variations. The hardness variations recorded a lower vertical heel velocity towards the floor than the thickness variations (e.g. H55 -0.294±0.055 v T35 -0.376±0.065), likely due to the differing uppers (Price et al., 2014). Also within the hardness variations the participants' heel velocity was systematically faster in the softest conditions after a hardness of 47 Shore A. These results demonstrate to a footwear designer or technologist that, within the hardness and thicknesses ranges tested in this study and population in this research, modifying hardness alters heel contact velocity, but modifying thickness does not. This means that if a footwear designer is to change the footbed hardness of walking footwear they must consider how this influences the velocity at heel-strike when considering aspects such as shock absorption, comfort and product longevity. Despite not influencing vertical heel-velocity at touchdown in this study, it is probable that modifications to footbed thickness may alter kinematics in terms of swing characteristics within footwear due to the demands of toe-clearance (Menant et al., 2009). Kersting and Brüggemann (2006) identified minimal and non-significant variations in the touchdown velocity of the malleoulus in trainers with differing midsole hardness (45-61 Shore C) in running, consistent with Nigg et al. (1987) in running shoes of 25-45 Shore A. Despite the changes in heel velocity apparent in the present research, no significant differences were evident in lower limb sagittal plane joint angles at heel-strike within the hardness (e.g. H55 v H47) or thickness (e.g. T41 v T28) variations in walking. Previous research has identified significant kinematic adaptations to knee flexion to mediate the stiffness of the limb and reduce impact energy, however this work is in running where limiting the maximum forces due to impact in the system may be more essential than in walking (impact forces may not exceed these limits in walking gait).

4.1 Thickness

It was hypothesised that decreasing the thickness of the footbed would increase the occurrence and magnitude of the HST and the magnitude of the peak positive axial tibial acceleration. The HST is caused by the force-time characteristics of the impact as the foot/shoe strikes the ground and is measured by the force plate. A stress wave from this impact travels proximally through the foot and into the limb. The magnitude of the force evident can be reduced by viscoelastic footbed material. The dissipation will be proportional to the damping coefficient of the material and the amount of material it travels through, hence thicker midsoles will reduce the magnitude of the HST. This is consistent with previous research in running footwear where increased peak acceleration values and a trend for increased force loading rate were evident in thicker footbeds (Hamill et al., 2011; TenBroek et al., 2013).

Both HST and peak positive axial tibial acceleration in the current research reduced with increasing midsole thickness, however differences were not statistically significant in the human walking data. . The drop-test protocol also largely failed to differentiate between the thickness variations tested. Maximum instantaneous loading rate of the vGRF was significantly higher in T28 than T35. These results suggest that potentially reducing an item of footwear with this construction from 41 to 35 mm in the heel may not results in any evident reduction in shock absorption properties, however further reductions may be detrimental. It may be apparent that the additional 13mm of EVA may be redundant in terms of shock absorption capacity for walking footwear. It is an example as to why other factors such as longevity of the foam at different thicknesses would also need to be considered in design. Thicker foam in a walking shoe may absorb slightly more shock and last longer, however the cost of manufacture and distribution is increased so the specific product requirements should be considered.

The identification of significant differences between conditions may have been limited by a large range in individual response, which resulted in large deviations about the mean values for the variables (for example standard deviations for peak acceleration were 8.4-10.6 $m{\cdot}s^{-2}$ and HST transient 113.3-155.8 N). A greater range of thicknesses may have identified further differences and also been more generalisable to the wider walking footwear market. Also the thinnest condition (28 mm) is also relatively thick for an EVA footbed in a walking shoe, but relevant to ‘health and well-being’ footwear.

4.2 Hardness

Consistent with the study hypothesis, the variations in hardness of the footbeds in the current study produced significantly lower peak axial tibial accelerations and reduced loading rates in softer footbeds. Also, the occurrence of HST reduced and the HST occurred later from heel contact in softer footbeds. The reduced occurrence is consistent with a reduced transmission of energy from impact in softer soled footwear. Less viscoelastic footbed materials, due primarily to reduced viscosity, absorb less energy such that recorded force is higher. Meaning that the magnitude of the HST is proportional to the viscoelasticity of the midsole when the thickness of the sole is un-changed. As the behaviour of the viscous component is rate-dependent it is essential that the rate and conditions of the loading reflect the intended use of the footwear, therefore data from running tests is not suitable to explain the response of footbeds in walking shoes. Contrasting this expectation, the peak positive axial tibial acceleration did not differ significantly between the two hardest and two softest conditions respectively. Similar to the thickness results, this identifies that footbed modification within certain ranges result in negligible alterations to the loading experienced by the wearer in walking.

The HST magnitude increased with decreasing hardness, which may be a function of the individual participant response. As the conditions became softer, fewer participants had evident HST which meant that the mean values were more heavily weighted toward participants with more severe HST. Similarly, Nigg et al. (1987) identified no difference in maximum force between hard (45 Shore A) and soft (25 Shore A) conditions. This was attributed to changes in initial eversion patterns. Further analysis of motion data in the current work would be required to determine adjustments are apparent in the present work, however this is beyond the scope of the present comparison. Contrasting the work by Nigg et al. (1987), in the current study the loading rates however did decrease as hardness decreased, consistent with other previous research in running (DeWit et al., 1995). This supports the suggestion by Hennig (2011) that the force loading rate is the most representative variable when considering the shock absorption properties of footwear in vivo and particularly due to the data analysis process implemented in the current research for HST variables.

The drop-test results reduced progressively with decreasing hardness until the softest condition where the peak acceleration and peak force variables increased to a level consistent with the H47 condition. This may be an indication of the material bottoming-out in response to the load applied and the rate of loading. As this was not evident in the walking data it is also an indication that, despite the modification, the mechanical testing methodology does not accurately represent the loading evident in these participants.

4.3 Limitations

Individual subject variability in the current study may have affected the HST magnitude, as the HST feature is not evident in all subjects for all conditions, so the mean data is influenced by which individual subject recorded a transient in each condition and variability between them. In running the first peak in vGRF is a feature apparent in all runners (Cavanagh and Lafortune, 1980) and therefore mean data between conditions includes all test subjects. Limitations are apparent in the present study, particularly the high vertical heel velocity and kinematics in the footwear tested due to the sandal upper means that the results may not be transferrable to all footwear styles and uppers (Lake and Robinson, 2005; Price et al., 2014; Shroyer and Weimar, 2010). The lack of testing of the interaction of material hardness and thickness also limits the application of results as footwear technologists are likely to manipulate thickness and hardness of EVA in combination as opposed to in isolation. Further work to quantify the influence of the thickness and hardness variations on foot motion and durability of footwear in walking is recommended.

4.4 Conclusions

The present study highlights that adaptations of footbed properties of daily walking footwear can significantly alter the impact characteristics experienced by the wearer. This study points to softer footbeds offering advantages in shock absorption, however their impact on motion at heel-strike as well as the longevity of softer foams should be considered prior to their recommendation for use in walking footwear manufacture. The differences evident in the thickness of the footbeds identified minimal differences in the shock absorption capability of 28-41 mm thick EVA footbeds in walking. The range of thicknesses employed in this study did not alter gait kinematics at heel-strike, however the alterations in hardness instigated altered heel contact velocity, which has implications for footwear design. Future work should determine the meaning of the magnitude of variables in terms of comfort or injury and potentially a recommended threshold for shock absorption properties in walking footwear.

Conflict of interest statement

The primary author of the paper worked on a research project which was funded by FitFlop ltd, supervised by Dr Jones. Dr G. Cooper was not funded by the research project. Work was undertaken with scientific diligence, data was collected, analysed and the paper written with no influence from the funding company.

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3.2.3 The effect of unstable sandals on single-leg standing.

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Abstract

Purpose: Unstable footwear lacks peer-review published research to support concepts and claims. The present study was therefore undertaken to quantify and compare the effect of commercially available unstable sandals on single-leg balance in a healthy female population. *Methods:* Fifteen participants stood on their right-leg in one control sandal (Earth) and four sandals that are marketed as unstable footwear (FitFlop, Masai Barefoot Technology, Reebok Easy-Tone and Skechers Tone-Ups). Centre of pressure trajectory, lower limb kinematics and lower limb muscle activation was recorded as participants undertook three 30 second trials in each sandal. *Results:* The unstable sandals altered parameters related to stability in participants. Namely Masai Barefoot Technology increased centre of pressure range in the anterior-posterior direction and concurrently increased sagittal ankle motion. Reebok EasyTone had a similar effect in the coronal plane at the ankle. Muscle activation increased in the unstable sandals, with significant differences apparent in the medial gastrocnemius, soleus and rectus femoris, predominantly in Masai Barefoot Technology. Findings were attributed to the large rocker sole on the Masai Barefoot Technology sandal and more subtle outsole designs in the other sandals. *Conclusions:* Overall minimal differences from the control sandal were evident and it is expected that dynamic tasks may elicit greater differences in stability. The instability imposed by the sandals is design-specific and consideration should be given to this when the footwear is recommended to specific individuals.

Keywords: footwear, centre of pressure, outsole, electromyography, postural sway.

1. Introduction

Instability in gait and standing is a concept that has been considered in clinical groups (Geuze, 2003), the elderly (Hijmans *et al.*, 2007) and in those rehabilitating from surgery (Harrison *et al.*, 1994). In these populations stability poses a particular problem and increasing stability to reduce risk of injury is the ultimate aim of research. In contrast, the concept of unstable footwear is to deliberately reduce stability in the wearer in order to increase muscle activation. If the footwear does increase muscle activation then it may be a useful tool for rehabilitation or in at-risk groups to increase stability (Nigg *et al.*, 2006a; Kaelin *et al.*, 2011). Recently unstable footwear has become increasingly popular and a wide range of styles incorporating different designs are available. Despite this there are few published studies quantifying and comparing the effects of these shoes on gait or standing to substantiate technologies.

Protocols to assess instability utilise walking, gait initiation and termination and double- and single-limb standing with eyes open and closed (Prieto *et al.*, 1996; Hatton *et al.*, 2009; Landry *et al.*, 2010). Single-leg standing is a widely used method to quantify postural instability, particularly in rehabilitation (Hadian *et al.*, 2008) and measured variables include centre of pressure (CoP) displacement, derivative and area quantities (Murray *et al.*, 1975; Raymakers *et al.*, 2005), muscle activation (Landry *et al.*, 2010) and lower limb kinematics (Prieto *et al.*, 1996). Generally CoP variables have been shown to be reliable for standing methodologies, particularly in younger populations (Santos *et al.*, 2008), with appropriate length trials and sampling frequencies (Lafond *et al.*, 2004; Raymakers *et al.*, 2005). Lower extremity joint angles were also quantified as previous studies have identified altered kinematics in the sagittal ankle angle in unstable footwear in standing (New *et al*., 2007) and walking (Nigg *et al*., 2006b). Previous walking studies have identified no differences between unstable footwear conditions or compared to a stable control (Elkjær *et al*., 2011; Porcari *et al.*, 2010). Therefore utilising a single-leg standing protocol may be an effective method to quantify and compare the effects of unstable footwear.

Previous research in unstable footwear has predominantly focused on Masai Barefoot Technology (MBT) (Nigg *et al.*, 2006b; Landry *et al.*, 2010). The rocker-soled footwear has extensive peer-reviewed investigation, with both independent and commissioned studies. Results demonstrate increased CoP range in the anterior-posterior direction and mean CoP velocity in walking (Buchecker *et al.*, 2012) and standing (Nigg *et al.*, 2006b; Landry *et al.*, 2010). The technology of the anterior-posterior rocker sole is derived from a clinical tool used to aid progression and reduce plantar pressure in clinical groups (Hutchins *et al.*, 2009). Other unstable footwear brands, such as Reebok, Skechers and FitFlop utilise different technologies which lack the publication record of MBT. In a study funded by Puma, Germano (2011) assessed four unstable footwear conditions against barefoot and a standard trainer during single-leg standing. CoP trajectory was recorded using an in-shoe pressure system and electromyography (EMG) of eight leg muscles was recorded. Increases in Integrated EMG (IEMG) and CoP excursion were evident in barefoot, however no differences existed between the unstable conditions or compared to the stable control trainer. The brands and technologies tested were not named or described in this study (Germano *et al.*, 2012). Therefore it is not clear specifically what footwear features were tested and found to be equally as stable as a trainer.

The present study was undertaken to quantify the effect of four commercially available unstable sandals on stability in single-leg standing. Variables deemed particularly relevant were lower limb joint angles, muscle activation, and CoP characteristics. It is hypothesised that the unstable shoes will decrease stability compared to the control, thus increasing EMG and CoP excursion. Apparent instability is expected to be design-specific with medial-lateral and anterior-posterior differences evident based on the sole design of each sandal.

2. Methods

2.1. Participants

Fifteen females participated in this study. The mean±1 SD age was 29±6.7 years, mass 62.6±6.9 kg, height 167.1±4.2 cm and shoe size 5 or 6. The study was approved by the University ethics committee and written informed consent obtained prior to participation. All participants reported themselves as in good health and with no recent lower limb injury prior to taking part in the study. Participants were not regular wearers of unstable footwear.

2.2. Sandal conditions

The control footwear tested was the Earth sandal (CO) alongside four unstable sandal conditions (Table 3.7; Figure 3.4). This was chosen as it is an alternative sandal which makes no claims regarding instability and is aimed at a similar population. Barefoot was not used as a control due to large decreases in stability apparent in previous studies (Germano, 2011).

Table 3.7 Footwear condition characteristics (size 6)

Sandal	Abbreviation	Mass (g)	Description
Earth Kalso	CO	193	3.7° incline in footbed from heel to toe with firm sole and flip flop upper.
FitFlop Walkstar	FF	187	Multi-density ethylene vinyl acetate (EVA) midsole incorporating high-density heel, low-density midfoot and a mid-density forefoot.
Masai Barefoot Technology Kisumu	MB	534	Thermoplastic polyurethane (TPU) heel and midfoot, compressible heel. Rocker-sole in the anterior-posterior direction.
Reebok Easy-Tone	RE	250	Air-filled compressible elliptical pods positioned under the heel and forefoot, which allow air to travel between the two.
Skechers Tone-Ups	SK	195	Multi-density polymer midsole with firm forefoot.

Figure 3.4 Footwear conditions left to right, Control (CO), FitFlop (FF), Masai Barefoot Technology (MB), Reebok (RE) and Skechers (SK).

2.3. Protocol

The protocol consisted of three-dimensional motion analysis, CoP excursions and lower extremity muscle activity measured for single-leg standing in the five sandal conditions. Sandal condition was randomised and participants were allowed a short familiarisation period in each sandal prior to testing. The participants were instructed to step directly onto the force plate with their right leg, place their hands on their hips and gain balance. Participants fixed their sight on a visual anchor, 2 metres away, at eye level. Three trials of 30 seconds were collected for each sandal condition (Pinsault & Vuillerme, 2009).

2.3.1 Kinematics

Three-dimensional kinematic data were collected using a sixteen camera motion capture system (OQUS, Qualisys, Gothenburg, Sweden) sampling at 100 Hz. Retro-reflective markers were positioned on the medial and lateral femoral condyles, medial and lateral malleoli, calcaneus and first, second and fifth metatarsal heads in order to define the right foot and shank. A cluster plate was attached to the shank and prior to each sandal condition a static, anatomically neutral, trial was recorded for each sandal, to define anatomical markers relative to dynamic clusters, similar to the CAST technique (Cappozzo, *et al.*, 1995). Kinematic data was filtered using a fourth order Butterworth low pass filter with a cut-off frequency of 10 Hz. Ankle joint angle ranges of motion were calculated in the sagittal and coronal plane. The Root Mean Square (RMS, window 50ms) was also calculated in the two planes to give a measure of deviation from neutral during the 30 second balance.

2.3.2 Centre of Pressure

Force data were simultaneously collected using an AMTI force plate (Advanced Medical Technologies Inc, Newton, Massachusetts, USA) built into the laboratory floor, sampling at 3000 Hz. CoP variables were defined relative to the foot segment, removing the effect of how participants positioned the foot on the force plate, which was not controlled. Data were exported into Visual 3D (Visual 3D Inc, Rockville, Maryland, USA) to calculate CoP variables, during which data was down-sampled to 100 Hz (Santos *et al.*, 2008) and filtered using a fourth order Butterworth low pass filter with a cut-off frequency of 25 Hz. During analysis CoP and GRF data were visually inspected to define when participants had gained balance, at least 10 seconds after their first contact with the force plate (Raymakers *et al.*, 2005). Variables were calculated for the 30 seconds starting at this event. CoP terminology and calculations are presented in table 3.8; example trajectories are represented in Figure 3.5.

Table 3.8 Centre of pressure variables calculated for the 30 second single-leg balance.

Variable	Unit	Calculation
Path length	mm	Total path length in millimetres for the CoP during single-leg stance. $\sum_{t=1}^{3000} t = \sqrt{(x_t - x_{t-1})2 + (y_t - y_{t-1})2}$
Anterior-posterior range	mm	Maximum posterior to maximum anterior position of the CoP co-ordinates: $\lvert ymax - ymin \rvert$
Medial-lateral range	mm	Maximum medial to maximum lateral position of the CoP co-ordinates: $\lvert xmax - xmin \rvert$
Ellipse Area	mm^2	Surface contained within an ellipse formed by the maximum ranges of the CoP: $\frac{\pi}{2} \cdot (x_{max} - x_{min}) \cdot (y_{max} - y_{min})$
Mean anterior-posterior velocity	$mm \cdot s^{-2}$	Mean velocity of the CoP in the anterior-posterior direction: $\frac{1}{3000} \sum_{t=1}^{3000} t = \sqrt{\frac{(y_t - y_{t-1})2}{t_t - t_{t-1}}}2$
Mean medial-lateral velocity	$mm \cdot s^{-2}$	Mean velocity of the CoP in the medial-lateral direction: $\frac{1}{3000} \sum_{t=1}^{3000} t = \sqrt{\frac{(x_t - x_{t-1})2}{t_t - t_{t-1}}}2$

Where *x* and *y* are anterior-posterior and medial-lateral centre of pressure (CoP) co-ordinates from the force-plate for sample time *t* (30 seconds at 100 Hz) respectively.

2.3.3 Electromyography

Electromyography activity was recorded at 3000 Hz using bipolar surface Ag/AgCl electrodes (Noraxon Inc, Scottsdale, Arizona, USA), with an electrode diameter of 10 mm and an inter-electrode spacing of 20 mm. Prior to electrode placement, in order to reduce noise, impedance and achieve an optimum EMG signal, hair was removed, skin exfoliated and cleaned with an isopropyl wipe. Electrodes were placed in accordance with the SENIAM recommendations (Hermens *et al.*, 1999) on the medial gastrocnemius, tibialis anterior, soleus, peroneus longus, biceps femoris, and rectus femoris of the right leg. The ground electrode was placed over the distal medial aspect of the medial tibial condyle. Cables were taped to the skin to reduce motion artefacts and participants wore a light jacket that housed the wireless transmitter such that data was collected within Qualisys software. EMG data was analysed in Visual 3D, data was zero-offset, full-wave rectified and smoothed with a root mean square (RMS) with a 200 ms window. The RMS value for the single-leg standing trials was averaged across 3 trials and presented as percentage change from CO condition for comparison (Nigg *et al.,* 2006b).

2.4. Statistics

Statistical comparison was undertaken in SPSS, EMG data was not normally distributed and therefore Friedman tests followed by Wilcoxon-signed rank test was utilised to detect significant differences. CoP and kinematic data were compared using repeated measures ANOVA. Both used Bonferroni correction for multiple comparisons (p value < 0.005 and <.05 respectively).

3. Results

3.1. Kinematics

Analysis of kinematic data identified differences in range of motion at the ankle joint in the different sandal conditions. A significantly higher sagittal range of motion at the ankle joint was recorded in MB than the other conditions (p = .000, Table 3.9). Increased coronal range of motion was evident in RE compared to the control (p = .000) and the SK unstable condition (p = .003). No measures of deviation from neutral (using the RMS data) identified significant differences between conditions (Table 3.9). Despite the differences in ankle range of motion, knee range of motion and RMS data did not differ significantly between conditions (Table 3.9).

3.2. Centre of Pressure

Centre of pressure data also identified significant differences between conditions. Specifically, as evident in Figure 3.5, the anterior-posterior range was significantly higher in MB compared to all other conditions (p = .008-.045). MB also elicited greater CoP ellipse area in participants than SK (p = .047, Table 3.10). No other variables demonstrated significant differences between conditions for CoP variables.

Table 3.9 Lower limb joint angle ranges of motion and root mean square data, statistically significant results are presented (determined using repeated measures ANOVA).

	CO	FF	MB	RE	SK	Significant *P* value
Ankle Sagittal ROM (º)	4.1 ±1.5	3.7 ±1.9	8.5 ±2.1	4.5 ±1.9	3.5 ±0.8	MB>CO p = .000, MB>FF p = .000, MB>RE p = .000, MB>SK p = .000
Ankle Coronal ROM (º)	7.1 ±3.5	8.9 ±2.1	9.7 ±2.4	13.5 ±4.7	7.6 ±2.5	RE>CO p = .000, RE>SK p = .003.
Ankle RMS sagittal (º)	6.8 ±3.0	7.6 ±3.2	7.4 ±4.1	6.4 ±2.8	7.3 ±3.1	-
Ankle RMS coronal (º)	6.7 ±1.6	6.7 ±2.1	8.5 ±2.5	8.4 ±2.8	6.8 ±2.3	-
Knee Sagittal ROM (º)	5.9 ±3.1	5.0 ±3.0	7.7 ±5.5	4.8 ±2.7	5.0 ±3.1	-
Knee RMS sagittal (º)	5.8 ±3.1	5.8 ±3.7	6.3 ±4.2	5.4 ±2.7	5.2 ±3.2	-

Data are expressed as Mean ± standard deviation. ROM= range of motion, RMS= root mean square, CO= control, FF= FitFlop, MB= Masai barefoot technology, RE= Reebok and SK= Skechers.

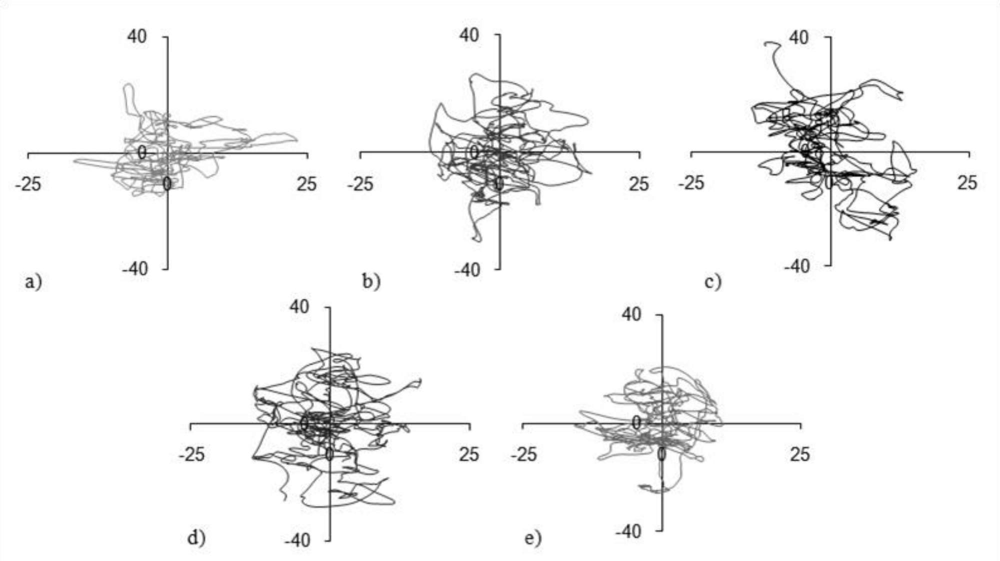

Figure 3.5 Example CoP trajectory (mm) of one participant for one 30 second balance trial in each condition.

Where x axis is medial-lateral distance and y axis is anterior-posterior. a) Control b) FitFlop c) Masai Barefoot Technology d) Reebok e) Skechers.

Table 3.10 Mean (±s) centre of pressure (CoP) variables, statistically significant results are presented (determined using ANOVA).

	CO	FF	MB	RE	SK	Significant P value
CoP Path Length (mm)	1219.2 ±260.6	1292.9 ±334.5	1322.1 ±265.0	1294.4 ±335.5	1244.5 ±290.7	-
CoP Anterior-posterior Range (mm)	48.9 ±11.0	53.0 ±8.4	64.0 ±10.9	50.3 ±15.0	49.3 ±12.3	MB>CO p =.008, MB>RE p =.030, MB>SK p =.011, MB>FF p =.045
CoP Medial-lateral Range (mm)	36.2 ±7.6	35.5 ±4.1	34.9 ±3.8	34.6 ±4.7	34.0 ±4.3	-
CoP Ellipse Area (mm^2)	1405 ±506	1433 ±290	1782 ±455	1286 ±488	1317 ±384	MB>SK P =.047
CoP Anterior-posterior Velocity ($mm \cdot s^{-1}$)	26.4 ±3.5	27.7 ±4.7	28.3 ±4.8	27.9 ±4.9	26.8 ±5.1	-
CoP Medial-lateral Velocity ($mm \cdot s^{-1}$)	29.5 ±3.5	28.7 ±4.9	29.9 ±5.9	29.3 ±5.6	28.5 ±6.2	-

CO= control, FF= FitFlop, MB= Masai Barefoot Technology, RE= Reebok and SK= Skechers.

3.3. Electromyography

In general EMG was greater in unstable footwear compared to the stable control, however a limited number of statistically significant differences were apparent. The FF (p = .002), MB (p = .003) and SK (p = .002) conditions all demonstrated greater RMS in medial gastrocnemius than the CO condition. Significant differences were also evident in the soleus and rectus femoris for muscle activation across the conditions tested (Figure 3.6).

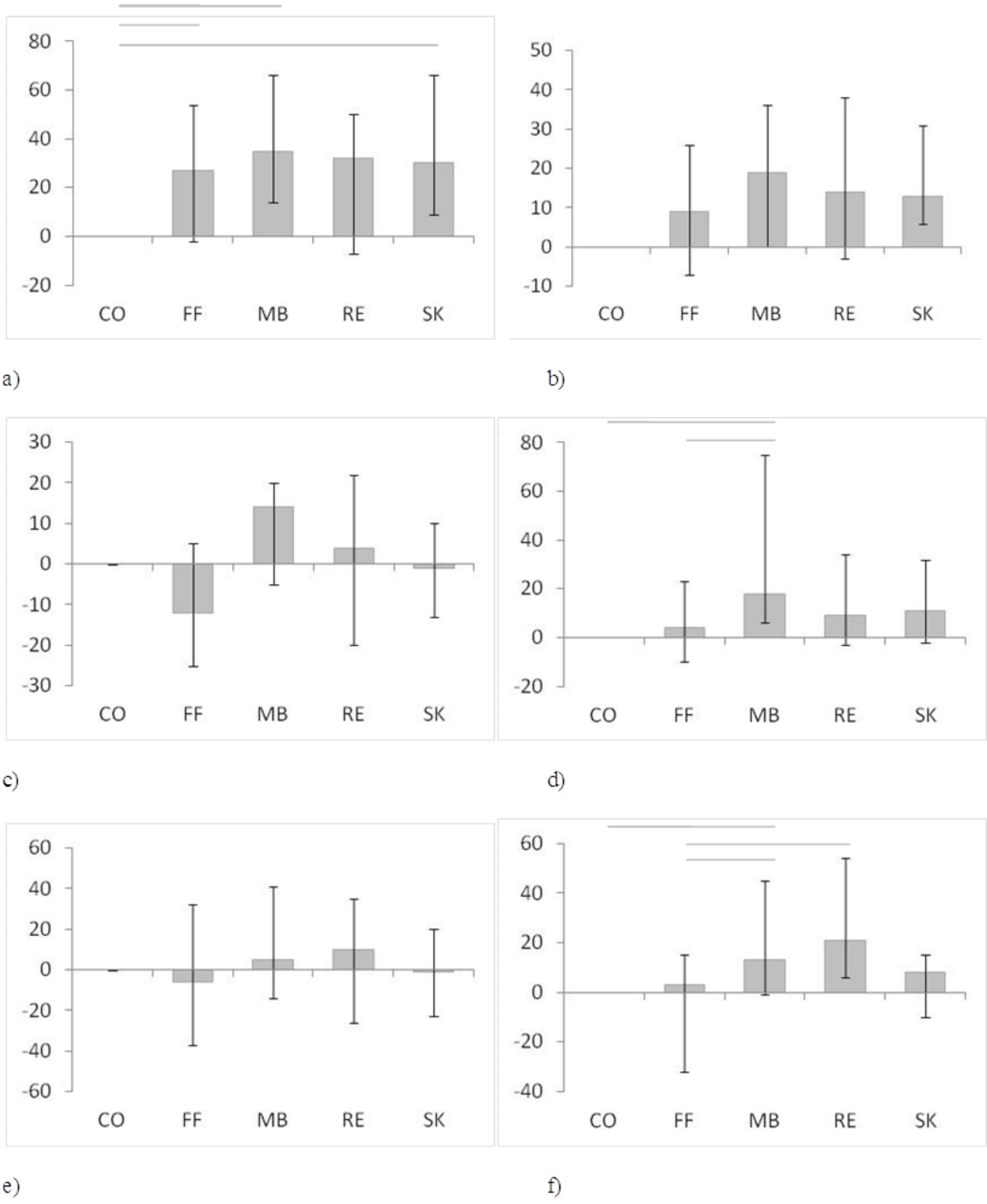

Figure 3.6 Median RMS (± inter-quartile range error bars) EMG for 30 second single-leg balance.

Where a) Medial gastrocnemius b) Peroneals c) Tibialis anterior d) Soleus e) Biceps femoris f) Rectus femoris. Median RMS (± inter-quartile range error bars) for each condition (x axis) as percentage difference to control (y axis) with significant differences (determined by Friedman then Wilcoxon signed-rank tests) indicated with horizontal bars (*P* <0.05).

4. Discussion

The study aimed to identify characteristics of single-leg standing in unstable sandals by quantifying lower limb kinematics, muscle activation and CoP characteristics. Lower limb kinematic data was collected to assess joint angle changes to maintain stability, which has previously not been quantified in unstable footwear.

The current study identified an increased sagittal range of motion at the ankle in MB. Stewart *et al.* (2007), from their plantar pressure analysis, postulated that participants sat on the fore-part of the MB shoe to control sway in standing. This technique for stance in MB would mask any instability features inherent in the footwear. This increased sagittal range at the ankle also suggests greater anterior-posterior movement; however the consistent RMS sagittal ankle value between conditions suggests that this was a one-off correction as opposed to a prolonged deviation from a neutral ankle position. The RE condition mirrored these results in the coronal plane. These findings presumably relate to the shape of the sandals, MB has a large anterior-posterior rocker profile whereas the RE balance pods appear to act as fore- and rear-foot medial-lateral rockers. The FF and SK conditions, however, do not include any rounded outsole design features. This would suggest that any instability is produced by the softer EVA in the midfoot, which is enclosed by the flat outsole surface. This may mean that these conditions are not as unstable within balance trials compared to gait. In balance the wearer is static on the footbed throughout, potentially maintaining stability on the firm heel and toe sections and not traversing the soft midfoot, therefore not being influenced by the characteristics of the shoe that were included to induce instability. No significant differences in knee kinematics were identified between conditions, identifying that any instability in the footwear could be controlled by adjustments at the ankle and did not impact higher up the limb.

In the present study MB had a significantly greater anterior-posterior range than all other test sandals, in which there was a mean 31% increase compared to CO. Numerous studies have identified increased CoP range in MB during standing (Nigg et *al.*, 2006b; Landry *et al.*, 2010). The studies report 65% (Landry *et al.*, 2010) and 105% (Nigg et *al.*, 2006b) increases in medial-lateral ranges in MB compared to control and 70% and 52% in anterior-posterior ranges respectively. In contrast to previous results, in the current data no significant difference in medial-lateral range was identified between conditions, despite the difference in ankle motion identified in the coronal plane in RE. Increased CoP anterior-posterior range is expected within MB footwear due to the larger rocker profile providing an unstable environment and ensuring the wearer must balance on the pivot point. The lack of differences found in the other unstable conditions may be explained by the RE balance pods being compressed when weight is evenly distributed across both forefoot and heel pods (as in single-leg standing in the current study), negating their effect. Similarly, FF and SK both incorporate a soft EVA construction under the midfoot, presumably during static activity the firm toe and forefoot and heel areas may eliminate any unstable midfoot effects by stabilising the foot. The MB sandal also incorporates higher elevation in the design, raising the centre of mass of the body and therefore increasing demand to maintain stability. The lack of medial-lateral differences within the current study may be due to methodological differences. Landry *et al.* (2010) utilised the participants' own work shoes as a control, which the subjects would have been more familiar with, a covered shoe would also likely be more stable for wearers. The current study tested sandals for which the fit of the upper on the foot is far less secure and less support is provided. Also both previous studies utilised double-limb standing, it is likely that subjects are less stable in single-leg standing (Hömme and Hennig, 2011) reducing differences between stable and unstable footwear conditions. CoP anterior-posterior velocity was higher, but not significantly, in all unstable conditions in the present study, with MB as the highest, consistent with Buchecker *et al*. (2012). CoP path length was also longer in the present study in all unstable sandals than CO, however this was not statistically significant, potentially due to the small sample size and large standard deviations.

In the present study total muscle activity over the 30 second balance task was greater in most muscles tested in the unstable sandals than CO (Figure 3.6), however consistent with previous literature, significant differences between conditions were limited (Germano, 2011, Landry *et al.*, 2010, Nigg *et al.*, 2006b). Landry *et al* (2010), using a linear array, studied EMG wavelet intensities of smaller lower limb extrinsic muscles; finding significant increases from CO in MB in flexor digitorum longus and anterior compartment muscles, including tibialis anterior. Nigg *et al.* (2006b), using wavelet analysis, reported a "trend" for an increase in muscle activity across tested lower limb muscles, particularly medial gastrocnemius, in standing, however no statistically significant differences were evident. Similarly, during single-leg standing Germano (2011) found no significant differences between unstable shoes and controls for both IEMG and RMS. In the present study medial gastrocnemius RMS was 25% higher in the unstable footwear than the control sandal (Figure 3.6). This increased RMS reached statistical significance in SK, MB and FF compared to CO, indicating a potential requirement to stabilise sagittal ankle motion. However this finding might also be due to the negative heel in the control condition. Soleus results support the stability concept, as there was a significant increase in the RMS in MB compared to CO and FF. This suggests that the large anterior-posterior rocker demanded more stabilisation at this joint than the other sandals to control balance and prevent resting on the forefoot in plantarflexion. These EMG findings also mirror the increase in anterior-posterior CoP trajectory in MB identified in this condition. In the peroneals and tibialis anterior no significant increases in muscle activation in unstable footwear were apparent, contrasting findings in MB (Buchecker *et al*., 2012). These differences may be due to the present study being more challenging in the control condition as subjects were tested in single-limb standing, compared to Buchecker *et al*. (2012) who utilised double-limb stance. Again, differences between the unstable sandals and the control in peroneal activation may have been affected by the CO sandal holding the foot in sight dorsiflexion and different results may be evident with an alternative control condition. At the knee, rectus femoris demonstrated increases in RMS activity in MB compared to FF. This may be indicative of an increased requirement to prevent the knee flexing, again relating to the large anterior-posterior rocker outsole of the MB condition.

Some limitations of the study include the use of sandals to assess unstable footwear, potentially decreasing the stability of the participants independent of any outsole or midsole features due to the lack of upper in the footwear, this may have been more like a barefoot control condition. Also, as previously noted, the control sandal selected may have impacted on muscle activation in the lower limb of participants. Characteristics of features of the subjects which may have influenced balance performance such as foot type (Hertel *et al*., 2002) may have also allowed better generalisation of study results to specific populations.

Conclusions

In summary, the results suggest that the tested unstable sandals did impact on parameters associated with stability in single-leg standing, although changes were subtle. Instability from the footwear was design-specific with the MB identifying anterior-posterior and the RE slight medial-lateral changes. It may be that the footwear has more impact on stability during walking as opposed to standing tasks due to walking dynamically moving across the footbed and using the full midsole design. Other population groups and comparison to other control conditions may also elicit more apparent instability in the footwear than identified in the current study design.

Conflict of Interest

The research has been co-funded by the U.K. government and FitFlop ltd as part of the Knowledge Transfer Partnership programme (KTP007228). The primary author works on the project, supervised by R.K.J. and P.G.S.. L.S. is not associated with the project. Work was undertaken with scientific diligence, data was collected, analysed and the paper written with no influence from the funding company.

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3.2.4 The effect of unstable sandals on instability in gait in healthy female subjects.

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Gait & Posture, 38(3),410-415.

Abstract

Unstable footwear generally lacks thorough peer-review published research to support concepts and marketing claims. The purpose of this study was to investigate the instability induced by four (FitFlop, Masai Barefoot Technology, Reebok Easy-Tone and Skechers Tone-Ups) commercially available unstable sandals and one stable control sandal (Earth) in walking in fifteen females (mean ± SD age was 29 ± 6.7 years, mass 62.6 ± 6.9 kg and height 167.1 ± 4.2 cm). Three-dimensional motion with synchronised electromyography and kinetic data were collected. Walking speed and step length remained consistent between conditions, however double support time decreased in Masai Barefoot Technology. Centre of pressure data identified no consistent difference between the stable control and the unstable sandals, however Masai Barefoot Technology reduced the anterior-posterior range of centre of pressure. Muscle activity differed significantly at the ankle in the unstable footwear. FitFlop, Reebok and Skechers increased peroneal activity during pre-swing, whereas Masai Barefoot Technology increased medial gastrocnemius and decreased tibialis anterior activity in loading response and mid-stance. The larger rocker sole of the Masai Barefoot Technology altered gait and muscle activation with regard to braking and progression in the sagittal plane. Reebok, Skechers and FitFlop, with softer, less stable foreparts increased evertor action at toe-off, having their effect in the coronal plane. The study highlighted that any instability induced by the shoes is design-specific.

Keywords: Gait, Instability, Footwear, Electromyography

1. Introduction

Within recent years unstable footwear has become increasingly popular as both a therapeutic and a functional tool [1,2]. The concept of unstable footwear is to deliberately reduce stability for directed outcomes, most commonly to increase muscle activation and "tone" the lower limb of the general population. Subsequently a myriad of brands have developed a wide range of unstable footwear for daily use in the form of shoes, sandals and boots. The brands utilise an array of technologies including rocker soles, balance pods and multi-density soles with the aim of inducing instability in the wearer to increase muscle activation and increase the demands of daily walking. Despite an array of marketing claims relating to muscle activity there are few empirical reviews comparing and quantifying the effects of unstable footwear on muscle activity, kinematics and kinetics in healthy individuals [3].

Quantifying stability in gait is a concept that has been considered in clinical groups [4], the elderly [5,6] and in relation to prosthetic feet and amputees [7]. Parameters to assess instability include variability of centre of pressure (CoP), vertical ground reaction force, electromyography (EMG) and joint angles. The application of these measurement and analysis techniques to unstable footwear may provide a more thorough assessment of the footwear than previously used protocols and identify small changes in gait induced by these footwear styles.

Previous unstable footwear research has predominantly focused on Masai Barefoot Technology (MBT). Nigg *et al*. [8] proposed MBT unstable shoes strengthen muscles which are anatomically closer to the axes of rotation, therefore reducing joint loading. Research on MBT has identified an increased range [9] and velocity [10] of CoP motion in standing, increased tibialis anterior activation during swing, increased gastrocnemius activation during early- and mid-stance [8,11] and improvements in reactive balance after an MBT intervention [12]. Other unstable footwear brands, such as Reebok, Skechers and FitFlop utilise different technologies, which lack the peer-reviewed published literature of MBT. Some independent comparative research has been undertaken, which does not support increases in muscle activation in MBT, Skechers Shape-Ups and Reebok Easytone when compared to a trainer [3,13], however the protocols utilised in these studies are less comprehensive including less stringent analysis of EMG than the MBT research. Small participant number unpublished studies commissioned by the footwear companies identify increases in lower limb muscle activation, both by prolonging and increasing the magnitude of muscle activation. The results from these studies are utilised in marketing material however the full protocols and analysis procedures of these studies are not currently in the public domain. Research commissioned by footwear companies acts as a starting point, with peer-reviewed published data being a must for this footwear category.

Due to existing research including low subject numbers, lacking peer-review and failing to compare commercially available brands, further study is warranted. Variables deemed particularly relevant but not previously compared between different unstable footwear styles were lower limb muscle activation in the phases of stance and CoP characteristics during walking. The present study was undertaken to quantify the immediate influence of the unstable sandals on gait to determine whether the footwear reduces stability in the wearer as is claimed. This aimed to clarify the differences and effectiveness in the technologies between sandals and characteristics of any instability induced in the wearer.

2. Methods

2.1. Participants

Fifteen female participants took part in this study. The mean ± SD age was 29 ± 6.7 years, mass 62.6 ± 6.9 kg and height 167.1 ± 4.2 cm. The study was approved by the University ethics committee and written informed consent obtained prior to participation. All participants reported themselves as in good health and with no recent lower limb injury prior to taking part in the study.

2.2. Sandal conditions

The control footwear tested was Earth and the unstable footwear conditions tested were FitFlop, Masai Barefoot Technology, Reebok Easy-Tone and Skechers Tone-Ups (Table 3.11). Earth was chosen as the control footwear as it makes no claims regarding instability, yet is a sandal which retails at a similar price and is aimed at a similar consumer as the unstable sandals.

Table 3.11 Footwear condition characteristics.

Image	Sandal	Abbreviation	Mass (g)	Description
	Earth Kalso	CO	193	3.7° incline in footbed from heel to toe with firm sole and flip flop upper.
	FitFlop Walkstar	FF	187	Multi-density ethylene vinyl acetate (EVA) midsole incorporating high-density heel, low-density midfoot and a mid-density forefoot.
	Masai Barefoot Technology Kisumu	MB	534	Thermoplastic polyurethane (TPU) heel and midfoot, compressible heel and pivot under the metatarsals with fibre glass forefoot. Rocker-sole in the anterior-posterior direction.
	Reebok Easy-Tone	RE	250	Air-filled compressible elliptical pods positioned under the heel and forefoot, which allow air to travel between the two.
	Skechers Tone-Ups	SK	195	Multi-density polymer midsole with firm forefoot.

2.3. Protocol

Three-dimensional kinematics of the lower limb and lower extremity muscle activity were measured for the right limb in each of the five conditions. For each condition, five trials of three gait cycles were collected at a self-selected walking speed. Condition order was randomised between participants. Kinetics and CoP characteristics were collected for one step, mid-trial, with a force plate embedded in the laboratory floor. Prior to data collection each participant performed practice walking trials to familiarise with the footwear and determine the starting position to enable successful force plate contacts without targeting. There was a maximum acclimatisation period of two minutes per sandal condition.

2.3.1. Kinematics

Three-dimensional kinematic data were collected using a sixteen camera motion capture system (OQUS, Qualisys AB, Gothenburg, Sweden) at a sampling rate of 100 Hz. Kinetic data were simultaneously collected with kinematics using an AMTI force plate (Advanced Medical Technologies Inc, Newton, Massachusetts, USA), which was embedded in the walkway, sampling at 3000 Hz.

Twenty four 14 mm retro-reflective spherical markers were placed on lower extremity joints in order to define the foot, shank, thigh and pelvis. Markers were placed bilaterally over the iliac crests, anterior superior iliac spines, posterior superior iliac spines, greater trochanters, the medial and lateral femoral condyles, the medial and lateral malleoli, calcaneous, and the first, second and fifth metatarsal heads. Marker cluster plates were attached to the pelvis, thigh and shank segments and prior to each sandal condition a static, anatomically neutral trial was recorded to define anatomical markers relative to dynamic clusters, similar to the CAST technique [14]. Data were exported into Visual 3D (Visual 3D Inc, Rockville, Maryland, USA) for processing and analysis. Joint angles were defined such that ankle dorsiflexion, inversion and knee and hip flexion were positive. Kinematic and kinetic data were filtered utilising a second order Butterworth low pass filter with cut-off frequencies of 10 and 25 Hz, respectively. Temporal-spatial data were output and joint angle ranges across stance were computed.

2.3.2. Centre of Pressure

Data were down-sampled to 100 Hz for analysis in Visual 3D; variables were calculated for each step then averaged across the 5 stance phases per condition. The CoP was defined relative to the foot segment and data were not normalised to foot size. Calculated variables were ranges and mean velocities in the medial-lateral and anterior-posterior directions. Anterior-posterior and medial-lateral ranges were defined as the maximum posterior to maximum anterior position and maximum medial to maximum lateral position of the CoP coordinates respectively.

2.3.3. Electromyography

Electromyography *(*EMG) activity were recorded simultaneously with the three-dimensional analysis at 3000 Hz using bipolar surface Ag/AgCl electrodes (Noraxon Inc, Scottsdale, Arizona, USA), with an electrode diameter of 10 mm and an inter-electrode spacing of 20 mm. Prior to electrode placement, hair was removed, skin exfoliated and cleaned with an isopropyl wipe. Electrodes were placed in accordance with the SENIAM recommendations [15] on the medial gastrocnemius, tibialis anterior, soleus, peroneus longus, biceps femoris and rectus femoris of the right leg. The ground electrode was placed over the distal medial aspect of the medial tibial condyle. Cables were taped to the skin to reduce motion artefacts and participants wore a light jacket to house the transmitter. EMG data were analysed in Visual 3D, data were zero-offset, full-wave rectified and smoothed with a Root Mean Square (RMS) (200 ms window). RMS was calculated for the phases of stance as defined by Perry's [16] subdivisions of the gait cycle; loading response (0-10%), mid-stance (10-30%), terminal stance (30-50%) and pre-swing (50-60%). Data were presented and compared for each subdivision and presented as a mean percentage change from the control sandal value.

2.4. Statistics

Statistical comparison was undertaken in SPSS, EMG data were not normally distributed and therefore Wilcoxon-signed rank test was utilised. CoP and kinematic data were compared using ANOVA. Both used Bonferroni correction for multiple comparison (*p* value<0.05). Individual participant differences were considered on a variable-by-variable basis.

3. Results

3.1. Temporal and Spatial parameters

Temporal and spatial parameters identified no significant differences for walking speed, step length or step and stance times between footwear conditions (Table 3.12). Swing and double-support time both differed significantly between conditions, with the MB footwear demonstrating longest swing time and shortest double support time.

Table 3.12 Mean ± SD temporal and spatial characteristics of gait, kinematic ranges of motion (ROM) and centre of pressure variables.

	CO	FF	MB	RE	SK	Significant Results
Walking speed ($m \cdot s^{-1}$)	1.28±.13	1.29±.14	1.31±.17	1.28±.12	1.28±.12	-
Step length (m)	.686±.045	.693±.041	.696±.047	.695±.047	.693±.041	-
Cadence (steps·min^{-1})	111.3±6.2	111.4±6.3	112.6±13.9	110.5±5.2	110.8±5.2	-
Stance time (s)	.635±.077	.630±.096	.657±.050	.638±.102	.659±.050	-
Swing time (s)	.429±.021	.422±.019	.445±.022	.431±.017	.419±.019	MB>FF *p* =.001, MB>SK *p* =.000
Double support Time (s)	.221±.038	.229±.036	.207±.033	.225±.033	.241±.035	FF>MB *p* =.000, SK>CO *p* =.000, SK>MB *p* =.000, SK>RE *p* =.000, RE>MB *p* =.003, SK>FF, *p* = .050.
Double support (% Gait Cycle)	20.7±3.1	19.7±2.6	18.7±2.2	18.7±1.8	20.1±1.8	CO>FF *p*= .046, CO>RE *p*= .030, MB<SK *p*= .021, SK>RE *p*= .000.
Ankle Sagittal ROM (º)	17.5±4.0	17.1±4.4	15.6±4.7	16.2±4.0	17.3±3.9	-
Ankle Frontal ROM (º)	14.5±5.5	14.8±5.5	14.2±5.4	16.0±5.6	14.6±5.0	-
Knee Sagittal ROM (º)	10.0±2.2	9.1±2.9	9.8±2.8	8.7±2.4	9.0±2.5	-
Hip Sagittal ROM (º)	45.1±6.2	44.8±7.4	43.2±5.2	46.4±5.0	46.4±4.7	-
CoP medial-lateral range (mm)	20.1±5.8	17.1±5.5	21.2±7.0	22.9±8.8	21.2±8.0	RE>FF *p* =.030
CoP anterior-posterior range (mm)	143.2± 40.6	138.5± 42.6	133.2± 45.2	139.4± 39.7	142.1± 46.3	CO>MB *p* =.004, SK>MB *p* =.001
CoP medial-lateral velocity ($mm \cdot s^{-1}$)	71.5±15.6	77.0±14.0	58.4±15.3	72.6±15.4	76.8±17.3	-
CoP anterior-posterior velocity ($mm \cdot s^{-1}$)	335.6± 39.9	335.6± 36.3	318.6± 43.2	345.5± 40.7	317.5± 91.4	-

Statistically significant results are presented (determined using ANOVA *p* values).

3.2. Kinematics

Kinematic joint ranges of motion in the lower limb were relatively consistent between conditions (Table 3.12). Ankle sagittal ranges were in greater dorsiflexion than in comparison to the CO condition, significantly greater in FF and SK (peak ankle values of 8.9± 1.3° FF (*p*

= .005), 8.7± 2.0° SK (p = .005), 9.4± 6.9° MB (p = .394) and 7.3± 1.8° RE (p = .363) compared to 7.0 ± 1.0° CO). Sagittal range of motion did not differ significantly between the shoes, highlighting that the apparent difference in peak dorsiflexion in CO may be due to the 3.7° increase in the static position due to the inclined footbed (Table 3.11).

3.3. Centre of Pressure

CoP data demonstrated mean reductions in medial-lateral range in the FF condition and anterior-posterior range in the MB condition (Table 3.12). Further examination of intra-participant data identified this pattern was consistent for 10 or more of the 15 participants for both variables. The MB mean CoP velocities were lower than the other footwear conditions, as only two participants had faster anterior-posterior velocity and medial-lateral velocity in MB than CO.

3.4. Electromyography

Electromyography demonstrated significant increases in muscle activity in the unstable footwear compared to CO, particularly in the peroneals (Fig 3.7, Table 3.13). During loading response and continuing into midstance, tibialis anterior was significantly higher in all conditions than MB (p = .005, Fig 3.7c). Contrasting this, the peroneus longus demonstrated increased activation in MB compared to CO (p = .020), FF (p = .015), and SK (p = .020) during loading response (Fig 3.7b, Table 3.13). Peroneus longus activation also differed significantly in pre-swing where FF (p = .025), SK (p = .010) and RE (p = .025) demonstrated greater activation than CO. Soleus activation during midstance was significantly lower in FF and SK than MB and CO (Fig 3.7d). This pattern was also apparent in medial gastrocnemius, where in midstance the muscle activity in the RE, CO and MB conditions exceeded that recorded in FF and SK (p = .005-.025) (Fig 3.7a).

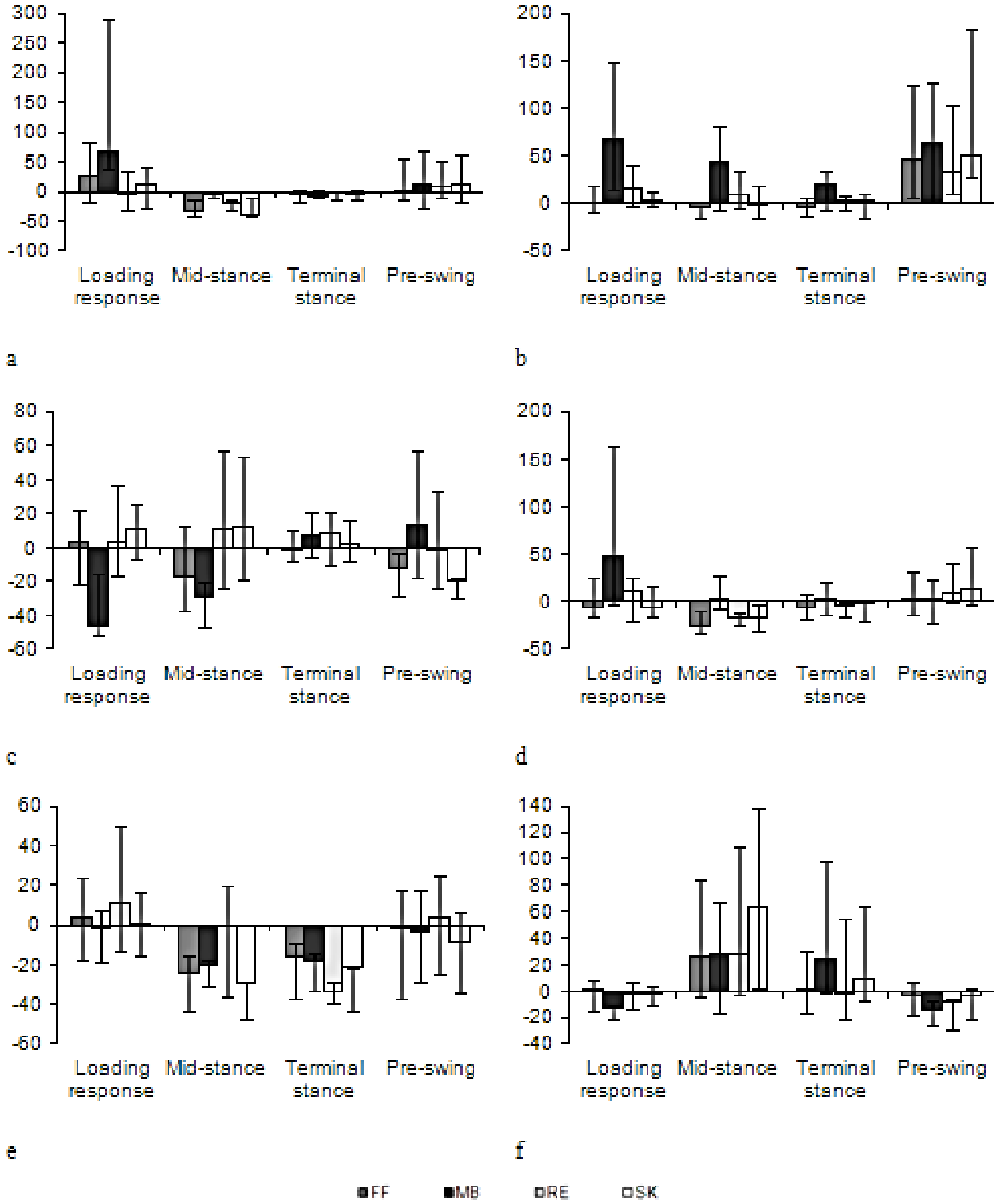

Figure 3.7 Median RMS (± inter-quartile range error bars) EMG for phases of stance (x axis) presented as percentage difference from control.

For a) Medial gastrocnemius b) Peroneus Longus c) Tibialis anterior d) Soleus e) Biceps femoris f) Rectus femoris.

Table 3.13 Electromyography statistically significant differences for the phases of stance

	Loading Response	Mid-Stance	Terminal Stance	Pre-swing
Tibialis anterior	MB<CO *p* = .005 MB<FF *p* = .005 MB<RE *p* = .005 MB<SK *p* = .005	MB<RE *p* =.005 MB<SK *p* =.015	-	-
Peroneus Longus	MB>CO *p* =.020 MB>FF *p* =.015 MB>SK *p* =.020	MB<FF *p* =.005	-	FF>CO *p* = .025 RE>CO *p* =.025 SK>CO *p* = .010
Soleus	-	FF<CO *p* = .005 FF<MB *p* = .005 SK<CO *p* = .005 SK<MB *p* =.005	-	-
Medial Gastrocnemius	MB>CO *p* = .010 MB>RE *p* = .020 MB>SK *p* = .010	FF<CO *p* = .005 FF<MB *p* = .010 SK<CO *p* = .005 SK<MB *p* = .005 SK<RE *p* = .025	-	-
Biceps femoris	-	FF< CO *p* = .025	-	-
Rectus femoris	-	MB<CO *p* = .020	-	-

Determined by Wilcoxon signed-rank tests with Bonferroni correction, $p < 0.05$.

4. Discussion

In the present study unstable footwear was shown to immediately alter parameters associated with stability in gait. All results discussed relate to immediate effects as opposed to long term effects. Longitudinal studies may elicit different results. Looking at temporal-spatial parameters; self-selected walking speed was consistent across tested footwear conditions, with no significant changes in step length or cadence. Contrary, Romkes *et al.* [11] identified a reduction in walking speed with MB, due to decreased cadence and step length. The control shoe utilised by Romkes *et al.* [11] was the participants' own street shoe, not another previously unworn unfamiliar sandal, which may in part account for these contrasting findings. Other research on MB has controlled walking speed which may reduce apparent differences between footwear conditions [8,17].

An increased double support time has been related to instability in gait [18]. The present study identified decreased double support times in MB in absolute terms, significantly lower than all other unstable conditions. As a percentage of the gait cycle, RE and MB showed shorter durations than CO and SK, FF showed no significant differences. This finding is potentially due to the previously proposed mid-foot contact of MB [8,19] and the pivot created beneath the foot in a rocker shoe increasing the speed of ambulation. Double support is indicative of the most stable phase of gait so reducing this time is theoretically increasing instability, however to claim any functional benefit an increase in EMG must be apparent. It may be that this variable would have more relevance in a different population as opposed to young, healthy females who are likely to remain stable.

The commonly reported anterior-shift in heel contact in MB [10,19] is evident in the reduced anterior-posterior CoP range in the present study alongside the decreased CoP velocity. The decreased CoP velocity is due to the increased stance time and due to the heel-strike occurring further down the foot and missing the CoP trajectory that would have been the fastest. The CoP ranges and velocities did not differ significantly between CO and unstable footwear conditions. The CoP trajectory in gait in unstable footwear has not previously been reported, however joint moment studies on MB allude to the current results with a reduction in dorsiflexor moment after initial contact, potentially a sign of the force vector position being more distal along the foot [20,8].

Muscle activation increases claimed by footwear companies marketing material range from 11-35% increases in the lower. Despite this, previous literature has failed to identify significant differences between MB and trainers [8] and other unstable footwear and trainers [3,13]. The two comparative instability papers normalised to maximal voluntary contraction and produced a measure of total muscle activity not divided into the phases of gait, potentially masking in-shoe differences. The present study divided the gait cycle and examined periods in stance in order to attempt to examine more closely the effects of the tested unstable footwear which allow more discrete analysis of the functional movements of gait.

The EMG results for MB mirror those previously reported in Romkes *et al*. [11]. At heel-strike and during loading, increased dorsiflexion occurs in conjunction with an increase in the activity of the peroneals, soleus and medial gastrocnemius. Alongside this there was a reduction in tibialis anterior activity which continues to mid-stance, also consistent with Romkes *et al*. [11], which the authors attributed to increased co-contraction to stabilise the ankle joint in the sagittal plane at heel-strike. The other unstable sandals show significantly greater activation in tibialis anterior during mid-stance, likely due to MB already having the ankle in dorsiflexion [8,21]. This pattern may also be linked to the upper of the shoes tested, the MB condition was a sandal, the other unstable conditions had flip-flop uppers and may have required muscle activation to control the shoe [22]. Medial gastocnemius and soleus also demonstrated significantly greater EMG patterns in mid-stance in the CO sandal than FF and SK. The higher activation in the plantarflexors in mid-stance is likely initiated to counteract the dorsiflexed position in the CO sole, in preparation for progression. Therefore this finding may not be repeated with a flat control shoe. Potentially muscle activation differences can be explained by the ankle position enforced by the sole shapes; FF and SK have effective heel heights of approximately 1.5 cm, keeping the ankle in relative plantarflexion, the other unstable sandals are flat in theory at this point, the RE air pods have compressed and MB heel section collapsed. During pre-swing the peroneal activity was significantly higher in the RE, SK and FF unstable sandals than the CO. This increase is likely attributable to the soft forefoot parts of the midsoles in these shoes. At toe-off the increased peroneals activity was likely initiated to counteract the lack of medial-lateral resistance to inversion from these midsoles. No increases in coronal ankle range of motion during stance would suggest that the muscle activation was enough to stabilise the joints and prevent the soft shoes allowing a large increase in eversion in SK and FF. It is apparent that the MB condition affects sagittal stability, while the SK and FF conditions effectively apply coronal plane instability to the wearer. The nature of the instability applied by each shoe should be considered by clinicians if the shoes are to be prescribed as a rehabilitation device.

5. Conclusions

Findings from the present study provide an overview of the effects of some unstable sandals that are currently popular. The research identifies that an increase in muscle activation is apparent using the unstable sandals, however the effects appear to be specific to phases of gait and sole-shape dependent. Alterations in CoP variables and joint angles also relate back to the method of instability imposed in the sandals. The long-term influence of the footwear on gait of wearers should be investigated further through a longitudinal study.

Conflict of Interest:

The research has been co-funded by the U.K. government and FitFlop ltd as part of the Knowledge Transfer Partnership programme (KTP007228). There is no conflict of interest. The primary author works on the project, supervised by Dr Jones and Dr Graham-Smith. L. Smith is not associated with the project. Work was undertaken with scientific diligence, data was collected, analysed and the paper written with no influence from the funding company.

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3.2.5 A comparison of plantar pressures in a standard flip-flop and a FitFlop using bespoke pressure insoles.

Price C, Smith L, Graham-Smith P and Jones R. (2013).
Footwear Science 5(2), 111-119.

Abstract

Purpose: The study was undertaken to quantify plantar pressures in a Havaiana flip-flop compared to a FitFlop. The FitFlop is a flip-flop designed to induce instability in the wearer during midstance as it incorporates a multi-density midsole in the design. It was hypothesised that in the Havaiana the toes are used to "grip" the shoe in swing and the loose upper and thin sole provide limited protection to the foot, producing higher plantar pressures than FitFlop. It is presumed that high plantar pressures are experienced in flip-flops and they may lead to discomfort in walking. Methods: Twenty female subjects walked in the footwear conditions while a bespoke instrumented insole quantified plantar pressures. Data analysis grouped sensors into regions for the heel, 1st MPJ and hallux to isolate pressures that have been linked to comfort and symptoms reportedly alleviated in FitFlop. Additional analysis was undertaken to measure hallux "gripping" during swing. Results: Significant reductions in plantar pressures in FitFlop, particularly in peak pressure in the heel (3.6%) and pressure time-integral in the 1st MPJ (12.0%) were identified. These findings were attributed to the thicker midsole with different EVA construction and a redistribution of load to the midfoot where contact area increased by 19.9% compared to Havaiana. Also evident were reductions in anterior-posterior centre of pressure velocity in FitFlop, attributed to its softer midfoot delaying progression. Hallux variables identified reductions in time spent "gripping" as well as the magnitude of force applied by the hallux in swing in FitFlop. Conclusions: Findings from the study identify that the FitFlop reduces pressure in key areas of the foot which are associated with walking comfort as well as clinical conditions. The "gripping" mechanism postulated to hold flip-flops on is lessened in the FitFlop, potentially reducing the likelihood of overuse injuries.

Keywords: plantar pressure, centre of pressure, footwear, peak pressure, sandal.

1. Introduction

Flip-flops break general recommendations for footwear by not covering and protecting the dorsal foot and toes, the upper being loose, not including a medial arch, by having a thin midsole and having no pitch from heel to toe (McPoil 1988, Barton *et al.* 2009). Because of these characteristics, tripping, puncture wounds, and cuts to the toes are associated with wearing flip-flops. Also localised heel pain is common for wearers (American College of Foot and Ankle Surgeons 2007). Despite extensive criticisms of this footwear style, a thorough search of peer-review published literature identifies limited scientific investigation into the effect of flip-flops on wearers (Carl and Barrett 2008, Chard and Smith 2011, Shroyer *et al.* 2010). Some research that is in existence alludes to flip-flops being potentially beneficial, by reducing knee loading in an osteoarthritis population (Shakoor *et al.* 2010) and by reducing pressures on the soles of the feet when compared to barefoot walking (Carl and Barrett 2008). Additionally there is a potential that open shoes reduce the prevalence of flat feet in children compared to closed (Rao and Joseph 1992), suggesting that flip-flops may offer some advantages despite general recommendations.

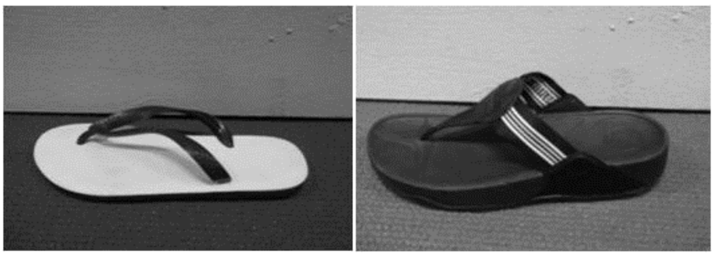
Studies of walking gait in flip-flops have identified a reduction in stride length compared to other footwear (Finnis and Walton 2008, Shroyer and Weimar 2010), moderations to ankle angle in swing and reductions in eversion in mid-stance compared to barefoot (Shroyer *et al.* 2010). These studies allude to alterations in gait when wearing flip-flops, which may affect loading on the joints of the lower limbs and the feet. Plantar pressure analysis is a tool that can be utilised to quantify the direct effect of wearing the footwear on the loading of the plantar tissue of the foot. Pressure variables have been linked to comfort (Che *et al.* 1994, Jordan *et al.* 1997) and compared in a variety of footwear conditions in published research including sandals (Song *et al.* 2005, Carl and Barrett 2008). Song *et al.* (2005) compared comfort, peak plantar pressures and pressure time integrals in five different Birkenstock sandals with differing arch heights. Findings suggested that mid-range arch heights produced the lowest pressure time integral beneath the 1st metatarsal-phalangeal joint (1st MPJ) and were reported to be the most comfortable. Che et al. (1994) also report that the 1st MPJ is the most sensitive region for differentiating between comfortable and uncomfortable footwear. Carl and Barrett (2008) compared peak plantar pressures during walking in a flip-flop to a trainer and barefoot, identifying that the flip-flop reduced peak pressures under the heel compared to barefoot, with trainers producing the greatest reduction. This is expected as a layer of viscoelastic material has been positioned beneath the barefoot heel in a flip-flop and this material is thinner than that of the trainer. The authors commented that the pressure measurement did not identify any changes associated with "gripping" the flip-flop with the toes to hold it on to the foot, however the analysis process undertaken would have been unlikely to capture such values as peak pressures throughout gait were compared. Further investigation of plantar pressure in flip-flops is therefore warranted, with a more stringent methodology to isolate key features associated with this footwear style.

The FitFlop is a flip-flop designed to induce instability in the wearer during midstance as it incorporates a multi-density midsole in the design (Figure 3.8; www.fitflop.co.uk). FitFlop publish numerous testimonials reporting high levels of comfort and reduced joint pain (Testimonials at FitFlop 2012). The reasons for these beneficial outcomes when wearing the shoe have not been identified, but may be attributable to reductions in plantar pressures. The present study, therefore, aims to quantify plantar pressures, contact areas and centre of pressure (CoP) trajectory in a standard flip-flop and FitFlop, which may allude to comfort in the footwear and reasons for reported relief of symptoms in testimonials. Testing will use stringent methodologies and a bespoke insole. It is hypothesised that FitFlop will reduce peak pressures in key areas such as under the 1^{s} MPJ and heel due to a thicker midsole, alongside reducing gripping with the hallux in swing due to a better fitting upper and increasing contact area due to the softer midsole under the medial midfoot compared to standard flip-flops.

2. Methods

2.1 Footwear Conditions

Participants walked two footwear conditions, a standard flip-flop (Havaiana) and a FitFlop (Figure 3.8). Conditions are described in Table 3.14.



a b

Figure 3.8 . Footwear conditions tested: Havaiana flip-flop (a), FitFlop, Walkstar I (b).

Table 3.14 Footwear features for the two test conditions.

Condition	Midsole Construction	Sole Depth (mm)	Hardness (Shore A)	Shoe Mass (g)
Flip-flop Havaiana Brazil	Ethylene Vinyl Acetate (EVA)	16	33	Size UK 6 140
FitFlop Walkstar	Multi-density EVA midsole. Rubber outsole.	Heel: 34 Midfoot:22 Toe 16	Heel: 55 Midfoot: 28 Toe 38	Size UK 6 172

2.2 Participants

20 female participants were tested, all wore a size U.K. 6 shoe, which was the size of the bespoke insole. Participants had a mean±*sd* age of 31±9 years, mass of 64.7±6.4 kg and height of 1.63±0.05 m. Participants were recruited from the University staff and student population. All indicated they were healthy and free of lower extremity injury and gave written informed consent fulfilling the requirements of the University Ethics Committee.

2.3 Protocol

Plantar pressure data was collected utilising the Medilogic (T&T Medilogic, gmbh, Germany) in-shoe pressure measurement system operating at 60 Hz with a bespoke insole (described later). Participants were instrumented with insoles in the footwear while a transmitter (fastened around their waist) wirelessly transferred data to a laptop. Insoles were secured in the footwear with small squares of double-sided tape positioned at the toe, mid-foot and heel. Footwear conditions were randomised between participants and 2 walks were undertaken in each condition to familiarise to each footwear condition. Participants undertook 4 walking trials of 15 m per condition on a concrete floor. Photoelectric timing gates, 10 m apart, were utilised to monitor walking speed, to ensure all trials were within 5% of each participants' first trial. Trials outside this boundary were excluded and additional trials were undertaken. Pressure data was recorded for the middle 10 m of the trial where walking speed was monitored.

2.4 Data Treatment

Data was exported from Medilogic and analysed using a custom-written analysis procedure in Microsoft Excel. Analysed variables were calculated only for the right foot for comparison (Menz 2004) and were specific to tested footwear styles and rationale. Analysis regions were chosen specifically as they are anatomical points that have previously been identified as being related to footwear comfort (Hong et al., 2005; Che et al., 1994). Masks were produced to capture the heel, 1st MPJ and hallux position in-shoe, grouping sensors in these regions for analysis (Figure 3.9). Divisions of the foot were based on a simplified version of a commonly used mask to capture only the regions desired in the current comparison (Cavanagh and Ulbrecht 1994). Hallux gripping was quantified by calculating the pressure applied during swing (Table 3.15). CoP variables were output unit-less so are presented as percentage change between conditions, the trajectory was then interpolated and normalised to stance to allow graphical comparison. Statistical tests were undertaken in SPSS, utilising Wilcoxon signed-rank test as data was non-parametric. Results are therefore presented as median±inter-quartile range, with $p<0.05$ chosen to denote significance.

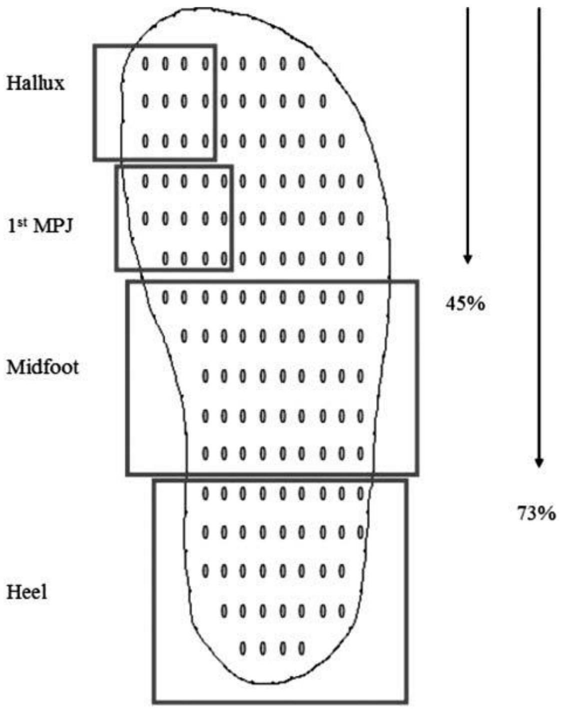


Figure 3.9 Region definition for the in-shoe plantar pressure.

0 denotes single-sensors and boxes group the sensors into the anatomical regions utilised in the study.

Table 3.15 Variables calculated from plantar pressure and centre of pressure data from Medilogic.

Variable	Definition
Whole foot	
Stance Time	Total time from initial pressure in heel to final pressure in hallux.
Total Contact Area	Total area of sensors loaded during stance as a percentage of total insole area.
Total Force-time Integral	Integral of the total force calculated across all sensors in the insole,
Total Pressure-time Integral	Integral of the pressure curve for the foot over stance.
Pressures for hallux, 1st MPJ and heel regions	
Peak Pressure	Mean of peak value recorded in each loaded sensor in the region in stance.
Pressure-Time Integral	Integral of the pressure curve for each region over the time that the region was loaded.
Grip	
Mean Pressure Hallux Swing	Mean value across sensors of the mean pressure recorded in the hallux during swing (when the hallux sensors were loaded prior to the heel sensors).
Time Pressure Hallux Swing	Time for which the mean pressure across all sensors in the hallux region was greater than zero.
Centre of Pressure	
CoP ML range	Range of CoP in medial-lateral direction (maximum-minimum)
Mean CoP ML position mid-stance	Mean value of CoP in medial-lateral direction during mid-stance as defined by Perry and Burnfield (2010)
CoP AP Range	Range of CoP in anterior-posterior direction (maximum-minimum)

2.5 Instrumented Insole

To conduct the study a bespoke insole was constructed to measure plantar pressure in flip-flops without any data loss associated with cutting the insole to accommodate a toe-post (Figure 3.10). The insole contained 150 surface resistive sensors (Figure 3.9) and was adapted with a small cut-away hole and slit from the medial edge to fit the toe-post. The insole had a larger area (≈190 cm^2) than standard insoles to cover a greater plantar area in open footwear due to the wider last and lack of rand for upper attachment. The repeatability and validity of data from the bespoke insole was established prior to testing with a two-session repeat test of 5 participants walking in the insole over a force plate. Correlation coefficients between sessions ranged from 0.63-0.99. The lower correlations were apparent in the average CoP anterior-posterior velocity ($r = 0.67$) and the pressure time integral in the heel ($r = 0.63$). All other reported variables exceeded 0.82, with contact area in the midfoot being particularly consistent ($r = 0.99$). Correlations between force plate and insole centre of pressure trajectories and velocities produced good to strong correlations. Medial-lateral variables demonstrated weaker relationships (range $r = 0.67$ and mean velocity $r = 0.50$) than anterior-posterior variables (range $r = 0.67$ and mean velocity $r = -0.84$). Correlation between the ground reaction force and total force calculated from the insoles was high, however a standard error of the estimate demonstrated a mean value of 39N, demonstrating a substantial underestimation by the insoles, a conversion factor of *1.6 produced comparable peak loading values from both sources consistent with findings using other in-shoe measurement systems (Hennig *et al*., 1996).

Figure 3.10 Bespoke instrumented insoles positioned and fastened with double-sided tape in the FitFlop test condition.

3. Results

Walking speed was controlled and therefore consistent between the two conditions, with Havaiana (1.27±0.12 $m{\cdot}s^{-1}$) and FitFlop (1.27±0.11 $m{\cdot}s^{-1}$) not differing significantly. Despite this, the analysis of plantar pressure variables in Havaiana and FitFlop identified significant differences between the two conditions. An increase in stance time was seen in the FitFlop (.540±.050 s) compared to Havaiana (.535±.050 s, p = .004). Total force-time integral was also increased by 10.4 % in the FitFlop condition compared to Havaiana (p = .009), however the pressure-time integral for the whole foot was 5.6 % higher in Havaiana than FitFlop (p = .004). Total contact area was higher in FitFlop (87.9±10.0%) than Havaiana (80.6±8.3%, p = .000), with this increase largely attributable to a median 19.9 % increase in the midfoot in this condition compared to the Havaiana.

Table 3.16 Median ± inter-quartile range for regional pressure variables.

Region	Variable	FitFlop	Havaiana	p values
Hallux				
	Peak Pressure (kPa)	156.6±226.9	186.5±228.1	.218
	Pressure-time Integral ($kPa{\cdot}^{s-1}$)	155.9±293.2	187.5±352.0	.313
	Contact Area (%)	78.1±31.8	78.1±37.0	.600
1st MPJ				
	Peak Pressure (kPa)	197.8±130.8	219.0±70.3	.073
	Pressure-time Integral ($kPa{\cdot}^{s-1}$)	444.7±224.1	508.3 ± 231.0	.001
	Contact Area (%)	100±0.0	99.2±6.3	.005
Heel				
	Peak Pressure (kPa)	170.3±120.9	176.6±135.0	.010
	Pressure-time Integral ($kPa{\cdot}^{s-1}$)	814.1±461.1	855.4±534.0	.247
	Contact Area (%)	80.7±3.6	78.8±7.5	.183

Foot-region results are presented in Table 3.16. Peak pressures were 16.0 % (p =.218) and 9.6 % (p=.073) higher in the Havaiana condition in the hallux and 1st MPJ respectively, although large ranges were evident and these differences were not statistically significant. Variables in the hallux showed greater intra-subject variability than other regions, with 12 participants demonstrating increased peak pressure and pressure-time integral under the hallux in FitFlop compared to Havaiana, despite the median value being higher in Havaiana. The pressure-time integral under the 1st MPJ was 12.5 % higher in Havaiana compared to FitFlop (p=.001), however no other regions pressure-time integral differed significantly. FitFlop demonstrated slight reductions in peak pressure (3.6 %, p =.010) and pressure-time integral (4.8%, p=.247) in the heel, a pattern which was consistent in three quarters of all participants. Contact areas for reported regions differed by only 0-1.9% between conditions, however in the 1st MPJ FitFlop had a significantly higher contact area (p=.005), with all participants demonstrating full coverage of the sensors in this region during stance.

The CoP variables identified significant differences between conditions, with the mean medial-lateral CoP position during mid-stance demonstrating the trajectory was more lateral in FitFlop (p = .000, Figure 3.11). Also evident in Figure 3.11, the Havaiana condition demonstrated an increased CoP anterior-posterior range, 8.2 % higher than the FitFlop condition (p = .002). FitFlop CoP trajectory moved faster toward the lateral (p = .048) and Havaiana CoP trajectory moved faster toward the medial (p = .002) side of the foot. Maximum and average velocity towards the toes was fastest in Havaiana, 22% and 6.6% faster than FitFlop respectively (p = .002). In FitFlop the CoP velocity towards the toes demonstrated that in FitFlop there was a backward motion of the trajectory present, with only 4 participants not demonstrating this, in Havaiana 9 of the twenty demonstrated this pattern. Further analysis of participants that demonstrated this backward velocity identified that the backward motion occurred in late stance and was 17% faster in FitFlop than Havaiana.

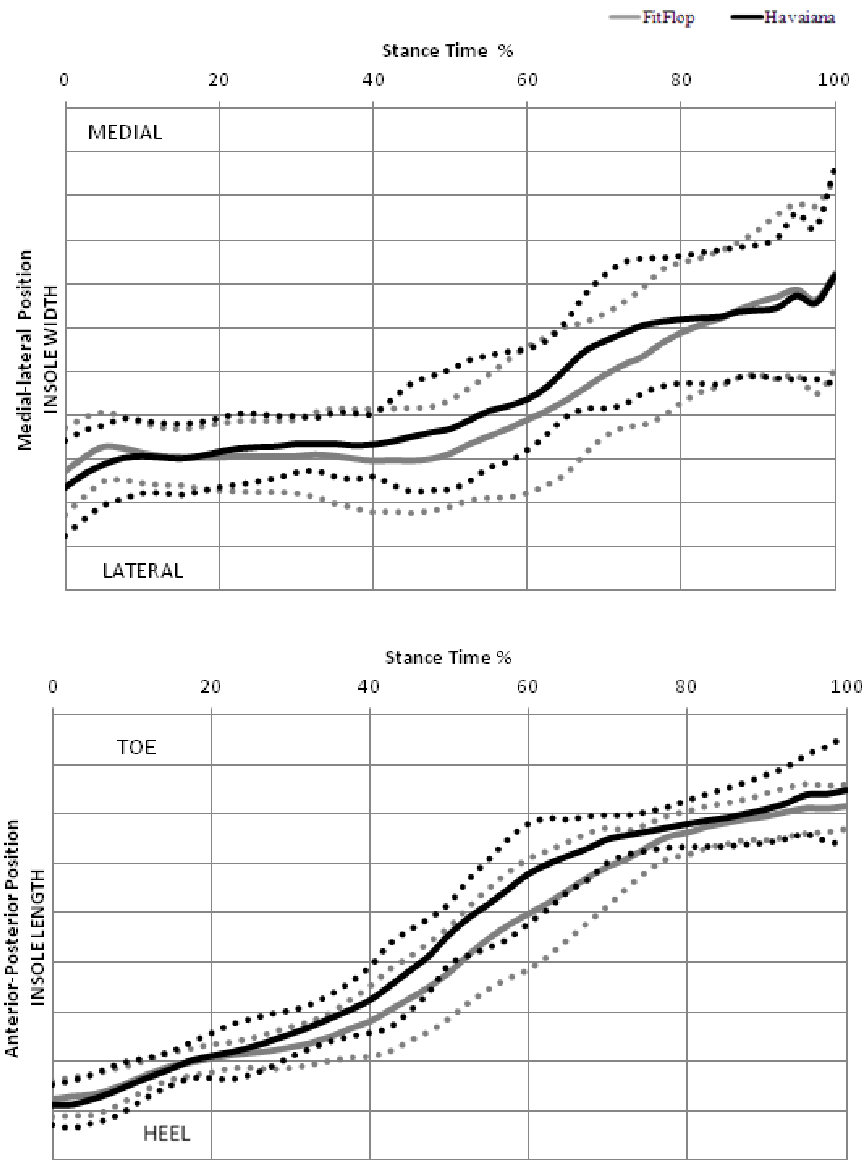


Figure 3.11 Median CoP trajectory in the medial-lateral and anterior-posterior directions with inter-quartile range denoted by dashed lines.

Variables to denote “gripping” with the hallux in swing identified significant differences between conditions; an exemplar participant is presented in Figure 3.12. The median participants results demonstrated that the average duration of time the hallux exerted pressure in swing was significantly longer in the Havaiana (.094±.210 s) than FitFlop (.002±.092 s, p = .001). In addition to a longer time spent gripping the sole, the pressure exerted under the hallux was of significantly higher magnitude in the Havaiana (.36±.62 kPa) compared to FitFlop (.05±.50 kPa, p = .020).

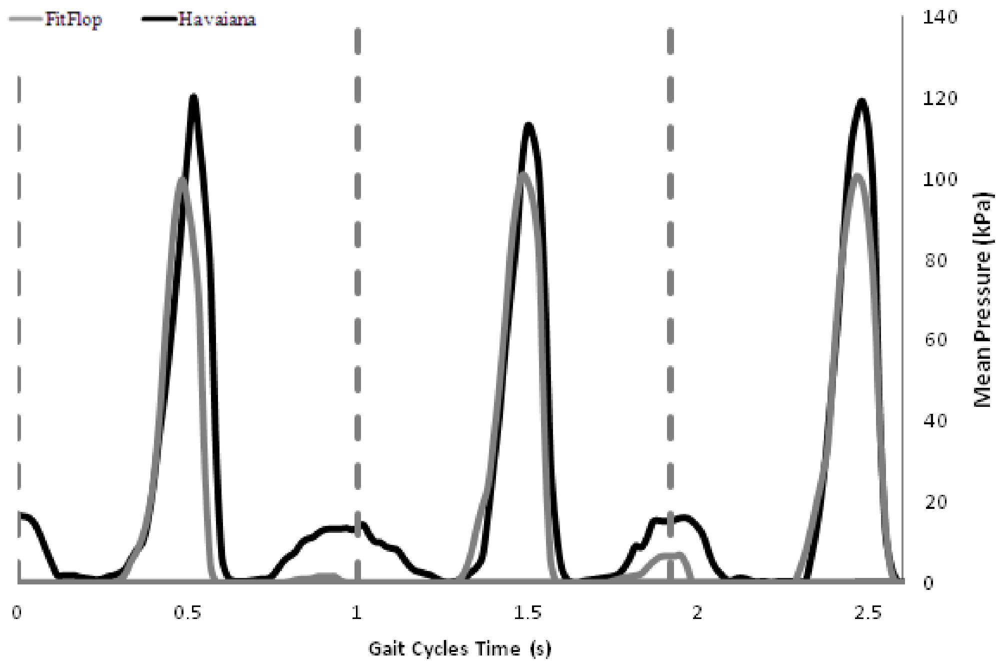


Figure 3.12 Example average hallux pressure during gait cycles for one subject where dashed lines denote heel-strike.

4. Discussion

Increased plantar pressure is a common factor in the development of foot pain and discomfort in walking in footwear (Che *et al.* 1994, Jannink *et al.* 2006, Witana *et al.* 2009). Therefore decreasing the pressure exerted at the foot-shoe interface is an important aspect of functional footwear. The purpose of this study was to assess the plantar pressures when walking in a conventional flip-flop (Havaiana) to walking in a FitFlop, exploring numerous testimonials regarding decreased lower limb complaints and increased comfort in FitFlop. This study found significant differences in plantar pressure between the two footwear conditions, which have not previously been quantified and reported. These findings were obtained by using a bespoke insole to allow pressures to be captured around the toe-post and an analysis methodology to quantify “gripping” with the Hallux in swing.

The results identified an increase in stance time in the FitFlop compared to the Havaiana, suggesting that, despite the fixed walking speed participants gait differed in the two styles (Cavanagh *et al.* 1992). An increase in total force-time integral in FitFlop was apparent alongside the increased stance time, which would have increased the duration of force application, explaining the variation in this variable. Increased stance time is consistent with previous literature investigating FitFlop, which was attributed to a thicker midsole making heel-strike earlier or instability from the midfoot of the midsole construction delaying progression to the toe (Price *et al.* 2010). This concept is supported by the reduced CoP anterior-posterior velocity identified in the present study. Also sixteen of the participants in the present study had a reversal of the CoP trajectory toward the heel near toe-off in FitFlop, this is likely caused by the softer EVA positioned under the metatarsal heads displacing at push-off causing a slight instability and slowing of progression during midstance. These findings are consistent with previous literature which reported reduced mean CoP velocity under the metatarsal heads in flexible footwear and footwear with a raised heel (Grundy and Tosh 1975), both features are apparent in the FitFlop design. This finding is also consistent with the increased pressure-time integral identified under the Hallux in some participants, if a prolonged contact is induced. Despite the increased force-time integral, overall pressure on the foot sole decreased each step in FitFlop compared to Havaiana. Plantar pressure in footwear is determined by the shape and properties of the footwear outsoles and insoles (Wenyan and Goonetilleke 2009). The FitFlop midsole is thicker than the Havaiana midsole and composed of rubber with different properties, likely reducing the force applied to some areas of the plantar surface. The softer midsole may have also deformed which enabled a median 7.3% increase in total contact area, thus also contributing to reduced overall pressures by dispersing pressure over a larger area. This increase in total contact area was largely attributable to an increase in midfoot contact area, which is likely due to the profiled footbed of the FitFlop compared to the flat footbed in Havaiana. Also the soft thicker EVA construction in the midfoot region may sculpt under the foot as it is compressed, this shaping would likely help to alleviate the risks of plantar fasciitis attributed to the standard flat footbed in flip-flops (American College of Foot and Ankle Surgeons 2007). The increased midfoot contact area (19.9%) in the present study is smaller than differences previously recorded in participants' most comfortable footwear compared to walking barefoot (74.2%, Burnfield *et al.* 2004). However it is apparent that increased perceived comfort is consistently reported in footwear with increased midfoot contact areas (Che *et al.* 1994, Jordan *et al.* 1997), as apparent in FitFlop.

Magnitudes of plantar pressure in the present study are consistent with values previously reported in footwear with insoles in healthy populations (Bus *et al.* 2004), and slightly lower than some reported values for standard footwear (Perry *et al.* 1995, Burnfield *et al.* 2004). Generally results identified reduced peak pressures and pressure-time integrals in the FitFlop compared to the Havaiana. Peak pressure in the heel and pressure-time integral in the 1st MPJ were significantly reduced in FitFlop compared to Havaiana, these are regions of the foot where peak pressure variables have been related to comfort (Che *et al.* 1994, Jordan *et al.* 1997). Che *et al.* (1994) also identified the most comfortable shoes in their study produced more lateral CoP trajectories, a feature also evident in the FitFlop in the present results. Pressure reduction in the heel region has also been related to therapeutic benefits in clinical conditions (Bonanno *et al.* 2011), potentially explaining the numerous consumer testimonials reporting reduction of symptoms relating to localised pressure in the heel, such as plantar fasciitis (Testimonials at FitFlop, 2012). However, a full clinical study needs to be performed to determine whether symptoms can be related to localised peak pressures in FitFlop in this condition. The pressure in the hallux region showed no consistent differences between conditions, potentially attributable to the contact area being consistent across both conditions and the soles being the most similar at this region in terms of thickness and hardness (Table 3.14). Pressure differences between the two conditions in the present study are similar to those identified in previous research when a custom insole was inserted into a leather soled shoe (Bus *et al.* 2004). Reductions are less than those reported in the same foot regions in a trainer compared to a leather soled shoe, which range from 32-45% (Perry *et al.* 1995, Kästenbauer *et al.* 1998). This is as expected as the Havaiana has a sole which is thicker and contains more viscoelastic material than leather soled footwear and therefore would be expected to provide some reduction in plantar pressures. Carl and Barrett (2008) compared plantar pressures walking in a flip-flop to a trainer and barefoot. The methodology for data collection did not control walking speed and therefore comparison between conditions is difficult. Also the study did not report absolute values for the pressure recorded so a direct comparison cannot be made to the current data. Pressure time integrals in the present study differed from those reported in the literature, however most studies report data using automated calculations from Pedar software (Novel, Gmbh, Munich, Germany) which utilise peak pressure values and are not true integrals of the pressure curve for a region (Melai *et al.* 2011).

A common notion held with flip-flop footwear is that the individual "grips" the toes to hold the footwear in place which limits toe extension, similar to the mechanism in functional hallux limitus (Shroyer 2010). The pressure recorded in swing under the hallux in the flip-flop styles in the current study supports this proposed "gripping" concept. In kinematic studies hallux motion in flip-flops has been quantified with Chard and Smith (2011) identifying reductions in hallux dorsiflexion pre-heel-strike in walking compared to barefoot, supporting Carl and Barretts' (2008) proposal that the toes are used to grip the footbed. Similarly, Shroyer (2009) reported no differences in peak hallux extension in swing between flip-flops and barefoot. Peak hallux flexion or mean hallux flexion or extension throughout swing may have been more appropriate variables to quantify "gripping" in this context. In the present study pressure analysis was used with the aim of quantifying the magnitude and duration of pressure applied by the hallux on the footwear midsole during swing. It was hypothesised that the superior fit and thickness of the FitFlop upper would reduce the requirement for "gripping" in this footwear. Participants gripped for a shorter duration in the FitFlop and with less force during swing phase. "Gripping" with the toes in toe-post footwear may lead to overuse of the toe flexor muscles, and it is apparent that the thicker, higher fitting upper reduces the requirement to hold the footwear on in FitFlop and may reduce the incidence of overuse injuries and attenuate any toe-flexor injuries such as hallux limitus.

Limitations exist in this study with previous literature demonstrating that foot type can influence plantar loading (Chuckpaiwong *et al.* 2008), which was not assessed during this study, therefore application to the wider population must consider this. Other aspects of flip-flop use that have not been considered include the longevity of the material and the potential that flip-flops are often used after they are substantially worn and degraded and the properties of the footbed may be altered. The footwear used in the current study was new and therefore plantar pressures in the footwear may increase as the shoe becomes more worn, resulting in higher pressures than participants in the current study experienced. The study also limited analysis to specific regions of the plantar surface and data treatment utilised general masks such that anatomical regions were not adapted for individuals.

Conclusions

Despite the aforementioned limitations, it is apparent in the present study that the FitFlop can significantly reduce pressures in key locations related to comfort on the plantar tissue compared to Havaiana, which likely makes the FitFlop more comfortable. Additionally, the FitFlop reduces "gripping" with the hallux on the footbed in swing, therefore potentially reducing overuse injuries of the flexor muscles that are traditionally associated with toe-post footwear styles. Therefore as toe-post footwear styles remain popular, convenient and fashionable it is recommended that the FitFlop is an alternative to a standard flip-flop offering benefits to the wearer. Further work will consider the relationship of this reduced pressure to other variables in relation to footwear comfort.

Acknowledgements

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Conflict of interest statement

This research was funded through the Knowledge Transfer Partnership programme (KTP007228) which was co-funded by FitFlop Limited, UK and the Technology Strategy Board, UK. This work was undertaken with scientific diligence and the funders had no influence on the research design, protocol and the reporting of the results.

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3.2.6 Does flip-flop style footwear modify ankle biomechanics and foot loading patterns?

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Abstract

Background Flip-flops are item of footwear which are rubber and loosely secured across the dorsal fore-foot. These are a popular item of footwear in warm climates; however they are widely criticised for being detrimental to foot health and potentially causing modifications to gait. Commercially available alternatives exist including FitFlop^{TM}, which has a wider strap which fits the foot closer to the ankle and a thicker, multi-density midsole. The current study investigated gait modifications when wearing flip-flop style footwear compared to barefoot walking. **Methods** Testing was undertaken on 40 participants (20 male and 20 female, mean age 35.2±10.2 years, B.M.I 24.8±4.7 kg.m^{-2}). Kinematic, kinetic and electromyographic gait parameters were collected while subjects. Participants walked through a 3D capture volume over a force plate with the lower limbs defined using retro-reflective markers. Ankle angle in swing, frontal plane motion in stance and force loading rates at initial-contact were compared. Statistical analysis utilised ANOVA to compare differences between conditions. **Results** The footwear conditions altered kinematics compared to barefoot. Maximum ankle dorsiflexion in swing was greater in the flip-flop (7.6±2.6°, P=.004) and FitFlop (8.5±3.4°, P<.001) than barefoot (6.7±2.6°). Significantly higher tibialis anterior activation was measured in terminal swing in FitFlop (32.6%, P<.001) and flip-flop (31.2%, P<.001) compared to barefoot. The FitFlop reduced frontal plane ankle peak eversion during stance by 0.9±1.7° compared to walking in the flip-flop (-4.4±1.9°, P=.008) and barefoot (-4.3±2.1°, P =.032). A faster heel velocity toward the floor was evident in the FitFlop (-.326±.068 m.s^{-1}, P<.001) and flip-flop (-.342±.074 m.s^{-1}, P<.001) compared to barefoot (-.170±.065 m.s^{-1}). The FitFlop more effectively attenuated impact compared to the flip-flop, reducing the maximal instantaneous loading rate by 19% (P <.001). **Conclusions** Modifications to the sagittal plane ankle angle, frontal plane motion and characteristics of initial contact in barefoot walking occur in flip-flop footwear. The FitFlop may reduce risks traditionally associated with flip-flop footwear by reducing loading rate at heel-strike and frontal plane motion at the ankle during stance.

Keywords: flip-flop, gait, electromyography, loading rate

Background

Flip-flops are a popular summer shoe in the United Kingdom and commonly worn throughout the year in warmer climates such as India and Australia. The footwear style is defined by having one strap across the dorsal fore-foot which attaches to the footbed between the hallux and second toe. Despite the popularity of flip-flops, heel pain and other conditions such as overuse injuries of the tibialis anterior and toes are implicated for the wearers of flip-flop style footwear by podiatrists [1]. Flip-flops differ from standard types walking footwear design by a thin sole, no medial arch support, no protection for the toes, being loose fitting and having no pitch from heel to toe [2,3]. Despite the popularity of flip-flops in warm climates, limited scientific investigation into their influence on adult gait has been published in peer-reviewed literature.

Children wearing flip-flops display a trend towards a more dorsiflexed, everted and abducted midfoot during walking [4] and reduced hallux dorsiflexion prior to contact during walking and jogging in children, and reduced eversion during midstance in adults, compared to barefoot [5,6]. Video data has been used in an observational study of pedestrians, identifying a reduction in average walking speeds when walking in flip-flops [7], which was attributed to a shorter stride length compared to other footwear and confirmed in a laboratory environment [8]. In addition, an experimental study using 2D gait analysis concluded that there was increase in ankle plantarflexion during swing, which the authors speculated could be due to contraction of the toe flexors to keep the flip-flop on the foot due to the lack of heel-strap or full upper [8]. This hypothesis of an increased toe-clearance during swing is reinforced by Chard et al. [4], who identified greater ankle dorsiflexion prior just before and at heel contact when compared to barefoot conditions. Additionally, a plantar pressure study on females walking in flip-flops postulated that “gripping” with the toes occurs to hold this footwear on the foot, however the variable used to speculate this may not have been relevant [9]. This study referred to the shoe providing protection to the plantar surface of the foot, reducing plantar pressures compared to barefoot walking due to the material. In contrast, the impact attenuation that flip-flops provide at heel-strike has been investigated by Zhang et al, but no significant difference in loading rates between barefoot and flip-flops was evident [10]. However, the parameter used was not maximum instantaneous loading rate, which may have attributed to identified differences between conditions. Describing gait in flip-flops is useful, but more useful is comparing the influence of these styles of footwear to a relevant replacement for specific characteristics and variables, which are currently deemed to be detrimental. This acts as a starting point to enable footwear design modifications to the flip-flop footwear style to reduce gait modifications.

The FitFlop was originally developed to increase muscle activation in the lower limb by incorporating a soft mid-foot to induce instability within the midsole design. This footwear encompasses a thick multi-density EVA sole with a wider and higher fitting flip-flop style upper, these features may reduce detrimental gait modifications so making this footwear a more suitable alternative to a flip-flop. A recent paper identified that the FitFlop reduces plantar pressures in walking compared to a flip-flop [11]. However, gait motion analysis in this footwear compared to a flip-flop comparator is yet to be fully investigated

This current research study aims to compare barefoot walking to walking in flip-flop style footwear; and walking in FitFlop to walking in flip-flop to see if this contemporary footwear design offers a potential advantage in terms of gait modifications. Firstly, it is hypothesised that shod conditions will increase ankle dorsiflexion and tibialis anterior muscle activation, during swing and at heel strike compared to barefoot to hold the footwear during swing. However, this increase in dorsiflexion is predicted to be less in FitFlop than flip-flop due to the size and position of the dorsal strap. Secondly, there is expected to be reduction in the frontal plane motion of the foot in the FitFlop condition during stance compared to both barefoot and flip-flop due to the ergonomically profiled sole and provision and features of dorsal strap. Thirdly, shod conditions are expected to attenuate force loading rates around heel strike compared to barefoot due to the inclusion of material under the calcaneus; with the FitFlop having a greater reduction in loading rate compared to the flip-flop due to its greater sole thickness.

Methods

Participants

Forty participants took part in the study, twenty female and twenty male (Table 3.17), recruited from the University staff and student population. All indicated they were asymptomatic i.e. no diagnosed gait pathologies (past or present) and gave written informed consent fulfilling the requirements of the Ethics committee.

Table 3.17 Participant characteristics (mean±sd).

	Overall	Male	Female
Age (years)	35.2±10.2	32.7±9.0	37.7±10.9
Mass (kg)	72.5±15.2	81.6±12.8	63.5±11.8
Height (m)	1.71±0.09	1.78±0.07	1.64±0.05
Body Mass Index (B.M.I. kg.m^{-2})	24.8±4.7	26.1±4.7	23.6±4.4
U.K. Shoe size	7±2	9±1	6±1

Footwear conditions

Testing utilised barefoot and two footwear conditions: flip-flop and FitFlop (Figure 3.13, Table 3.18). The FitFlop varied between genders as a single style does not span the size range of participants. These variations included the last shape (used in the manufacture) and the dorsal strap material and the sole was thicker in the male version. The coverage and position of the dorsal strap to the foot however, were consistent between shoes. The male and female data was combined for comparison between conditions.

Figure 3.13 Footwear conditions tested: Havaiana flip-flop (a), Female FitFlop, Walkstar I (b) and Male FitFlop, Dass (c).

Table 3.18 Footwear characteristics for an example male and female shoe size from each condition

Condition	Style	Midsole Construction	Heel Depth (mm)	Hardness (Shore A)	Shoe Mass (g)
Flip-flop	Havaiana Brazil	EVA	Size UK 6 16 Size UK 9 18	33	Size UK 6 140 Size UK 9 172
FitFlop	Female = FitFlop Walkstar I. Male = Dass	Multi-density EVA in heel, midfoot and toe. Rubber outsole.	Size UK 6 33 Size UK 9 37	Heel: 55 Midfoot: 28 Toe 38	Size UK 6 172 Size UK 9 270

Where EVA is ethylene vinyl acetate.

Protocol

Participants undertook five walking trials for each condition, the order of which was randomised. Participants walked at their self-selected speed through the laboratory while kinematic, kinetic and Electromyography (EMG) data were recorded. Data from the right leg only was used for all analysis.

Kinematics and Kinetics

Three-dimensional motion data were collected using a 12 Camera Qualisys Opus system (Qualisys, Gothenburg, Sweden) sampling at 100 Hz. Spherical retro-reflective markers were positioned on the anatomical landmarks of the lower extremity to define the foot, leg, thigh, and pelvis. Rigid plates with reflective markers attached were fastened to the segments to enable the CAST technique to be utilised [12]. This technique uses a static trial recorded at outset, which enables the rigid plates to be defined relative to the anatomical landmarks for each segment. Joint angles were defined such that a static posture was zero and ankle dorsiflexion and was positive. Kinetic data were collected simultaneously using two force platforms (Advanced Medical Technologies Inc., Newton, Massachusetts, USA) at 3000 Hz. These data were exported to Visual 3D (C-Motion Inc., Rockville, Maryland, USA) where kinematic and kinetic data were filtered using second-order Butterworth filters at 10 and 25 Hz respectively.

Visual 3D software was utilised to build a six degree-of-freedom model of the lower limbs. A pre-written pipeline was used to calculate kinematic and kinetic variables including joint angles and internal joint moments for the right leg normalised to body weight and gait cycle time where appropriate. Heel strike was defined at the point where the vertical GRF exceeded 10 N [13]. Peak values and magnitudes at heel strike and toe-off for relevant kinematic variables were identified for statistical analysis. GRF impulse and maximum instantaneous loading rate from heel-strike to 65ms were calculated to enable comparison of trials with and without heel-strike transients consistently [14,15]. Walking speed was computed from kinematic data within Visual 3D.

Electromyography

28 participants from the 40 were tested for muscle activation (mean±1 standard deviation, Male, N=15, age=30±8 years, B.M.I=25.9±4.5 $kg.m^{-2}$; Female, N=13, age=37.8±12.4 years, B.M.I=23.0±4.7 $kg.m^{-2}$) due to technical difficulties. EMG was recorded simultaneously at 3000 Hz using bipolar surface Ag/AgCl electrodes (Noraxon Inc, Scottsdale, Arizona, USA), with an electrode diameter of 10 mm and an inter-electrode spacing of 20 mm. Prior to electrode placement, hair was removed, skin exfoliated and cleaned. Electrodes were placed in accordance with the SENIAM recommendations on the tibialis anterior and peroneus longus [16]. The ground electrode was placed overlying the medial condyle of the tibia.

EMG analysis was undertaken in Visual 3D. Data was filtered to remove zero-offset (high-pass Butterworth 20 Hz) and a linear envelope was produced using a 10 Hz low-pass Butterworth filter. The linear envelope EMG was integrated (EMGLI) within the Gait Cycle events: initial contact (0–2%), loading response (0–10%), midstance (10–30%), terminal stance (30–50%), pre-swing (50–60%), initial swing (60–73%), mid-swing (73–87%) and terminal swing (87–100%) [17]. The gait cycle phases relating to the specific hypothesis above were compared only for the specific muscles.

Statistical Analysis

Variables were calculated on an individual participant basis from non-normalised data for statistical comparison averaged across trials then participants to produce an individual then group mean (±1 standard deviation). Figures present ensemble average data normalised to gait cycle/stance time. SPSS (Version 17.0, SPSS Inc., Chicago, U.S.A.) software was utilised for statistical testing specific to study hypotheses, utilising between participants analysis of variance (ANOVA) to identify differences between conditions. Data that was not normally distributed (electromyography and joint moments) was square-root transformed, checked for normality, and treated as parametric. Holm adjustment for multiple comparisons was used and effect size (Cohen's d, *d*) was reported for significant differences.

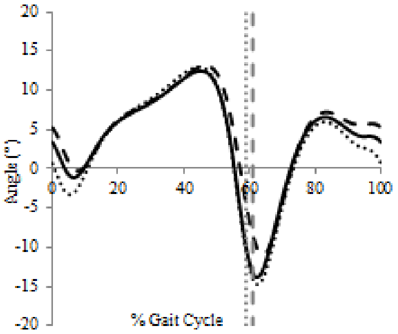
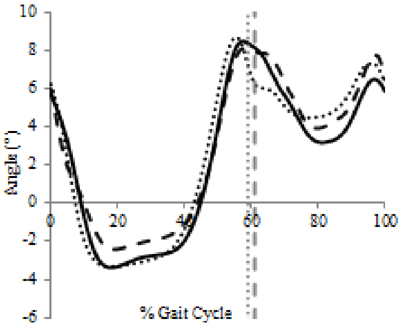
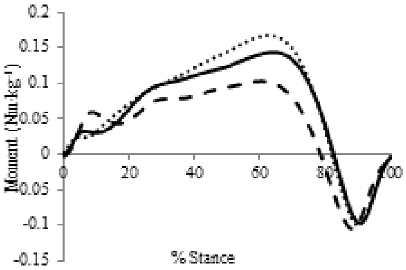
Results

Ankle angle swing

Ankle joint angles and muscle activation differed between conditions in the sagittal plane (Figure 2, 3). In the sagittal plane, peak dorsiflexion and plantarflexion angles in swing differed significantly between conditions with small to large effect sizes. The flip-flop style footwear recorded greater (FitFlop 8.5±3.4°, flip-flop 7.6±2.6°, barefoot 6.7±2.6°; Figure 2) maximum dorsiflexion values and reduced maximum plantarflexion angles (FitFlop -12.4±4.4°, flip-flop -15.4±5.1°, barefoot -16.8±4.7°; Figure 2) compared to barefoot. Muscle activation measurement recorded significantly higher tibialis anterior activation in terminal swing in FitFlop (mean 32.6%, $P<.001$, $d=-0.11$) and flip-flop (31.2%, $P<.001$, $d=0.27$) compared to barefoot.

Frontal plane ankle in stance

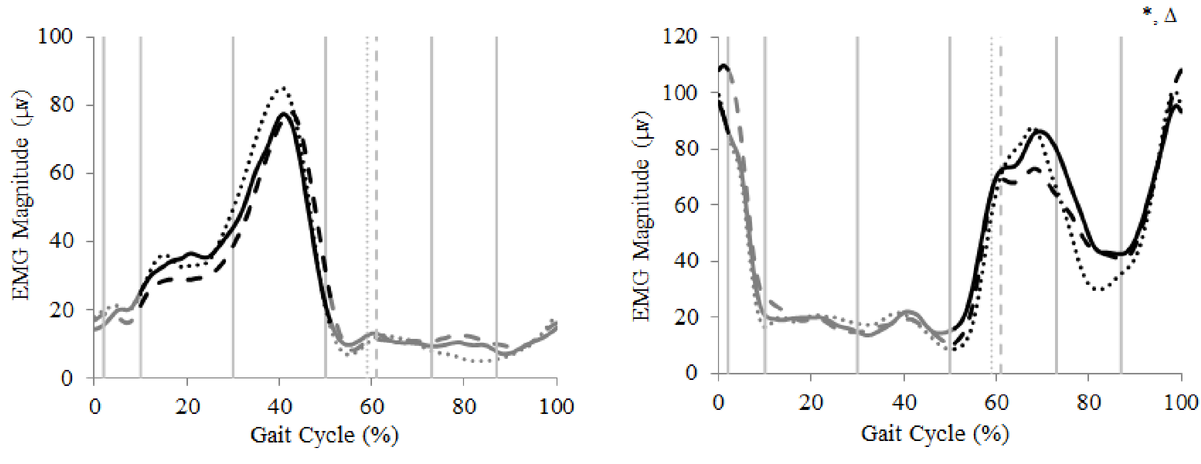
In the frontal plane the FitFlop reduced the range-of-motion compared to the other conditions by approximately 10% (Figure 3.14). Maximum eversion was significantly lower in FitFlop (-3.5±2.2°) compared to in the flip-flop (-4.4±1.9°, $P=.008$, $d=-0.44$) and barefoot (-4.3±2.1°, $P=.032$, $d=-0.37$) conditions (Figure 3.14). Alongside alterations in joint angle, significantly lower inversion moment was recorded during late stance in FitFlop than flip-flop ($P<.001$, $d=1.04$) and barefoot ($P<.001$, $d=1.36$) (Figure 3.14). Peroneus longus muscle activation did not differ between conditions during stance (Figure 3.15).

			Barefoot	flip-flop	FitFlop	Significance (Cohen's *d*)
Ankle Angle (Sagittal) Dorsiflexion positive		Heel-strike (°)	1.6±2.7	4.1±3.	5.7±3.6	∆ (-0.88) * (-1.29) □ (-0.48)
		Maximum dorsiflexion stance (°)	13.5±2.4	13.0±2.9	13.7±2.7	∆ (0.19) □ (-0.25)
		Maximum dorsiflexion swing (°)	6.7±2.6	7.6±2.6	8.5±3.4	∆ (-0.35) * (-0.60) □ (-0.30)
		Toe-off (°)	-12.6±4.3	-11.8±4.5	-8.9±4.0	* (-0.93) □ (-0.68)
		Maximum plantarflexion Swing (°)	-16.8±4.7	-15.4±5.1	-12.4±4.4	□ (-0.29) * (-0.97) ∆ (-0.63)
Ankle Angle (Frontal) Inversion positive		Heel-strike (°)	6.2±3.3	5.5±3.1	6.5±3.5	
		Maximum inversion stance (°)	10.8±3.9	10.7±4.3	10.3±3.5	
		Maximum eversion stance (°)	-4.3±2.1	-4.4±1.9	-3.5±2.2	* (-0.37) □ (-0.44)
		Toe-off (°)	6.8±5.9	8.5±6.1	8.3±4.7	∆ (-0.28) * (-0.28)
		Maximum inversion swing (°)	10.8±4.5	11.4±5.0	11.5±4.0	
		Minimum inversion swing (°)	1.6±4.3	1.0±4.1	1.8±4.1	
Ankle Moment (Frontal) Inversion positive		Max inversion moment ($Nm \cdot kg^{-1}$)	0.194± 0.107	0.185± 0.090	0.142± 0.068	* (0.58) □ (0.54)
		Max eversion moment ($Nm \cdot kg^{-1}$)	-0.122± 0.103	-0.122± 0.090	-0.129± 0.093	

- - - FitFlop, — flip-flop, ···· barefoot

Figure 3.14 Ensemble average ankle kinematics and kinetics.

Kinematics normalised to the gait cycle, kinetics normalised to stance time, where vertical lines denote toe-off. With values calculated prior to normalisation and presented as mean±*sd*.
Statistical significance denoted by:
*barefoot significantly different to FitFlop,
∆ barefoot significantly different to flip-flop
□FitFlop significantly different to flip-flop.



a) b)

--- FitFlop, — flip-flop, ···· barefoot.

Figure 3.15 Mean of all participant (N = 28) data for electromyography linear envelope (µv) normalised to the gait cycle.

For the a) peroneus longusb) tibialis anterior.
Vertical dashed lines denote toe-off and black highlighs regions that were compared statistically. Statistical significance denoted by:
*barefoot significantly different to FitFlop,
Δ barefoot significantly different to flip-flop
□FitFlop significantly different to flip-flop.

Loading rate

The nature of the impact with floor differed between the three conditions with significant differences evident in the GRF variables as well as vertical heel velocities toward the floor. The maximum loading rate significantly differed between conditions with both the flip-flop (26.7±5.6 BW·s^{-1}) and FitFlop (21.7±5.4 BW·s^{-1}) providing significant reductions compared to Barefoot (41.4±22.9 BW·s^{-1}P<.001, d>0.91). Furthermore, the FitFlop condition reduced loading rate by 19% compared to flip-flop (P<.001, d= 0.88). The impulse of the vertical GRF from heel-strike to 65 ms was significantly lower in FitFlop (.029±.006 BW·s, P<.001, d=1.09) and flip-flop (.034±.005 BW·s, P=.032, d=0.20) than barefoot (.035±.005 BW·s). These were despite the walking speed being significantly higher in the FitFlop (1.32±.10 m.s^{-1}) compared to the flip-flop (1.29±.11 m.s^{-1}) condition. The vertical heel velocity at heel-strike was significantly faster toward the ground in both flip-flop (-.342±.074 m·s^{-1}, P<.001, d=2.5) and FitFlop (-.326±.068 m·s^{-1}, P<.001, d=2.3) than the barefoot condition (-.170±.065 m·s^{-1}).

Discussion

This study has undertaken an assessment of the biomechanics of gait when walking in flip-flop style footwear and compared it to barefoot walking and a contemporary version of this style of footwear: the FitFlop. The research has highlighted statistically significant differences in ankle angle in swing, frontal plane motion and loading rate gait parameters, while walking in a flip-flop and FitFlop compared to barefoot walking. Contrasting gait patterns are evident in the FitFlop, which may pose advantages to the wearer of flip-flop style footwear. Walking speeds in all conditions in the present study were comparable to the speed that people walk in flip-flops in their daily lives (1.31 $m.s^{-1}$) [7]. This suggests that results are generalizable to adults walking in flip-flop and FitFlop footwear in a real-world environment.

Ankle angle swing

As hypothesised, the shod conditions demonstrated moderations to sagittal plane motion at the ankle joint motion at heel-strike, toe-off and during swing compared to barefoot. This trend toward dorsiflexion, or reduced plantarflexion, in shod conditions is consistent with previous literature [4], particularly during swing, and is potentially a mechanism to keep the shoes on the foot. The dorsal strap for both conditions of footwear only cover the front of the foot and thus, gait may be adapted to hold the shoe on the foot [8]. In contrast to the current study, Shroyer et al., identified increased plantarflexion in swing when participants wore flip-flops compared to trainers [8]. The authors attributed their finding to contraction of the toe flexors to hold the flip-flop, creating a plantar-flexor moment at the ankle, however no electromyography data was collected. A previous study, however, discounted gripping of the flip-flop by the hallux in swing to control the flip-flop [9], although the plantar pressure analysis undertaken may not have been sensitive to pressures in swing. Contrasting these findings, a recent plantar pressure analysis identified gripping in swing in both flip-flops and FitFlops [11]. The FitFlop demonstrated reductions in magnitude and duration of gripping [11], potentially reducing any resultant plantar-flexor moment at the ankle, allowing greater dorsiflexion compared to the flip-flop. Inferences from the current data and literature imply that ankle dorsiflexion and toe flexion may combine to hold toe-post footwear on the foot during swing, however toe motion must be quantified to confirm this. Contradicting the original hypothesis, the FitFlop increased dorsiflexion in swing and tibialis anterior activation compared to flip-flop, as opposed to reducing this mechanism. This may be due to the aforementioned reduced toe-flexor moment, or the increased mass and thicker sole of this shoe requiring greater ground clearance than the flip-flop condition. Results from the present study demonstrate significantly higher tibialis anterior activation in shod conditions than barefoot in terminal swing, consistent with the increase in dorsiflexion in swing and reports from other authors [10].

Frontal plane ankle in stance

As anticipated, in the frontal plane the FitFlop reduced the joint excursion compared to other conditions by approximately 10%, in particular eversion was reduced during stance (Figure 1). The flip-flop condition showed consistent patterns to barefoot as would be expected with a flat, flexible sole and a thin, loose fitting upper (Figure 3.13), foot motion is unchanged during stance [19]. Previous studies are inconsistent reporting increased midfoot eversion [4], no significant differences in frontal plane range-of-motion [10] and reduced eversion in midstance in a flip-flop compared to barefoot [6]. The FitFlop thicker upper and soft profiled footbed therefore appear to interact to control frontal plane motion of the ankle and tarsal joints. This may be potentially beneficial to wearers and may in-part, explain positive testimonials from consumers as excessive frontal plane motion of the ankle has been repeatedly linked to overuse injuries [20.21]. A significantly reduced inversion moment was recorded in FitFlop throughout stance, with a significantly lower peak moment in terminal stance than both flip-flop and barefoot. This reduction may be attributed to the less everted foot position reducing the distance between the GRF and the ankle joint centre [21]. Increased ankle external eversion moments have been linked to increased injury potential in running [22].

The flip-flop peak inversion moment in the current study was equivalent to that in barefoot. This contrasts previous research which has identified increased maximum external inversion moment in a flip-flop compared to other footwear conditions and barefoot [23]. This may have been due to pathology related motion present in one of the previous studies knee osteoarthritic population [23]. Similarly, another study reports reduced peak ankle inversion moment in late stance in flip-flops compared to barefoot in a male population [10]. This may be a function of gender, familiarity with the footwear or differences between the specific styles utilised in the study. Gender differences have previously been identified in sagittal ankle angle walking in flip-flops [8] and future work should consider both gender differences and footwear style familiarity differences to clarify any gait modifications in flip-flop style footwear in specific groups. This is a limitation of the current research, as any evident differences between genders could not be isolated to gender alone as opposed to footwear differences in the styles, interactions were not compared.

Loading rate

The flip-flop was expected to reduce loading rate at ground impact compared to barefoot and the FitFlop was expected to further reduce this loading, which was confirmed. Analysis of the GRF was designed to allow comparison of heel strike loading features when not all trials included a transient feature. Loading rates quantified in the present study were consistent with previous literature for the shod values, barefoot values were lower than the 117.8±27.5 $BW.s^{-1}$ reported in previous literature; however this literature utilised a fixed walking speed of $1.5m.s^{-1}$, faster than the current study [14,15].

Velocity of the heel toward the floor was twice as fast in the flip-flop style conditions compared to the barefoot, consistent with previous findings in flip-flops and sandals [23,24]. Explanation of increased heel velocity in this shoe style may be proprioceptive due to the shoe leaving the foot at the heel or due to protective kinematic adaptations in barefoot gait to reduce impact energy, as evident in running [25]. Despite the higher heel velocity and therefore higher impact energy in both flip-flop style conditions, force loading rate was lower than in barefoot. This contrasts previous research which reports no difference in loading rate between barefoot and flip-flop conditions, but calculated loading rate to loading peak of GRF as opposed to the maximum instantaneous value [10]. This may have masked differences at heel-contact due to the loading peak of the GRF largely being a function of body mass, kinematics and walking velocity as opposed to features of the footwear or heel velocity. The flip-flop would be expected to attenuate shock at initial contact compared to barefoot as a layer of EVA is placed under the foot. Viscoelastic material absorbs energy and therefore can attenuate the foot impact with the floor [26,27], whereas the barefoot condition only has the internal structures of the foot as protective materials. A plantar pressure study has suggested that results show the flip-flop protecting the body at heel-strike compared to barefoot.[9] The FitFlop absorbed greater shock at heel-strike, evident by 19% and 15% reductions in loading rate and impulse compared to flip-flop, with strong effect. This is likely due to the thicker construction of EVA in the heel section of the FitFlop compared to flip-flop (Table 3.18). Despite the flip-flop having a softer EVA construction (Table 3.18), the thickness of the EVA is the most important factor when considering shock absorption properties [25]. Reduced loading rate of the ground reaction force likely reduces the potential for skeletal injury during walking [26,27].

Conclusions

The current study identified increased ankle dorsiflexor activity in flip-flop style footwear compared to barefoot, coupled with increased dorsiflexion in swing, assumed to be a mechanism to hold the shoe on the foot. The FitFlop limited foot motion in the frontal plane and significantly reduced loading at impact, compared to flip-flop and barefoot. However, it is not clear whether these reductions are enough to reduce any potential injury or overuse injuries associated with flip-flop footwear and further, longitudinal, research would be needed to clarify this relationship.

Competing Interests

The research has been co-funded by the Technology Strategy Board and FitFlop ltd as part of the Knowledge Transfer Partnership programme (KTP007228). The primary author worked on the programme, supervised by RJ and PGS. VA and AF are not specifically associated with the programme. Data was collected, analysed and the paper written with no influence from the funders and no author will receive anything of value from the commercial products included in this paper.

Authors' Contributions

CP designed the study, collected and analysed data, drafted and critically revised the manuscript. VA acquired and processed data for the paper, was involved in the drafting process and approved the final draft. RK, AF and PGS were involved in the study design and conception in addition to critically revising the final manuscript and approving the final draft.

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3.2.7 Subjective and objective variables to quantify comfort in walking footwear.

Price, C & Jones, R. (2013).

To submit.

Comfort is one of the most important aspects for a consumer when purchasing footwear. It is user and situation specific and therefore must be measured with relevant protocols, on relevant consumers. The study was undertaken to present a protocol to quantify and compare comfort in walking footwear using subjective and objective measures with the aim of providing useful data in the design process. A modified version of a regularly used running footwear questionnaire was developed. This was used to compare comfort in walking in two different walking shoes (C and S) The purpose was to develop a protocol which could be used to subjectively and objectively compare comfort in walking shoes for relevant consumers and shoes. *Methods*: 40 female subjects were tested walking on a treadmill in two covered footwear styles while plantar pressure, peak axial tibial acceleration and questionnaire data were collected. Comfort was reported on the modified questionnaire. Peak and regional pressures, peak axial tibial acceleration and comfort scores were compared between shoes. The relationship between objective and subjective methods for specific anatomical regions was explored using correlations. *Results:* Both subjective and objective comfort measures differed significantly between shoes. Questionnaire data demonstrated increased comfort perception under the ball of the foot (+19 comfort points, p=.016) and increased perception of cushioning under the heel (+12 comfort points, p=.001) in shoe S. All anatomical regions demonstrated significantly different pressures between shoes. Pressure in the metatarsals decreased in shoe S (p<.001) and medial midfoot pressure increased (p<.001). Correlation of subjective and objective difference scores identified one significant correlation between maximum heel pressure and perceived cushioning in the heel (r=.341, p=.039). As a decrease in pressure in condition S occurred compared to C, there was an increase in perceived cushioning in S compared to C. *Conclusions:* This work has demonstrated a holistic protocol for footwear companies to quantify and benchmark walking footwear comfort from subjective and objective sources. The use of this methodology could be continued with footwear products throughout the design process to adapt design features for comfort improvement.

1. Introduction

Comfort is an important factor in the decision of a consumer to purchase footwear. It is subjective and therefore influenced by an interaction of psychological, physiological and mechanical factors (Alcántara et al., 2005; Au and Goonetilleke, 2007; Mills et al., 2010). Literature has reported perceptions of comfort being affected by material properties of the upper and footbed (Jordan et al., 1997; Yung-Hui and Wei-Hsien, 2005), consumer characteristics such as anthropometry (Wunderlich and Cavanagh, 2001), activity undertaken (Miller et al., 2000) and shoe dimensions and fit (Au and Goonetilleke, 2007; Hagen et al., 2010).

Objective measures of comfort have been considered extensively in human factors literature, identifying that pressure distribution demonstrates the clearest association with subjective ratings of comfort (De Looze et al., 2003). In footwear research, plantar and dorsal pressure measurement have both been utilised in studies relating to footwear comfort (Jordan et al., 1997; Wegener et al., 2008). Increased peak plantar pressures in the rearfoot and forefoot and decreased contact areas have been associated with footwear conditions that have been rated least comfortable (Che et al., 1994; Jordan et al., 1997; Witana et al., 2009). Che et al. (1994) related subjective comfort to plantar pressures, comparing each participants' most and least comfortable insole. In walking there were significantly higher peak pressures in the midfoot (+16.5%) and lower in the medial forefoot (-16%) and hallux (-23%) in the most comfortable compared to the least comfortable insole. Shock absorption has also been considered by researchers investigating comfort and cushioning perception in running (Hennig et al., 1996a; Lake and Lafortune, 1998). Lake and Lafortune (1998) assessed the link between perceptions of impact severity and force loading rate using a human pendulum protocol. The outcome variables demonstrated that perception of impact severity increased with faster impacts and harder surfaces, significantly correlating with the biomechanical variables. Similarly, Hennig et al. (1996a) related reported comfort to peak force loading and peak pressure in the heel during running in different midsole constructions. Maximum rate of force loading ($r^2 = 0.89$) and peak pressures in the heel ($r^2 = 0.97$) increased with perceptions of less cushioning in the three running shoes tested.

Methods used to quantify subjective comfort are generally scales (Au and Goonetilleke, 2007; Mündermann et al., 2002), or ranking shoes in order of preference (Che et al., 1994; Mills et al., 2010). Mündermann et al. (2002) developed, validated, and assessed the reliability of a comfort protocol and scale. This tool was validated by assessing comfort of insoles in running shoes on a healthy running population. The protocol utilised a control trainer as every other condition for a stable baseline and calculated the mean results from sessions 4-6 to compare comfort (Mündermann et al., 2002). Despite this specific protocol, the scale, and slightly adapted versions have been commonly and widely used in a range of footwear studies including walking (Davis et al., 2008), heeled shoes (Yung-Hui and Wei-Hsien, 2005) and on specific population groups (Wegener et al., 2008). Furthermore, the questionnaire has been applied to compare numerous footwear conditions without the recommended control condition to provide a stable baseline between conditions (Davis et al., 2008; Wegener et al., 2008). The application of this scale to shoes that are not running shoes and consumers who are not runners is questionable. The specific terminology relates to running shoes and consequently answers may be misleading with other footwear. The terminology within the current scale may not be relevant or clear to the general consumer (for example: "medial-lateral control") and has previously caused confusion for participants in other research projects within this department. It is therefore evident that the moderation and re-validation of such a scale may be more appropriate for measuring subjective comfort in a walking shoe in the general population.

The purpose of this study was to present a methodology to assess comfort in walking footwear for benchmarking and testing purposes in footwear development. The process will modify a comfort questionnaire to be wearer and footwear specific. Following this, the questionnaire will be included as part of a protocol to objectively and subjectively quantify comfort in two commercially available shoes for walking. The relationship between objective and subjective comfort measures for specific aspects of the footwear styles will then be explored.

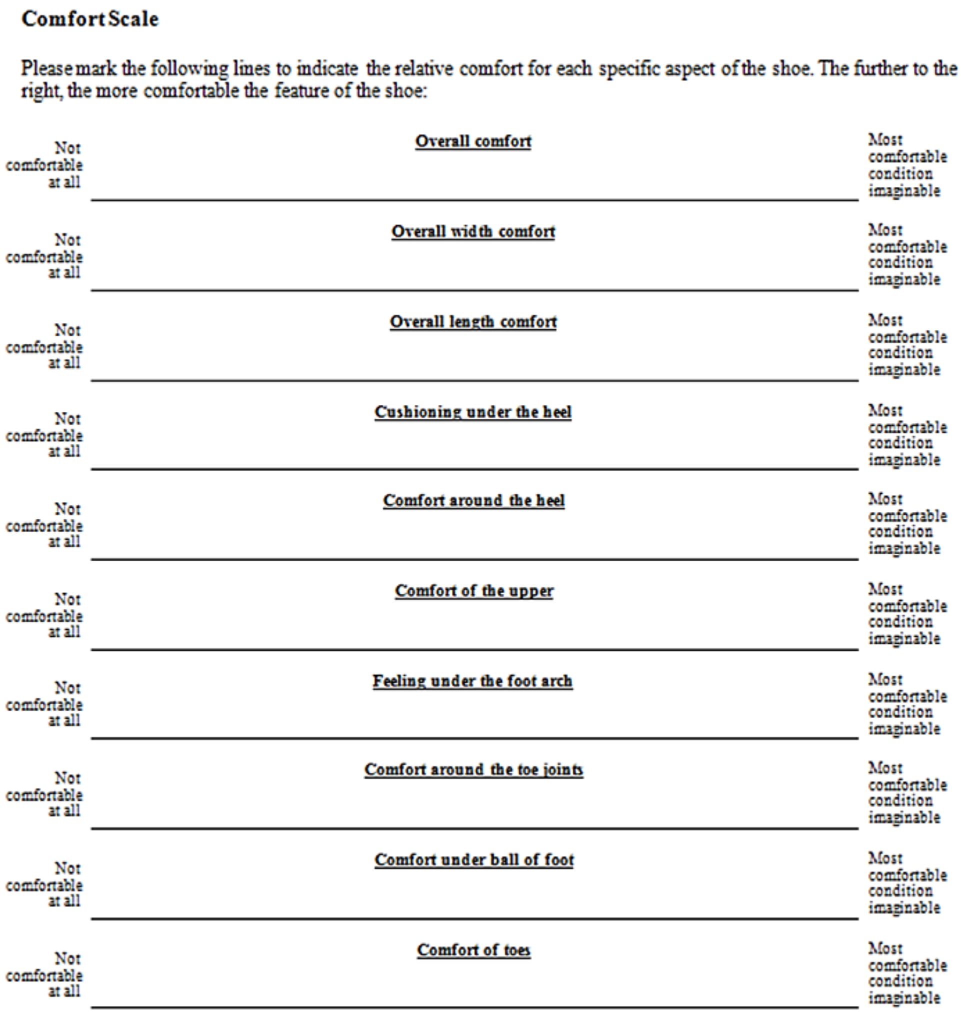
2. Methods

The study was split into two distinct phases: Phase one modified the questionnaire and explored its repeatability. Phase two of the study utilised the modified questionnaire within a protocol to compare comfort subjectively and objectively in two commercially available shoes for walking. All participants were recruited from University staff and student populations and gave written informed consent fulfilling the requirements of the University Ethics Committee.

2.1 Phase One

2.1.1. Development

10 female participants (mean±1 standard deviation: age 39±11 years) were interviewed to determine the terminology that they would use to refer to specific anatomical locations on the foot and lower leg as well as features of footwear. A flat covered shoe was used as a sample to determine how participants would refer to specific parts of the shoe. A simple tally chart was utilised to construct a list of commonly used phrases by the participants. These phrases were then compiled and used to modify some aspects of the Mündermann et al. (2002) comfort scale, which had previously caused confusion in study participants (Figure 3.16).



Comfort Scale

Please mark the following lines to indicate the relative comfort for each specific aspect of the shoe. The further to the right, the more comfortable the feature of the shoe:

Not comfortable at all	**Overall comfort**	Most comfortable condition imaginable
Not comfortable at all	**Overall width comfort**	Most comfortable condition imaginable
Not comfortable at all	**Overall length comfort**	Most comfortable condition imaginable
Not comfortable at all	**Cushioning under the heel**	Most comfortable condition imaginable
Not comfortable at all	**Comfort around the heel**	Most comfortable condition imaginable
Not comfortable at all	**Comfort of the upper**	Most comfortable condition imaginable
Not comfortable at all	**Feeling under the foot arch**	Most comfortable condition imaginable
Not comfortable at all	**Comfort around the toe joints**	Most comfortable condition imaginable
Not comfortable at all	**Comfort under ball of foot**	Most comfortable condition imaginable
Not comfortable at all	**Comfort of toes**	Most comfortable condition imaginable

Figure 3.16 Modified comfort visual analogue scale.

Note that the scales are 150mm in length when completed in the study. Original scales read “overall comfort”, “heel cushioning”, “forefoot cushioning”, “arch height”, “heel cup fit”, “shoe heel width”, “shoe forefoot width” and “shoe length”.

2.1.2. Repeatability

The repeatability of the modified scale was established by 5 participants (mean±1 standard deviation: age 24.8±3.1 years, mass 62.8±3.0 kg and height 1.66±0.08 m) completing the questionnaire in 3 shoes (one shoe rated twice each visit) twice in one day (4 hours apart) and again the next day. Repeatability was established with statistical analysis of the within-session, between-session and between-day scores for each participant and each aspect of the comfort scale. Combining all participant data produced intra-class correlation coefficients of 0.778-0.967 between-session for the different scales making up the questionnaire. A minimal important difference to denote change in comfort was estimated at 10.1±3.3 (range 5.2-14.1), comparable with the 10.2±6.1 (range 5.0-25.0) identified by (Mills et al., 2010b) with a participant-nominated approach.

2.2 Phase Two

2.2.1. Footwear Conditions

Two footwear conditions were tested (C and S) (Table 3.19, Figure 3.17). Participants did not have the shoe they were walking in described during the testing and shoes were chosen deliberately to look similar in attempt to minimise any potential influence of appearance on findings (Alcántara et al, 2005). After all participants had completed the trials, the footwear was tested for shock absorption properties utilising a protocol developed for walking footwear (Price et al., 2014) (Table 3.19).

Figure 3.17 The two footwear conditions utilised for the footwear comparison: Shoe S (left) and Shoe C (right).

2.2.2 Participant

40 female participants were recruited (mean±1 standard deviation age of 40±14 years, mass of 63.2±9.0 kg and height of 1.63±0.06 m). Participants wore shoe size UK 5 (N=26), 6 (N=10) or 7 (N=4).

Table 3.19 Footwear features for the two test conditions compared for the size 5 condition.

Condition	S	C
Upper Construction	Leather-lined canvas	Canvas
Midsole Construction	Multi-density EVA	Rubber Sole
Heel Sole Depth (mm)	41	19
Heel Hardness (Shore A)	45	48
Forefoot Sole Depth (mm)	19	19
Forefoot Hardness (Shore A)	30	48
Shoe Heel Breadth (mm)	70	69
Mass (g)	307	388
Peak g in drop test	2.2	3.3
Style name	FitFlop SuperT	Converse All-Star

Where EVA = Ethylene Vinyl Acetate and Shore A hardness was measured prior to testing using a Shore A durometer and at the factory using a custom-made device with a larger foot.

2.2.3 Protocol

Prior to testing, participants' feet were measured for joint and instep girth and length to ensure that subjects wore the correct shoe size. Participants were all familiar with treadmill walking and practiced walking on the treadmill and established their self-selected speed, which was set and used throughout the study. The testing protocol consisted of two stages; the questionnaire/accelerometer stage and the plantar pressure stage, each consisting of a 2 minute walk. This was structured so that the comfort questionnaire could be completed without the pressure insoles influencing the footwear fit and comfort (Che et al., 1994). Footwear condition order was randomised. To complete the questionnaire/accelerometer stage, an accelerometer (Noraxon Inc, Scottsdale, Arizona, USA) was fastened to the tibia of the subjects with double-sided tape, and then wrapped with a bandaged as tight as possible without causing discomfort. Data was collected via wireless telemetry (1200 Hz) in the MyoResearch software (Noraxon Inc, Scottsdale, Arizona, USA). Following walking with the accelerometer in each footwear condition the subjects sat down and completed the modified questionnaire (Figure 3.16) without removing the footwear. A baseline condition of the participants own walking shoe was used prior to the test conditions to act as a stable baseline to complete the questionnaire, consistent with the protocol recommended by Mündermann et al. (2002). Plantar pressure data was then collected utilising the Medilogic (T&T Medilogic, Gmbh, Germany) in-shoe pressure measurement system (100 Hz). Participants were instrumented with pressure insoles in the footwear while a transmitter (fastened around their waist) wirelessly transferred data to a laptop. Pressure insoles were secured in the footwear with small squares of double-sided tape positioned at the toe, mid-foot and heel.

2.4 Data Treatment

Accelerometer and pressure data sets were analysed using custom-written Matlab scripts, analysing all complete steps in the 2 minutes of recorded data (mean N = 104±12). Analysed variables were calculated only for the right foot for comparison (Menz 2004). Peak axial tibial acceleration (PA) was calculated from the accelerometer data. Peak pressures (PP) were calculated from the plantar pressure sensors associated with specific foot regions (Figure 3.18). This computed both the regional and single-sensor peak pressures for the region. Regional pressure (RP) grouped all sensors from a region and was included alongside the more traditionally reported single sensors as it was deemed potentially more relevant to comfort sensation. Total foot and regional contact area for the medial midfoot were also computed. Questionnaire scales were measured, reported as 'comfort points' (from 0-150) and data entered into SPSS (V17, SPSS Inc., Chicago, U.S.A.) for analysis.

Statistical comparisons were undertaken in SPSS, utilising Wilcoxon signed-rank test as data was non-parametric, identified by Kolmogorov-Smirnov tests and examination of box-plots. Results are therefore presented as median±inter-quartile range, with $p<0.05$ chosen to denote significance. Percentage data was log transformed prior to statistical comparison. As a measure of the magnitude of effect, the Cliff's delta (δ) was computed, a measure of the probability of the score in one shoe being higher than the other (Cliff, 1993). This represents the probability that a wearer will find one shoe more comfortable than another for the specific feature. Correlations were undertaken to compare objective and subjective measures of comfort. The individual differences between the two footwear conditions were quantified and compared to remove any individual-subject bias to using different lengths of the scale. Additionally, a "Distribution score" was calculated to determine whether the relationship between objective and subjective measures followed the anticipated distribution as opposed to whether the relationship was linear or not e.g. as pressures decreased comfort increased.

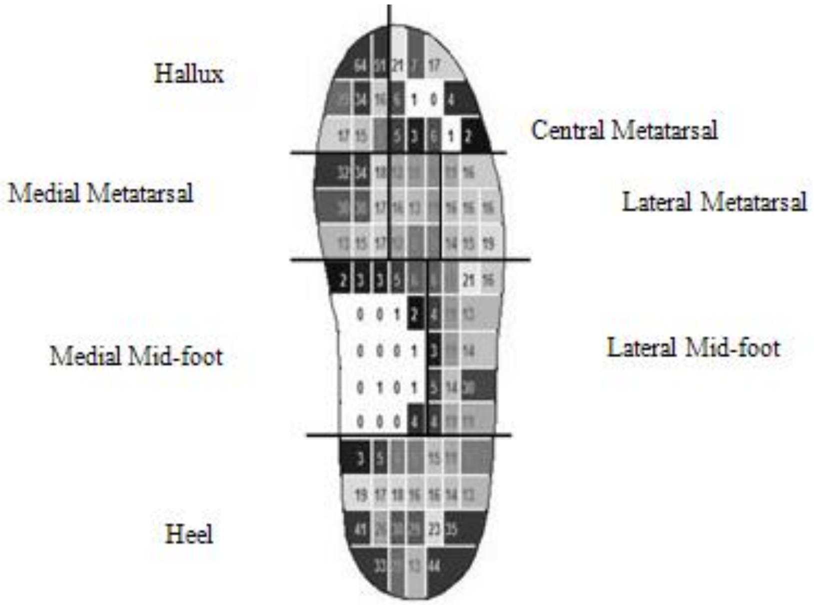


Figure 3.18 Anatomical regions defined on the Medilogic insole utilised for the study.

Based on an adapted version of the Pedar prc mask and the foot divisions described by Cavanagh and Ulbrecht (1994) and Hennig (1993).

3. Results

Following data analysis for outliers (using box-plots) in objective and subjective measures, 37 participants remained (age 39±13 years, mass of 64.0±7.3 kg and height of 1.63±0.06 m).

3.1 Footwear condition comparison

The questionnaire data demonstrated two significant differences between the footwear conditions tested for subjective measures of comfort. Two scales demonstrated significantly higher scores for condition S; "cushioning under the heel" and "comfort under the ball of the foot" (Table 3.20). Both differences exceeded the minimal difference established in the repeatability testing of the questionnaire, however the Cliff's delta demonstrated relatively large overlap between the data-sets (≈0.3).

Table 3.20 Absolute comfort scores (where maximum is 150) for footwear tested.

Questionnaire Scale	**S**	**C**	***p* value**	δ
Overall Comfort	78 (60-106)	76 (58-99)	*.814*	0.03
Overall Width Comfort	76 (61-102)	77 (49-100)	*.980*	0.04
Overall length comfort	77 (75-95)	76 (57-101)	*.789*	-0.08
Cushioning under the heel	94 (78-120)	75 (45-95)	* *.001*	-0.39
Comfort around the heel	90 (65-102)	78 (70-101)	*.502*	0.07
Comfort of the upper	75 (59-100)	75 (52-110)	*.987*	0.05
Feeling under the foot arch	77 (67-102)	75 (39-92)	*.080*	-0.19
Comfort around the toe joints	76 (54-102)	75 (49-103)	*.973*	0.04
Comfort under the ball of the foot	88 (75-100)	76 (42-90)	* *.016*	-0.26
Comfort of toes	86 (66-110)	76 (49-100)	*.392*	-0.05

Median values are presented (25th and 75th percentile), where * indicates statistically significant differences alongside *p* value and Cliff's delta δ.

There were significant differences identified in the in-shoe plantar pressure measures between conditions and Cliff's delta values identified substantial to minimal overlap in the data-sets (Table 3.21). No significant difference existed between the Hallux and two mid-foot regions for the standard PP values, however RP, where sensors were summed, were significantly higher in shoe S. PA recorded in the current study did not differ significantly between the two footwear conditions tested (Table 3.21).

Table 3.21 Plantar pressure and contact area results for the two tested footwear conditions

Variable	Region	S	C	*p* value	*δ*
Peak Regional Pressure (kPa)	Heel	5882.8 (4492.0-6917.8)	6110.1 (4969.7-7670.3)	**.010*	0.03
	Medial Midfoot	600.7 (369.7-1031.3)	342.0 (193.5-543.9)	**.<.001*	-0.34
	Lateral Midfoot	1239.7 (846.0-1640.3)	1040.1 (488.5-1360.5)	**.<.001*	-0.22
	Central Met Head	2093.8 (1797.5-2350.9)	2573.7 (2336.9-3155.8)	**.<.001*	0.70
	Lateral Met Head	1973.2 (1789.8-2280.9)	2329.7 (1926.4-2653.2)	**.<.001*	0.39
	Medial Met Head	3043.5 (2546.4-3480.2)	3175.1 (2718.6-3837.6)	**.008*	0.21
	Hallux	2551.5 (1948.4-2858.0)	2156.8 (1601.2-2558.7)	**.001*	-0.25
Peak Pressure (kPa)	Heel	398.0 (350.9-472.7)	509.1 (438.5-623.4)	**.<.001*	0.50
	Medial Midfoot	244.1 (210.8-284.5)	237.2 (167.6-256.8)	*.105*	-0.08
	Lateral Midfoot	224.8 (183.9-275.8)	236.8 (143.8-288.1)	*.197*	0.01
	Central Met Head	345.7 (300.8-374.9)	543.0 (486.2-610.5)	**.<.001*	0.84
	Lateral Met Head	356.0 (314.1-396.2)	463.2 (410.2-508.6)	**.<.001*	0.77
	Medial Met Head	370.8 (325.2-452.5)	576.6 (472.3-624.7)	**.<.001*	0.56
	Hallux	562.2 (430.2-631.3)	480.4 (590.8-635.6)	*.177*	0.15
Contact Area (%)	Total	91.1 (82.6-94.1)	74.7 (71.2-80.3)	**.<.001*	-0.69
	Medial Midfoot	93.3 (77.4-98.6)	52.8 (42.1-72.3)	**.<.001*	-0.25
Peak Tibial Acceleration (g)		1.02 (0.86-1.16)	1.01 (0.81-1.15)	*.946*	-0.00

Where data is median value (25th and 75th percentile), Met = metatarsal, * indicates statistically significant differences alongside *p* value and Cliff's delta *δ*.

3.2. Relationship between objective and subjective measures

The relationships between PP, RP and PA to relevant anchors of the subjective questionnaire were explored with correlations (Table 3.22) and scatter-graphs (Figure 3.19). The difference between the two footwear conditions for subjective variables was compared to the difference between objective variables that relate to the specific footwear region. Significant linear relationships were not commonly evident, however maximum heel RP and difference in "cushioning in the heel" were significantly correlated (r = .341, p = .039). The direction of the relationship identified that as a decrease in pressure in condition S occurred compared to C, there was an increase in perceived cushioning in S compared to C, evident by the distribution of the data (Figure 3.19a). Despite the lack of significant linear relationships, examination of the distribution of the scatter-plots through the "Distribution scores" identified patterns. In the central metatarsal head as RP or PP under the central metatarsal head decreased, perceptions of comfort increased in 70% of participants. In the midfoot the "Distribution scores" were relatively consistent across all the variables considered, as midfoot pressures and contact areas increased the feeling under the foot arch was rated higher on the subjective scale.

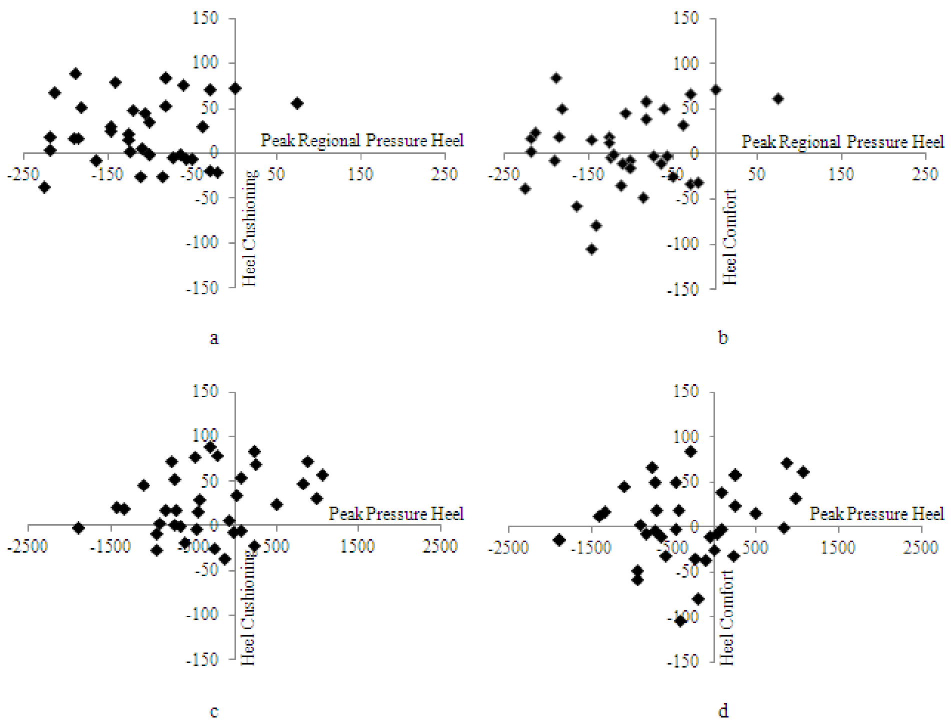


Figure 3.19 Scatter-graphs for difference scores between the two footwear conditions for subjective and objective measures.

Where units for subjective (Heel Comfort and Heel Cushioning) are comfort points (max 150) and objective (Peak Regional Pressure and Peak Pressure in the Heel) are kPa.

Table 3.22 Correlations between difference in scores for the two footwear conditions (Shoe S- Shoe C) for relevant objective measures and relevant subjective scores.

Region	**Variable**	**Questionnaire**	**Distribution Score**	**r value**	***p* value**
Overall	Total Contact Area	Overall Comfort	18	-.017	*p= .919*
	Contact Area Medial Mid-foot	Overall Comfort	19	.068	*p=.687*
Forefoot	Medial Met Head Peak Regional Pressure	Overall Comfort	15	.151	*p= .371*
	Medial Met Head Peak Regional Pressure	Comfort under the ball of the foot	14	.189	*p=.263*
	Central Met Head Peak Regional Pressure	Comfort under the ball of the foot	25	.043	*p=.802*
	Hallux Peak Regional Pressure	Comfort of toes	16	.111	*p=.512*
	Hallux Peak Pressure	Comfort of toes	23	.112	*p=.510*
Midfoot	Medial Midfoot Peak Pressure	Feeling under the foot arch	23	.134	*p=.429*
	Medial Midfoot Peak Regional Pressure	Feeling under the foot arch	21	-.147	*p=.385*
	Medial Midfoot Contact Area	Feeling under the foot arch	23	.079	*p=.644*
Heel	Heel Peak Pressure Regional	Cushioning under the heel	17	.341	**p=.039*
	Heel Peak Pressure	Cushioning under the heel	23	.081	*p=.635*
Peak Tibial Acceleration		Cushioning under the heel	22	-.201	*p = .233*

The distribution score is calculated for a total of N=37 and calculates the number of difference-pairs between subjective and objective measures, which followed the expected relationship. For example, as peak tibial acceleration increased perceptions of ‘cushioning under the heel’ decreased. Where Met = metatarsal, * indicates statistically significant differences alongside *p* values.

4. Discussion

Comfort is one of the most important aspects for a consumers' decision to purchase footwear (Jordan et al., 1997; Papuga and Burke, 2012) and therefore a valid protocol to quantify and compare comfort in walking footwear is essential for manufactures to develop comfortable footwear ranges. The variable nature of comfort makes quantifying the concept subjectively and objectively a sensible, yet challenging approach. This study modified a questionnaire to measure subjective comfort in walking footwear and then coupled this tool with objective measures of in-shoe pressure and PA to present a protocol to quantify aspects relating to comfort in walking footwear. The questionnaire developed as the first phase of the study was repeatable, for the five participants tested, and subsequently was utilised to compare two footwear conditions. Throughout the testing, no participants asked for clarification of any of the scales and semantics utilised and therefore no additional direction was required aside from that consistently printed on each questionnaire. This was noted as further evidence that the questionnaire was a successful modification of the running-shoe based questionnaire of Mündermann et al. (2002).

4.1. Footwear condition comparison

The variability in reported comfort was high in the present study, for example comfort around the toe joints was rated from 56-105 by different participants in condition S, consistent with ranges in previous research (approximately 10-110 in subjective reports from 9 participants, Mündermann et al., 2002). Additionally, consistent with other research, this range in response between participants is dependent on the scale being considered (Mündermann et al., 2002).

The outcome variables from the questionnaire were able to distinguish differences in the perception of the footbed between the two footwear conditions chosen for this comparison. Aspects relating to perceptions of the rearfoot and forefoot sensations were rated as significantly higher in condition S compared to C. "Cushioning under the heel" and "Comfort under the ball of the foot" differed significantly between the two conditions. Footwear features that have been linked to comfort in previous research can rationalise the response from the participants in the current study. These features include the construction of the footbed, anthropometric characteristics and upper characteristics (Hennig et al., 1996b; Jordan et al., 1997; Miller et al., 2000; Mills et al., 2011). The upper characteristics, size and pitch (from heel to toe) of the footwear styles were relatively consistent across the two styles (Table 3.19) and no questionnaire scales relating to these aspects produced significantly different responses from participants. Difference in hardness of the material under the ball of the foot and the heel (Table 3.19) in the two conditions may likely explain the significantly different reported comfort between the two shoes. Condition S included a thicker, softer footbed throughout the length of the shoe (Table 3.19), which was likely perceived, and reported, to be more comfortable than the footbed in shoe C. This is consistent with previous findings that participants can perceive difference in hardness of materials underfoot (Hennig et al., 1996a) and preference is generally towards softer materials (Mills et al., 2011; Mündermann et al., 2001).

In addition to the questionnaire, the plantar pressure data in regions of the foot was able to distinguish between footwear conditions with significant differences in pressure in all foot regions (Table 3.21). In the heel region, both the RP (+227.3 kPa, p=.010) and PP (+111.1 kPa, p=.000) pressures were significantly higher in condition C than S. This is consistent with the perception of improved heel cushioning in shoe S and may be due to the differing footbed constructions (Table 3.19) (Hennig et al., 1996a). Also in the Midfoot region, higher medial (+258.7 kPa, p =.000, δ = -0.30) and lateral (+199.6 kPa, p =.000, δ = -0.84) RP were evident in condition S than C. The traditional PP variable identified no significant difference, suggesting that more sensors were loaded, but the maximum pressure in a single area of the region did not increase significantly between conditions. This could be due to the Soft EVA (Ethylene Vinyl Acetate) construction in shoe S under the midfoot deforming slightly on weight-bearing and increasing mid-foot contact area (+40%, p=.000). The difference in contact area in the medial midfoot between the two conditions is similar to the 34% reported in custom-orthotics compared to a shoe (Redmond et al., 2009). Despite the contrasting contact area and pressure variables between the two conditions, the scale relating to “Comfort under the foot arch” did not differ significantly between the two (Table 3.20). This may be indicative of a preference for some participants finding increased arch support more comfortable, which may be related to foot anthropometry (Mündermann et al., 2001). It contrasts previous research where increased contact area is consistently recorded in footwear where subjective comfort scores are higher than comparators (Che et al., 1994; Jordan et al., 1997).

The plantar pressure in the two footwear conditions also differed significantly in the Metatarsal regions. The Metatarsal heads peak and regional peak pressures were lower in shoe S than C for all three Metatarsal regions ($p<.001$, $-1.00 \leq \delta \geq -0.89$). Previous research identifies reductions in pressure in the Metatarsal heads in footwear where comfort is reported higher (Che et al., 1994; Jordan et al., 1997). Consistent with this previous research, “Comfort under the ball of the foot” was significantly higher in shoe S condition, where the metatarsal pressures were lower. Despite this significant difference, however, the overlap between the two data-sets was large and the difference of 9.0 comfort points between the two conditions did not exceed the potential error of 10.1±3.3 comfort points highlighted in the repeatability work.

Tibial acceleration data has previously been utilised to relate to footwear or surface comfort and perceptions of impact severity (Lake and Lafortune, 1998; Whittle et al., 1994). PA measured during walking in the two footwear conditions did not differ in the current research. It would be expected that the thinner sole of shoe C would have increased PA, consistent with previous research in walking (Price et al., 2014) and running (DeWit et al., 2000.; Hamill et al., 2011; Nigg et al., 1987). However, the peak tibial accelerations were comparatively low (Price et al., 2014) in the present study, likely due to the treadmill and low self-selected velocity.

4.2 Relationship between objective and subjective measures

The current correlation analysis was undertaken on the difference between the scores of the conditions for the objective and subjective measures in an attempt to control intra-participant variability between scores. Previous authors have normalised subjective scores to mean scores (Clinghan et al., 2007) or to control values (Hennig et al., 1996; Lake and Lafortune, 1998). Data from a visual analogue scale would ideally not be normalised to mean scores as it is assuming information about the data e.g. that all participants would use the whole length of the scale. This is why the method of comparing differences between footwear conditions was chosen in the current approach as it in-part controls for plantar pressure and subjective rating variations between footwear styles. The inclusion of the “Distribution score” analysis was to compare the distribution of the data in regions of the scatter-plot, without the requirement of relationships to be linear.

In the correlation analysis, only one correlation undertaken identified a significant linear relationship between the subjective and objective aspects. The direction of the relationship identified that as pressure in the heel in condition S decreased compared to C, there was an increase in perceived cushioning in condition S. This meets expectations in that as pressures in the heel decrease, perceptions of heel cushioning increase, consistent with findings by Hennig et al (1996a). In contrast, tibial acceleration in the current research did not demonstrate a significant relationship with subjective measures of cushioning, contrasting previous research (Goonetilleke, 1999, Lake and Lafortune, 1998). This may be due to the limited difference in hardness between the two conditions tested or other studies including a far wider range of recorded acceleration values (Goonetilleke, 1999).

In the present research, no other correlation between subjective and objective measures demonstrated a significant relationship. Similarly, Clinghan et al. (2007) correlated plantar pressure data with ratings of perceived comfort in a range of retail price running shoes, identifying no significant correlation, despite differences in recorded plantar pressures. It may be that the footwear in the current research was too similar, eliciting minor differences in sensation, which could not be interpreted and thus conveyed effectively via the questionnaire. This potentially confounds this correlation-approach to analysis as participants could not differentiate aspects of the footwear meaning difference scores are limited in range.

The “Distribution scores” in the present research, however, identified a trend, particularly in the midfoot for an increase in pressure and contact area to be rated as more comfortable. This reflects findings in previous research where higher peak pressures in the medial midfoot have been identified in insoles rated the most comfortable (Chen et al., 1994) and that the provision of an arch support in a high heel increases midfoot pressures by 126%, coupled with comfort ratings of 28 mm (Yung-Hui and Wei-Hsien, 2005). Lange et al (2009) identified increased force-time integral in the heel region was apparent alongside higher comfort values. This pattern was not apparent in the heel region when Chen et al (1993) compared the force-time integral in the running footwear rated the most and least comfortable on an individual participant basis. The authors, however, identified a decreased pressure-time integral in the heel of the least comfortable footwear during walking. This is consistent with the findings of the “Distribution score” in the present research for the heel region. The regional peak pressure decreased as the comfort scores increased in only 17 of the 37 participants, therefore 20 of the participants rated heel regions with higher regional peak pressures as more comfortable. The importance and sensation of different footwear regions is likely to be a function of the gait style of the wearer, but it is apparent that responses also vary between participants and footwear styles.

4.3. Conclusions

Despite the objective measure of plantar pressure identifying differences between conditions. The tibial acceleration data and subjective questionnaire data did not establish many differences between regional comfort in the shoes. Thus the relationship between the two was generally weak. Further research is required to explore the relationship between subjective and objective measures of comfort with systematically modified shoes to compare outcomes with correlation analysis to pin-point the specific sources of differences in reported comfort or plantar pressure measures in order to specifically direct the design process such as footbed materials and last form. However this work has successfully demonstrated a holistic protocol for footwear companies to compare footwear comfort in two shoes from subjective and objective sources, which highlights that modifying the scale to be consumer- and footwear-specific poses benefits.

4.4 Limitations

The present study, consistent with other reported research quantifying comfort subjectively and objectively, applied the questionnaire twice only and did not repeat the comfort reporting for each shoe (Wegener et al., 2008; Zifchock and Davis, 2008). The protocol has been adapted to remove the recommended 6 repeated questionnaires required for an interpretation of long-term comfort in order to make testing more feasible for high participant number tests of fewer conditions. This protocol thus lends itself to footwear companies to conduct with manageable data output and feasible timescales as part of a design process. This potentially provides a more short-term reported comfort, more akin to a consumer trying on the shoe in a shoe shop (Kong and Bagdon, 2010; Mündermann et al., 2002). The aforementioned use of a treadmill to control walking speed during this testing is a limitation as the interaction of the foot with the floor is altered (Hardin et al., 2004). However this is the chosen methodology for the comfort assessment at the leading UK footwear testing company (SATRA). Foot type can influence plantar loading and future work should categorise more than just foot size to generalise results to specific populations (Chuckpaiwong et al., 2008). Whilst the conditions were not blinded to the individuals, efforts were made to ensure that the individuals were not aware of which condition they were wearing, but it is appreciated that some bias may have been introduced. Future studies should blind brand names from study participants if possible.

Conflict of Interest Statement

The primary author of the paper works on a project funded by FitFlop ltd, supervised by Dr Jones. Work was undertaken with scientific diligence, data was collected, analysed and the paper written with no influence from the funding company.

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Chapter 4 Critique

The critical appraisal of the current work begins with a critique of the research designs under subheadings to address aspects of the research designs. Specific methodological choices relevant to the protocols are then discussed. Following this, a summary of the findings from the four biomechanical concepts being explored in 'health and well-being' footwear are included:

- "Shock Absorption"
- "Instability"
- "Gait modifications"
- "Comfort"

The research is positioned in the existing literature base and the novelty and addition of the work discussed. The four objectives presented at the outset of the thesis are explored, with a discussion as to how the thesis met these aims:

- Provide data on the influence of FitFlop^{TM} footwear on walking and standing
- Demonstrate the importance of biomechanical testing in walking footwear
- Modify testing from running footwear protocols for walking footwear
- Measure footwear biomechanics concepts relating to 'health and well-being' footwear

Further to this the dissemination and wider impact of the work is considered and summarised.

4.1 Critical Appraisal of Research Designs

Specific aspects relating to the research designs of the Papers are addressed, relating to the research questions, populations, footwear conditions, familiarisation period and data collection/protocol and statistical approach.

4.1.1 Research Questions

The work presented herein was completed as part of a Knowledge Transfer Partnership (KTP007228) and then a commercial research contract with a company (FitFlop ltd). The nature of the research, being driven by a small organisation with specific motivations, has led to the research questions, in part, being somewhat descriptive. The testing of items of footwear as opposed to specific footwear features means that the ability to perform stringently controlled hypothesis testing is somewhat limited. For example the data relating to "Gait Modifications" was collected alongside other studies and collected as additional conditions in research projects which were to be completed to meet the requirements of the research contract. As a consequence, the research questions were more comparative than might have traditionally been seen in footwear research. Irrespective of this the studies provide useful information and comparisons between a FitFlop™ shoe, a flip-flop and barefoot walking. The seemingly independent, descriptive aims of some aspects of the KTP and research projects were altered, and other footwear conditions added to produce research questions that enabled the quantification of different aspects of biomechanics in 'health and well-being' footwear, which were publishable, using the FitFlop™ shoe as a tool.

4.1.2 Populations

The populations included in each of the studies are presented in Table 4.1. Numerous aspects of the participants such as age, gender and health status would have influenced the study outcomes in this thesis.

Table 4.1 Subject characteristics for research papers included in the submission for PhD.

Paper (*Biomechanics Concept*)	N Number	Gender	Mass (kg)	Height (m)	Age (years)	Shoe size	Health Status
1,2 (*Shock Absorption*)	13	2 M, 11F	62.0±10.3	1.65±0.05	27.5±8.8	UK 6	Healthy
3,4 (*Instability*)	15	15 F	62.6±6.9	167.1±4.2	29±6.7	UK 5&6	Healthy
5 (*Gait Modifications*)	20	20 F	64.7±6.4	1.63±0.05	31±9	UK 6	Healthy
6 (*Gait Modifications*)	40	20M, 20F	72.5±15.2	1.71±0.09	35.2±10.2	UK 7±2	Healthy
7 (*Comfort*)	40	40 F	63.2±9.0	1.63±0.06	40±14	UK 5,6&7	Healthy

4.1.2.1 Population Age

The youngest person who participated in this range of studies was 19 and the oldest 65 years of age. The emphasis when recruiting for the studies was to recruit representative participants in contrast to previous research which focuses on younger populations (Burgess and Swinton, 2012; Horsak and Baca, 2013). The study participants needed to represent the wearers of the footwear to provide a representative interpretation of function, as highlighted by Sterzing et al. (2012) in their review of the integration of research into the footwear industry. Thoroughly documented age-related changes to proprioception, reaction times, muscular strength, foot shape, kinaesthesis and sensitivity to touch infer that footwear function and preferences would interact with age (Holland et al., 2002; Kaplan et al., 1985; Mickle et al. 2009). Therefore, the focus of recruitment was post-graduate students, staff and mature students at the University. A combined mean age of over 27 years for the papers within this thesis better reflects the users of footwear in the 'health and well-being' footwear market.

Testing an age to reflect the appropriate footwear wearers was particularly relevant for the studies of instability footwear where younger, more athletic participants may be less responsive to any inherent instability in the footwear (Burgess and Swinton, 2012). Following an assessment of unstable footwear, numerous authors have reflected on their study participants potentially being too stable to be influenced by the intervention applied (Burgess and Swinton, 2012; Germano et al., 2012; Plom et al., 2014). Additionally, in barefoot or stable footwear conditions, younger people have an improved ability to maintain balance, attributed to age-related factors such as declines in visual function and reductions in muscle strength (Lin et al., 2008; Prieto et al., 1997). Consistently, Buchecker et al. (2013) reported different responses in knee loading to wearing unstable footwear in an elderly group (>55 years) compared with a young group (18-35 years). Thus suggesting that greater perturbations than footwear can offer might be required to remove stability in young healthy populations.

Additionally, within Paper 7, the participants wearing the footwear would be influential on subjective study variables. The demographics of the participant would influence subjective responses to the comfort questionnaire, such as their footwear habits and familiarity mediating their underlying preference of footwear in-terms of comfort (Au and Goonetilleke, 2012). Thus the relevance of a user to the footwear being compared was considered. Nevertheless, when applying the results to the wider population the specific demographics of the participants, such as being associated with higher education, should similarly be considered. This background of the participants would influence the language that they understand and the footwear they wear on a daily basis (due to income and work clothing for example) and therefore their preceding expectations. The age of the participants would also influence the sensitivity of their foot to areas of poor fit or discomfort induced by the footwear conditions tested. In the foot dorsal and plantar-surface vibration and tactile sensation decreases as a function of age due to changes in mechanoreceptor morphology and density, decreased nerve conduction and elasticity of the skin (Perry, 2006; Scott et al., 2007). This age-related decline would likely have an influence on responses to the comfort questionnaire for specific foot regions in the current study protocol. This decline in sensitivity has been identified to take effect at approximately 70-80 years of age (Perry, 2006; Scott et al., 2007), therefore with a later onset than the participants included in the present research.

4.1.2.2 Population Gender

The research focused on female participants as when the research began, FitFlop™ only produced footwear for women. Male participants were tested later (in time) in the series of studies (Papers 1,2&6). The footwear they were tested wearing differed for Paper 6 as the FitFlop™ does not have a unisex design which spans UK 4-11 sizing. The male example of FitFlop™ footwear has a thicker footbed compared to the female. This unfortunately limits which study conclusions can specifically be attributed to gender differences as opposed to differences in the footwear styles, hence the lack of assessment of gender interaction in that paper. This, however, induces limitations into the interpretations of results as grouping differing populations that may respond differently to the footwear conditions applied may reduce both the conclusions that can be reached and the external validity of study results. Future work should further explore the gender differences and interactions highlighted by Shroyer and Weimar (2010) when walking in flip-flop footwear.

4.1.2.3 Population Health Status

As evident in Table 4.1, all studies were undertaken on 'healthy' asymptomatic participants. Recruited subjects completed forms to demonstrate that they suffered from no lower limb injuries and limitations to their walking. An obvious progression of the research would be to consider symptomatic patients and how 'health and well-being' footwear could be used as a treatment or prevention tool. For example, this has been conducted, in MBT™ footwear in patients with lower back pain (MacRae et al., 2013; Nigg et al., 2009). The instability footwear papers (Papers 3&4) attempted to quantify the differing nature of the instability induced by footwear and this could be inferred to provide specific instability challenges to patients.

4.1.2.4 Population Shoe Size

The shoe size of participants tested was limited based on the availability of footwear items. The unstable footwear ranges (five different shoes) were purchased in size U.K. 5 and 6 as these represent a mean size for females (Goonetilleke, 2012; Wunderlich and Cavanagh, 2001) and the master shoe last size for FitFlop. Papers 1&2 were limited by the customised footwear only being manufactured in size U.K. 6 for the testing. These were prototype footbeds that required new EVA moulds to be opened at the factory and bespoke hand-stitched uppers at significant cost. It would not have been realistic to expect multiple samples to be produced. Similarly, Paper 5 utilised a bespoke pressure insole that was only produced in a size U.K. 6. A greater range of foot sizes was tested in the gait kinematics study with footwear conditions being available in size U.K. 4-11. As shoes are scaled to foot size there is no obvious reason why testing footwear on different foot sizes would be a limitation generally. However, there are two examples of exceptions to this assumption in this research:

- The resolution of the pressure system (Medilogic) alters with increases in shoe sizes as sensor sizes change and consequently peak pressures may change
- People at extreme sizes of footwear may have different comfort expectations than those with the most common sizes as they may be used to wearing ill-fitting shoes.

The limited sample shoes for testing in Papers 6&7 has controlled for these two interactions by only testing small or limited shoe size ranges. This, however, does limit the application of results to the wider population as wearers of footwear that is at the extreme size ranges cannot be assumed to be represented within this research.

4.1.2.5 Population Sample Size

Sample sizes in this research were somewhat limited considering the number of variables that have been compared. The effect size of variables was added to Papers 6&7 to demonstrate the limitations of the sample size on the data dependent on the publication journal. Generally, sample sizes have been limited by the time-constraints placed on the research due to the commercial nature of the project. The sample sizes (Table 4.1) were comparable to participant numbers in recent and similar research (Arezes et al., 2013; Zhang et al., 2012). Furthermore, the number of individual comparisons (for example 34, in Paper 5) was lower than in other similar research. For example Hennig and Milani (1995) compared 20 plantar pressure and accelerometer variables for 19 running shoe conditions with only 22 participants. Recently, Healy et al. (2012) compared over 40 plantar pressure and kinematic variables with data from only 10 participants. Future research should attempt to address this imbalance in number of variables versus participant number, in the footwear biomechanics field.

4.1.2.6 Other Considerations

Participant characteristics were recorded for each experiment including mass, height and age. These were detailed to demonstrate the population represented by the participants with the aim of this resembling the relevant groups for the results to be generalised to. For example the body mass index of the participants in Paper 6 closely matched that of the British population (NHS, 2010). In Paper 7, the foot dimensions of the participants were recorded (instep girth, metatarsal-phalangeal joint girth and foot length). This was included to enable a measure of how much the participant foot size differed from the last size of the shoes as this measure has previously been discussed in relation to perceptions of footwear comfort (Cheng and Perng, 1999; Miller et al., 2000; Mündermann et al., 2001). It was determined in the final stage of the research that heel breadth was an important measure for comfort, however, due to the similar heel width of the two comparison footwear conditions this measure was omitted (Paper 7, Table 3.19, pg. 155). In retrospect, the measurement of foot dimensions would have been of benefit at each stage of the research, particularly in the studies including in-shoe pressure measurement in the protocol. Other characteristics of the subjects that would have impacted on the results such as their foot shape would have been beneficial to include in the participant information sections of the publications to enable the results to be generalised to specific groups. Foot type has been related to both balance (Hertel et al., 2002) and plantar pressures (Chuckpaiwong et al., 2008). Other similar research has characterised participants pre-testing, for example, Hennig and Milani (1995) determined their participants' foot structure was "normal" via clinical examination and inspection of footprints. This provided additional information to the reader to infer generalizability of the research to a population or sub-groups thereof.

4.1.3 Footwear Conditions

The commercial nature of the work impacted on some aspect of study design, in particular the concluding comfort concept of work and the footwear conditions that were compared. The nature of the research enforced a study design which replicated a product comparison study for product development and marketing purposes. Consequently, the two shoes differed quite distinctively on numerous aspects including footbed construction, upper material stiffness, shoe volume and fit. The ideal situation would have been to systematically modify specific footwear design features and performance criteria as the independent variables in studies. Then see how this influenced dependent variables such as plantar pressure and/or subjective comfort scores, such as undertaken by Mills et al. (2011) who modified only the hardness of orthoses in running. The comparison was still valuable in that it enabled the comfort methodology to be presented and discussed in relation to the previous papers and research included in wider literature. Additionally, it enabled the demonstration of a potential bench-marking tool for footwear companies to test products in the developmental stage, or to compare and benchmark to market-leaders. It did not, however, enable a true validation of the modified comfort scales regional reporting of comfort. A systematically modified independent variable (e.g. altered footbed hardness) would enable the quantification of influences on dependent variables (e.g. plantar pressures and reported comfort) to be compared and confidently attributed to changes in the independent variable. Additionally, the interaction of thickness and hardness of the footwear midsole for Paper 2 would have been of benefit as in a true footwear manufacturing environment these two features would not have to be manipulated independently. The constraints on the number of test shoes which could be provided for the testing limited the investigation of interactions. This constraint is a reality of commercial research projects, where aspects of the research design, such as resources and time, are restricted by a product cycle which ultimately has an economic driven emphasis.

The input and insight of FitFlop ltd. encouraged a more function driven approach to comparator shoes. From a scientific perspective a comparator is often chosen as a standard trainer (e.g. Landry et al., 2010; Sacco et al., 2012) or barefoot (Germano, 2011; Light et al., 1980). These items of footwear are often not a comparable shoe from a 'real life' perspective in that they are not what would be worn by a similar wearer in a similar context. For example, the FitFlop™ would be worn in place of another sandal (as compared in Papers 1&2), or a flip-flop (as compared in Papers 5&6). The studies undertaken comparing FitFlop™ to a flip-flop utilised barefoot as an additional control condition to give a baseline measure more consistent with, and comparable to, existing research. A barefoot condition was not chosen for the instability work as this has previously been demonstrated to be significantly less stable than shod conditions (Robbins et al., 1994) and is also not a comparable 'real life' wear condition. Thus, the control sandal utilised in the papers relating to instability was chosen due to it having a similar price and wearer as the unstable sandals, however, it made no marketing claims regarding instability. However, the chosen shoe did have a negative heel, which likely altered muscle activation and therefore impacted on the interpretation of results (Paper 3, pg. 101; Paper 4, pg. 116).

4.1.3.1 Footwear Properties

The footwear properties of the footwear conditions included in the studies were generally reported in the methodologies (e.g. Paper 1 pg. 63, Paper 2 pg. 77, Paper 3 pg. 93). More extensive details could have been quantified and reported for all papers such as footwear mass, sole material properties, footwear longitudinal stiffness, heel heights/sole thickness and sole contact areas. These characteristics of the footwear would have impacted on the variables being quantified within the studies. Footwear mass has been considered in literature pertaining to energy expenditure in footwear (Theusen and Lindahl, 2009) and alterations to swing kinematics (Shroyer et al., 2010). Altering footwear material properties (eg. Paper 2) and upper styles (e.g. Paper 1) will impact on the mass of the footwear being tested and thus potentially these associated variables. The ranges of footwear mass in the test conditions within the research studies within this thesis were 140-534 g, which are of a Havaiana™ flip-flop and MBT™ sandal respectively. From empirical literature this suggests minimal impact on energy expenditure, substantially lower than the 1% increase in energy expenditure per 100 g of footwear mass reported in running (Frederick, 1984). The mass of the FitFlop™ compared to the flip-flop may have increased the tibialis anterior activation during swing in Paper 6 (pg. 145); moreover a similar response may be expected in the MBT™ condition in Paper 4 had the muscle activity in swing been quantified, which was beyond the scope of the current comparison (pg. 114).

The sole contact area in contact with the floor would have been particularly relevant at different phases in different studies. For Papers 1&2, at initial contact the shape and material of the heel section of the footwear would have been influential in determining the characteristics of the impact (assuming a heel contact at initial contact). The shape of the heel is a factor in determining the contact position and the volume of material between the calcaneus and the floor. Additionally, in Papers 3&4, the sole shape and total contact area with the floor are influential in determining the stability of the wearer due to larger weight-bearing areas increasing stability (Menant et al., 2009). This variable is a function of sole shape and sole properties, for example the stiff sole of MBT^{TM} (Figure 2.3c, pg. 25) will not deform under weight-bearing and the contact area would vary based on the position on the sole on which the wearer is loading. However, the more compressible nature of the Reebok Easytone^{TM} balance pods (Figure 2.3b, pg. 25) would increase the contact area as body weight is applied during stance. The design of the Skechers^{TM} and FitFlop^{TM} footwear (Figure 2.3d, pg. 25) includes outsoles which are flat and not deformable and therefore in single-leg standing these conditions represent the largest contact areas of the outsoles compared in the unstable shoes in this research. Future research pertaining to quantifying biomechanical variables related to stability should measure the actual and functional contact area of the footwear conditions tested.

The sole material properties are also a functionally significant feature of the footwear due to the influence on the shock absorption properties (as discussed in Papers 1&2). Additionally, the material properties of the sole influence the stability of the wearer in terms of providing firm support surface and also, alongside the dimensions, influence the proprioception of the wearer (Robbins et al., 1994; Robbins and McClaran, 1995). These factors would influence study outcomes relating both to the biomechanical concepts of instability (Papers 3&4) and the gait modifications in the footwear (Papers 5&6). Furthermore, the sole material properties contribute to determining the longitudinal stiffness of the footwear alongside other characteristics such as the upper material and outsole design. Manipulating the longitudinal stiffness of footwear can alter the biomechanics of gait such as foot kinematics at toe-off and comfort (Branthwaite et al., 2013; Roy and Stefanyshyn, 2006) and thus would be a worthwhile factor to have been quantified in the research papers included within this thesis.

4.1.3.2 Footwear age

All footwear conditions tested in this research were new. An extension of the work would be to consider the influence of degrading shoes with wear. This would be relevant research for investigations relating to shock absorption properties of footwear, in-shoe plantar pressures and perceptions of comfort. It has been reported that wearers often continue to wear flip-flops past the point where they are worn out (Shroyer et al., 2010), thus testing an aged shoe may demonstrate significant changes in plantar pressures in this footwear style. It may be that as the shoe is worn and the EVA becomes degraded the reported negative effects of the footwear style arise in wearers. Compression of the footwear over long periods is likely to degrade materials and reduce the shock absorption properties of the heel section of the footwear (Saito et al., 2007). Testing the footwear using the mechanical and human protocols discussed in the papers relating to shock absorption pre- and post- long-term wear would be informative. Additionally, investigating the influence of long-term wear on plantar pressure and comfort variables may ne informative for footwear technologists to improve longevity of products and components (House et al., 2002). Fully defining standards for the longevity of products and components efficacy would be informative for clinicians and technologists alike. It would also enable a footwear company to define a life-cycle for their products effective functionality. Responses will differ dependent on the material construction of the footwear and how well this withstands repeated loading over time. Pratt et al. (1986) reported a reduced shock absorption capability of a 6 mm Plastazote insole that was worn for only 72 hours by a participant. Kong et al. (2009) identified significant alterations to running kinematics, particularly at the ankle, after 200 miles of road running in a pair of shoes, suggesting that gait may be altered due to prolonged wear in footwear. Dixon et al. (2003) considered the influence of degrading on the shock absorption properties of military boots. The protocol consisted of phases of mechanical (using an Instrom device) and human degrading replicating wear of 0.25-250 km. All test insoles increased stiffness and demonstrated a decreased ability to absorb shock following the degradation. The mechanical protocol loaded 500 kPa in the heel region 50 ms after initial contact and was demonstrated to degrade the insoles further than the human protocol. Therefore, if further work is to be conducted, the Instrom testing may not replicate wear of the footwear and thus appropriate testing protocols and standards should be a consideration. As well as shock absorption properties, the plantar pressures in mechanically degraded insoles has been considered (House et al., 2002). The study identified no significant change in plantar pressures when subjects wore the insoles in their own boots following 130 km of simulated running compared to new insoles. This was despite bench-top methods identifying changes in both stiffness and peak deceleration. The study, however, compared plantar pressures using regional peak analysis of the rearfoot and forefoot peak pressures where the same sensors may not have been compared from each test condition. Summing sensors to produce a total measure of pressure may have been more sensitive to change after long-term wear or direct comparison of single-sensor values from the in-shoe pressure device and should be considered for future longevity comparisons.

4.1.4 Data Collection and Protocols

The protocols within the research papers controlled some aspects of the data collection in order to aide interpretation or treatment of data and inference of study findings. One aspect specifically influencing all the four biomechanical concepts was walking velocity. The walking velocity utilised in the studies determined both the internal and external validity of the data and thus specific methodologies were selected based on study outcomes (Table 4.2). All walking velocities in the present research were self-selected, contrasting other similar literature, which prescribed a walking velocity to participants (Nigg et al., 2010). The methodology utilised in the current research had the advantage of enabling the data to be collected on a realistic wearer walking at a velocity which resembled a ‘real life’ gait in the footwear style tested. This enabled the external validity of the research results to be high. This was particularly relevant when comparing specific features such as in Paper 4 where the flip-flop may have reduced gait velocity compared to other footwear styles (Finnis and Walton, 2008; Shroyer and Weimar, 2010). Papers 1&2 utilised a self-selected velocity, which was then fixed. This was implemented in an attempt to isolate specific footwear difference, such as a thicker footbed on impact characteristics and thus walking velocity needed to be consistent. In contrast, the investigation in to instability utilised a self-selected velocity, which was not controlled between conditions (Paper 4) such that the quantification of style/brand differences on temporal-spatial characteristics of gait could be quantified and external validity was high for each individual shoe.

Table 4.2 Walking velocity approach for papers

Publication	Walking velocity
Paper 1	Self-selected then fixed (timing gates)
Paper 2	Self-selected then fixed (timing gates)
Paper 4	Self-selected for each shoe condition
Paper 5	Self-selected then fixed (timing gates)
Paper 6	Self-selected for each shoe condition
Paper 7	Self-selected then fixed (treadmill)

Note: Paper 3 was a single-leg standing protocol

4.1.5 Statistical Approach

The study designs for the research undertaken in this body of work were repeated measures of subjects with the application of different conditions. Hence, the statistical approach implemented was repeated-measures ANOVA or Friedman and Wilcoxon signed-rank tests dependent on the parametric nature of the data. In Paper 6 (pg.141-142), linear transformation (square-root) of non-parametric variables was undertaken, such that they were treated as parametric and analysed consistently with other variables in the paper. This was advantageous in that it enabled a more simple interpretation of the paper and simultaneously increased the power of the statistical analysis (Field, 2005). Correction for multiple comparisons was included when more than two conditions were compared and described in the methodologies, using either Bonferroni correction or Holm-adjustment. The Holm adjustment was advantageous in that it is stepwise and less conservative than Bonferroni and thus decreased the likelihood of type II errors (Aickin and Gensler, 1996; Knudson, 2009).

The statistical approach assumed a group level effect; whereas it might be that an effect is created by the shoe, but on a subset of participants. Thus, the study designs may have benefited from the consideration of individual differences as comparisons of mean data may mask these individual responses. Research relating to orthotics and footwear identifies subject-dependent responses (De Wit et al., 1995; Nigg, 1986). These responses are likely determined by pre-existing features of the subjects such as lower limb alignment (Davis et al., 2008) and foot sensitivity (Mündermann et al., 2001), which then interact with the treatments or conditions applied. Considerations for these individual subject differences may have allowed more thorough conclusions to be drawn from the work. Nigg et al. (2012) highlight the importance of considering individual differences due to the large variation in individual responses to unstable footwear. A later study quantifying muscle activation in MBT™ shoes justified this suggestion by presenting a vast range of muscle activation responses from 15 participants (Branthwaite et al., 2013). Further, comparisons of the unstable footwear effects on subjects may be aided by grouping participants based on their response, reflecting clinical methods to determine who may benefit from specific footwear treatments; for example, ‘responders’ versus ‘non-responders’ (Jones et al., 2014). This would separate wearers who would be influenced by unstable shoes and those that will not and could increase the sensitivity of research studies at differentiating between styles and footwear features by removing the assumption of a group-level effect.

4.1.6 Familiarisation Period

An aspect of methodologies that was considered was the familiarisation period allowed by subjects prior to trials in the laboratory test sessions. The familiarisation period in the current body of work ranged from no specific footwear familiarisation (Paper 7), to two walks of 15 metres (Paper 5) to five minutes (Paper 6). The range in the present studies was consistent with familiarisation periods in other literature (Table 4.3) (Dinato et al., n.d.; Hamill et al., 2011; Hennig et al., 1996; Zhang et al., 2013).

Table 4.3 Papers and familiarisation periods included in the protocols.

Publication	Familiarisation period	Data collected
Paper 1	"a familiarisation period"	Kinematics, kinetics and tibial acceleration
Paper 2	"a familiarisation period"	Kinematics, kinetics and tibial acceleration
Paper 3	"short familiarisation"	Kinematics, kinetics and electromyography for single-leg standing
Paper 4	Practice walking trials to familiarise	Kinematics, kinetics and electromyography for walking
Paper 5	2 walks of 15 m	In-shoe plantar pressure
Paper 6	Up to 5 minutes	Kinematics, kinetics and electromyography for walking
Paper 7	Practice walk on treadmill No in-shoe familiarisation	In-shoe pressure, tibial acceleration and questionnaire for treadmill walking

As in the included publications, the duration of a familiarisation period is not always reported or defined in literature and some studies do not include a time period where the subject wears the shoe prior to data collection (Buchecker et al., 2012; Horsak and Baca, 2013a). The familiarisation period will influence the outcome of test results, particularly in footwear which may be new to the wearer such as flip-flops or unstable footwear. Additionally, recording the subjects' previous experience with the footwear style would have enhanced the generalisation of the results to the population. For example, some participants may have been common wearers of flip-flop footwear or have had previous experience wearing unstable footwear. Other study participants may have never worn a flip-flop or an unstable shoe prior to volunteering. Some research papers, particularly those relating to unstable footwear, report that participants have not previously utilised the footwear prior to embarking on the research study (Branthwaite et al., 2013; Buchecker et al., 2012; Turbanski et al., 2011). Despite not being reported in the publication, this was the case for the participants in Papers 3&4. It has been demonstrated that a familiarisation period significantly alters gait kinematics when walking in unstable footwear; in particular it reduces the variability between steps in kinematic and electromyography data after a long-term intervention wearing the shoe (Stöggl et al., 2010). These findings are relevant for Paper 6 relating to reducing gait modifications, as highlighted in the discussion (pg. 146). Other authors have provided test footwear up to two weeks prior to testing for subjects to become familiar walking in the footwear (Buchecker et al., 2012; Horsak and Baca, 2013).

In addition to the familiarity with the specific style of footwear, the participants' familiarity with the testing protocol would influence the findings from the research. For example, if a subject had never walked on a treadmill before or worn in-shoe pressure insoles then their variability at the start of the protocol may have been greater than other participants and their own variability later in a protocol. This justifies why a familiarisation period was included specifically for walking on the treadmill in the protocol included in Paper 7, similar to other authors (Hennig et al., 1996).

4.2 Critical Appraisal of Specific Methodological Choices

Each assessment of biomechanical concepts pertaining to 'health and well-being' footwear within the thesis and body of research included specific methodological choices, which will be discussed in turn.

4.2.1 "Shock Absorption"

Quantification of the shock absorption properties of walking footwear is important for comfort and the minimisation of discomfort from any potential symptoms (Voloshin and

Wosk, 1982; Whittle et al., 1994). This research, therefore, included the manipulation of a commonly reported mechanical drop-test method to better replicate the impact with the ground in the footwear conditions being compared. This was a novel approach to footwear comparisons and the publication promoted the consideration of the specificity of testing protocols to both the wearer and activity. Although the methodology employed has some weaknesses (Paper 1, pg. 71), it was demonstrated to be repeatable and valid and acts to initiate the development of specific testing methodologies for 'health and well-being' and walking footwear in the future, consistent with objectives two and three defined out the outset of this thesis.

4.2.1.1 Footwear Hardness Measurements

The quantification of the hardness of the footwear was undertaken with a bespoke Shore A device utilised in the factory producing the test footwear and with a Shore A durometer both at the University and the factory. The hardness range of the footwear was produced specifically for the study by the manufacturer for the company which funded the research. The number of EVA "beads" was modified to alter the hardness of the heel section of the footwear specimens produced for the testing. The main measurement device for the manufacturers is a Shore A durometer at the end of the manufacturing process, for quality control, which is their standard tool. Also implemented was a bespoke device with an increased surface area "plate" which is expected to be less affected by the positioning of the device on the non-uniform EVA surface. The samples were also tested for hardness using a Shore A durometer at the university on receipt of the delivery and after the testing sessions.

The use of an Instrom device to characterise footbed stiffness would be a good supplement to the methodology sections of papers to characterise the stiffness of footwear included in studies. The currently reported measure, within these papers and the wider research field, is a Shore value. This quantifies indentation hardness, a function of the yield strength of the surface of the material; a characteristic unrelated to shock absorption properties (Shorten and Mientjes, 2011). The relevance of this measure to the function of footwear in a dynamic situation is questionable. Shore hardness, however, is commonly reported and has relevance to industry and footwear companies who purchase materials defined by Shore values and also enables comparison across studies with the absence of a more practical and accessible measure.

4.2.1.2 Bespoke Drop-Device Assumptions

4.2.1.2.1 Initial Contact

The bespoke drop-device was constructed in order to replicate the impact conditions at initial contact in walking. The device therefore needs to replicate as many as the initial conditions as possible (pg. 20) as validly as possible. The original ASTM protocol was defined based on the landing kinematics of male rearfoot runners running at 3.6 m.s^{-1} captured by Cavanagh et al (1984). Cavanagh et al (1984) utilised high-speed cinematography and tested 10 male runners to define initial contact with the ground. The velocity at contact was 0.90 m.s^{-1} in the direction of running for the horizontal component. The vertical component of velocity was a mean value of -0.70 m.s^{-1}. Thus the resultant velocity of the heel region of the foot at initial contact was reportedly 1.12±0.41 m.s^{-1} at an angle of 40° to the horizontal (Cavanagh et al., 1984). The resultant vertical ground reaction force was also aligned with the direction of the shank at initial contact. These conditions are not replicated in walking (Figure 4.1).

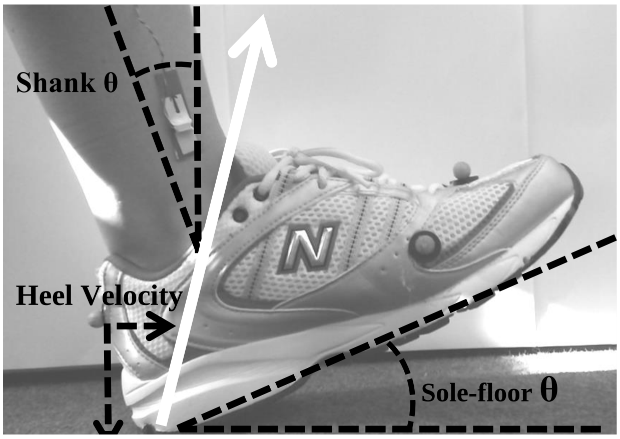


Figure 4.1 Example characeristiccs of initial contact in walking.

Data taken as an example from a participants walking from Paper 6: Shank θ =17°, Sole-floor θ =23°, Force θ =7°, Heel velocity horizontal 0.12 m.s^{-1}, Heel velocity vertical -0.41 m.s^{-1}.

As evident in Figure 4.1, the resultant force vector is not aligned with the shank, with an offset of approximately 24° in this example. Also, the horizontal component of the heel velocity comprises a substantial part of the resultant heel velocity at initial contact during walking. However, magnitudes of the variables presented in this example were variable between footwear conditions and participants. This is consistent with other literature, where reported values range from 0.2-0.4m.s^{-1} in similar participants (Lockhart et al., 2003; Winter, 1995). It is also consistent with the large range in vertical heel velocities originally reported by Cavanagh et al (1984) of -0.16 to -1.20 m.s^{-1}. The data indicates that the drop-test protocol over-simplifies the impact with the floor in walking and that the inclusion of further information would increase the validity of this protocol for quantifying the shock absorption properties of walking footwear. In particular, applying the force in a representative direction and calculating resultant heel velocities to be utilised within the drop-test protocol would further increase the validity of the protocol implemented in Papers 1&2 and the specificity to replicating the biomechanics of walking.

4.2.1.2.2 Individual Participant Calculation

A further limitation of this method involves the calculation of the drop-device data with individual participant values. Despite not being reported in the methodology (Paper 1, pg. 65), the peak acceleration and peak force values utilised to calculate the effective mass of the device (Figure 3.1, pg. 66) were from the individual subject which most closely matched the mean data. This decision was made to reflect the nature of the personalisation of the process to individual subjects or participant groups. This however means that the comparison of data from an individual is related back to a mean value from the alternative protocols. Utilising the mean data from the participants would have altered the effective mass values for the Triple-density sandal (to 13.7 kg) and Shoe (to 17.5 kg) and thus would have altered the results from the drop-test for these two conditions. This would have implications for the comparison of the three methodologies (ASTM, adapted and human protocols) within this paper and future research should consider mean data from relevant sample populations.

4.2.1.2.3 Effective Mass Calculation

The calculation of the effective mass can be undertaken using a peak force method or an impulse-velocity method (Kessler et al., 2003). The first method had been previously used in footwear research and employs Equation 2 (Nigg, 2010) and the variables denoted in Figure 4.2. This method was selected for the protocol development for Papers 1&2 to develop a protocol to replicate walking, consistent with the third objective of this research.

Equation 1 Effective mass peak acceleration method.

$$m_{eff} = \frac{F}{a}$$

Where *F* and *a* are the peak vertical force and the peak tibial acceleration respectively (Figure 4.2).

Calculating the effective mass of the lower limb in walking was influenced by the peak tibial acceleration and vertical ground reaction force transient not occurring simultaneously as they would in running or drop test methodologies (Figure 4.2). Equation 3 provides an example to describe the methodology utilised in the Papers with data from an example participant presented in Figure 4.3.

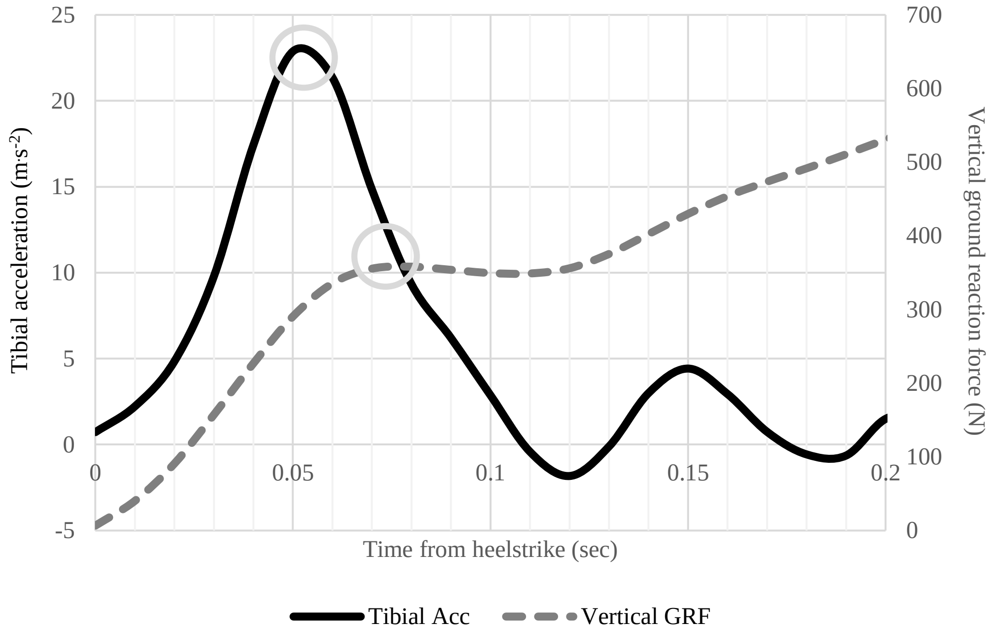


Figure 4.2 Effective mass calculation example

Where peak tibial acceleration and the transient peak of the vertical ground reaction force are approximated with circles.

Equation 2 Effective mass peak acceleration method example

$$m_{eff} = \frac{358.2\ \text{N}}{22.9\ \text{m.s}^{-2}} = 15.7\ \text{kg}$$

Where *F* and *a* are the peak vertical force and the peak tibial acceleration respectively (Figure 4.2).

For Paper 1, the use of Equation 2 resulted in an effective mass of 10.6-17.3 kg to replicate the energy in walking in the different styles of footwear (Paper 1, Figure 3.1, pg. 66). The values for effective mass and drop-height established for the Health Sandal/FitFlop^{TM} in Paper 1 were implemented in Paper 2 to present a methodology to estimate the influence of thickness and hardness of the footbed through drop-testing, which could be incorporated into the research and development processes for waking footwear companies.

The second method to estimate effective mass is an impulse-momentum method and has also been used in relevant research considering drop-test devices (Kessler et al., 2003) (Equation 4).

Equation 3 Effective mass impulse-momentum method

$$m_{eff} = \frac{\int_{t1}^{t2} F\,dt}{\int_{t1}^{t2} a\,dt}$$

Where F and a are the force and the tibial acceleration respectively and t_1 and t_2 are the times at which the force value crosses zero.

This impulse-momentum method does not have as simple definition within walking studies as available in drop-test protocols. The application to footwear assessment during walking requires the definition of the boundaries of the integral (t_1 and t_2; Equation 4), which is problematic as the ground reaction force value does not cross zero after impact due to the reaction to the body mass being applied to the force plate. This impulse-momentum method has been implemented by (Chi and Schmitt, 2005) to estimate the energy absorbed by the heel pad during impact in walking and running. The authors utilised the velocity and change in time evident from the initial contact to the end of the impact phase. The methodology defined the impact phase as the peak of the heel-strike transient, thus defining the boundaries (t_1 and t_2) (Equation 4 and Figure 4.3).

The impulse-momentum method for this specific trial estimated an effective mass of 15.7 kg using the method from Paper 1 and 11.5 kg using the definition proposed by Chi and Schmitt (2005). The differences in calculated effective mass from the two methodologies resulted in a difference in energy of 0.28 J when the vertical heel velocity established in Paper 1 for drop-tests on a flip-flop was implemented. Further work to replicate walking impacts utilising drop-tests should establish the most accurate and valid method for estimating effective mass in walking data, nevertheless these methodologies will continue to neglect to quantify the complex interaction of the human and environment (Sterzing et al., 2012).

The effective mass of the limb has been considered in reference to walking by some authors in walking protocols, pendulum protocols and isolated heel pad samples (Aerts and De

Clercq, 1993; Chi and Schmitt, 2005; Jørgensen and Bojsen-Møller, 1989). Jørgensen and Bojsen-Møller (1989) dropped a mass of 1.6 kg on heel pad specimens to represent the effective mass and lower shank at touchdown in fast walking. The impact velocity chosen was 1.4$m{\cdot}s^{-1}$, reported by the authors to replicate heel velocity in fast walking, however this is more representative of heel velocities toward the floor recorded in running, exceeding the maximum value recorded in 10 male subjects running at 3.6 $m{\cdot}s^{-1}$ (-1.20 $m{\cdot}s^{-1}$) (Cavanagh et al., 1984). Despite the substantially lower mass employed in their study, Jørgensen and Bojsen-Møller (1989), applied greater energies at impact (1.57 J) than the highest energy drop-test in the current work (Paper 1; 1.05 J). Similarly, Whittle et al. (1994) implemented a lower mass (2.27 kg) dropped from a greater height (178 mm) to characterise the shock absorption properties of surfaces. Aerts and De Clercq (1993) utilised a pendulum methodology with a mass more similar to the current work (11.6 kg) striking the heel of participants at velocities ranging from 0.37-1.37 $m{\cdot}s^{-1}$, also consistent with the current research (Paper 1, Figure 3.1, pg. 66). As undertaken in Paper 1, altering the effective mass applied for each footwear condition was advantageous in that it incorporated different joint kinematics or limb postures which may be evident at heel-strike in different footwear styles or in different populations. This was particularly relevant for footwear with open uppers, as identified in the kinematic testing within the papers (Paper 1, Figure 3.2, pg. 68) and could be integrated and modified to add valuable information into the footwear product cycle of walking footwear.

4.2.1.3 Bespoke Drop-Device Repeatability

The methodology adapted for the research was the ASTM F1614 Procedure A, which mimics the effective mass and heel velocity in running to apply an impact energy of approximately 5 Joules to the footwear heel (Figures 2.2 & 4.3). The modified protocol presented in Paper 1 (Figure 3.1, pg. 66) adapted the drop height (and consequently impact velocity) and mass to better replicate that which was evident in walking. Subject specific velocities (and therefore drop-heights) were calculated for the compared footwear styles (e.g. flip-flop = 7 mm). This method over-simplifies the initial contact in walking due to the nature of the double-limb support and the incline of the foot not being flat as the heel contacts the floor. Other inherent assumptions and limitations include to the aforementioned initial conditions which was previously discussed and reviewed.

Two footwear conditions of varying composition, to represent the two extremes of the footwear tested, were compared from two repeat tests with the drop-testing protocol to consider the repeatability of the device and protocol (Table 4.4). The repeat testing was undertaken six hours after the original test to enable any compression of the footwear midsoles to return to the original condition prior to secondary testing. The same laboratory was utilised for the testing and the footwear was stored at a stable and moderate temperature.

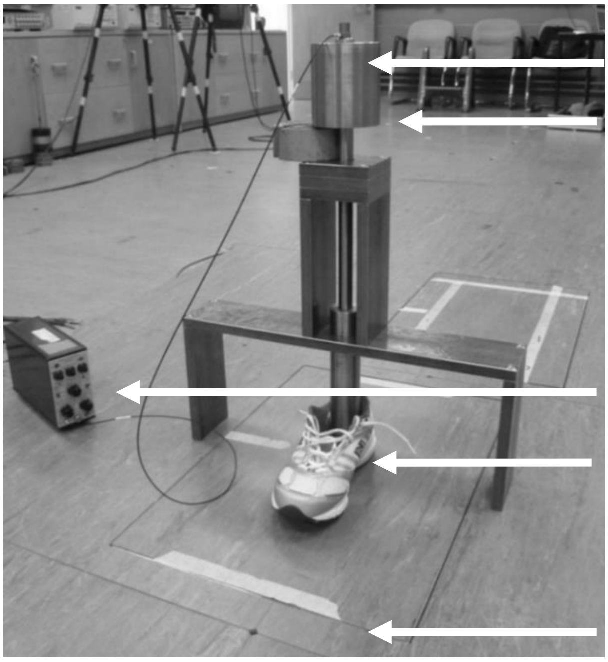


Figure 4.3 Drop-testing device for testing footwear shock absorption capabilities.

Table 4.4 Footwear conditions tested for drop-test repeatability

Condition	Midsole Construction	Heel Sole Depth (mm)	Heel Hardness (Shore A)	Mass (g)	Image
6B	Ethylene Vinyl Acetate (EVA)	310	38	244	
KG	Mixed synthetic construction	263	58	315	

Where KG and 6B are two footwear conditions from research projects not presented in this thesis, but representative of extreme range of shoes in terms of construction and hardness.

Table 4.5 Comparison of results from two repeat sessions of the drop-testing.

Footwear	KG			6B		
Test Session	1	2	*p* value	1	2	*p* value
Peak Acceleration ($m.s^{-2}$)	58.2±4.1	54.8±3.6	.208	19.8±2.2	21.1±1.6	.353
Peak Force (N)	743.3±61.8	785.2±56.3	.286	223.7±28.4	226.3±19.4	.877

Comparison of mean ± standard deviation results from trials 26-30 for two repeat sessions of the drop-device testing. T-test comparison of results from time one and time two are presented ($p < .05$).

Comparison of the values from the two test sessions identified no significant difference between each shoe for the two tests ($p \geq .208$; Table 4.5). The minimal detectable change was calculated to indicate a magnitude of difference that is greater than the error expected from the protocol and measurement device:

Equation 4 Minimal detectable change (MDC)

$$MDC_{95}=1.96\times 2\sqrt{}\times(\ s\sqrt{}(1-r))$$

Where s = mean standard deviation of time one and time two, r = the reliability coefficient of the test i.e. Pearson's correlation co-efficient between times one and two.

The calculation of minimal detectable change values from these shoes determined maximum values of 3.2 $m{\cdot}s^{-2}$ for acceleration and 44 N for force. For there to be a detectable difference between the conditions the difference must have exceeded these values. The author is not aware of any other repeatability studies for bench-top testing devices or methods reported in the literature, which present a measure to define detectable change. This may be due to most devices being mechanically driven and thus the variation being substantially lower between trials (e.g. SD ≈10% for the ASTM protocol; Schwanitz et al., 2010). Jørgensen and Bojsen-Møller (1989) reported a reproducibility of 96% when using a drop-test setup to test EVA foam specimens at five minute intervals. However, it was not clear how this was calculated and therefore how the values compare.

4.2.1.4 Human Testing

Limitation to the in vivo data collected during walking in the current research is the lack of correction for the angular motion of the limb in the peak tibial accelerations. This approach is consistent with other studies (Lafortune and Hennig, 1992), however even in walking the angular motion of the limb may still account for 16% of the accelerometer signal that is recorded (Lafortune and Hennig, 1989). This contribution cannot be considered consistent between the footwear styles due to altered kinematics so should be accounted for in future work. Other limitations to the accelerometer protocol are that skin mounted transducers over-represent bone accelerations (Lafortune and Hennig, 1989) and the placement of the accelerometer distally effects the variability of the signal. These limitations are inherent to this methodology and consistent with all other research using similar protocols. However, the accelerometer was not removed between-conditions, thus limiting the influence of any potential error from these sources on the within-subject comparisons in this research design.

In addition to time domain analysis, future research could incorporate frequency domain analysis of the accelerometer and force signals, which enables the researcher to isolate frequency components of ground reaction force and accelerometer data (Shorten et al., 2003). This permits the separation of portions resulting from the active and passive impact phases of the ground reaction force and the resonant frequency of the accelerometer (Nigg et al., 1981; Shorten and Winslow, 1992). This is therefore a more specific analysis of the features of the accelerometer signal relevant to the shock absorption properties of the footwear. Frequency aspects of the vertical ground reaction force and accelerometer signals in walking have also been associated with soft-tissue injury, with the proposal that it is the frequency content of resulting transients that determines the extent of cartilage injury (Gillespie and Dickey, 2003). Research using frequency domain techniques include shock transmission in different running shoes (Light et al., 1980; Shorten and Winslow, 1992) and the effectiveness of insole materials in walking (Gillespie and Dickey, 2003). It has been demonstrated that frequency domain analysis is more sensitive to differences in shock absorption properties of insoles and footwear than time domain variables (Healy et al., 2012). Additionally, frequency domain variables have been demonstrated as more linearly associated with subjective perceptions of impact than time domain variables in running (Hennig et al., 1996; Milani et al., 1997). Frequency domain comparisons require large numbers of trials to compare conditions and as a result this method is not always feasible in a gait laboratory setting, particularly with clinical patients walking where trials may be minimal. This analysis technique is therefore often reported in studies that use a treadmill in the protocol (e.g. Hamill et al., 1995). The analysis process also requires that the frequency content is stable over time, which can not necessarily be assumed for the duration of walking stance or shorter subsections of this (Gillespie and Dickey, 2003; Shorten et al., 2003). The time domain approach was chosen due to the limited number of foot ground contacts in the current protocol. Additionally, at the outset the protocol was obtaining variables to replicate with a mechanical test device, thus comparing the frequency content of accelerometer data from a device mounted on a human limb to a metal device is unlikely to produce worthwhile information due to the presence of soft tissue in-vivo. Despite the aforementioned methodological constraints and assumptions, the comparison of frequency domain variables to differentiate between shock absorption properties of the conditions in future research in addition to those in Paper 2 may add value.

4.2.2 "Instability"

The papers included in the second biomechanical concept of the research compare instability in commercially available unstable sandals. Instability is quantified in terms of ankle kinematics, centre of pressure trajectory and lower limb muscle activation. The analysis of frontal plane range of ankle motion in unstable footwear during standing was novel and had not been undertaken previously as a measure of instability, with other authors considering sagittal plane alterations at the lower limb joints in MBT^{TM} (New and Pearce, 2007) and centre of pressure and electromyography data with no kinematic information (Nigg et al., 2006b). Increased ankle inversion angle has since been demonstrated in unstable footwear (Debbi et al., 2012). In addition to quantifying range of motion, the measurement of kinematics at the ankle in the current research enabled the subjects' body position to be compared in the footwear styles during single-leg standing. Other authors have speculated that wearers of anterior-posterior rocker shoes, such as MBT^{TM}, sit back in the footwear in standing tasks, altering their ankle angle to find a stable base (Stewart et al., 2007). Comparing the mean ankle angle in the sagittal plane enabled this potential mechanism to gain stability to be recorded if present in the participants.

4.2.2.1 Tasks

The tasks undertaken to investigate the biomechanical concept of 'instability' may limit the generalisation of results to 'real life' situations when wearing 'health and well-being' footwear. Literature comparing muscle activation in footwear has focused on standing and walking as used in Papers 3&4 (Germano et al., 2012; Hömme and Hennig, 2011; Landry et al., 2010; Nigg et al., 2010; Plom et al., 2014). These tasks are common in daily life and act as a relevant starting point for comparative studies. Single-leg standing may be more demanding, particularly for younger, more stable populations and therefore may be more effective at inducing instability in different footwear and differentiating between conditions (Hömme and Hennig, 2011). Burgess and Swinton (2012) undertook a more varied protocol that included treadmill, stair and cone walking. This study included turning in the protocol to make the walking more similar to a daily walking situations. These tasks would be a progression to standard protocols once a holistic comparative study has been undertaken.

The single-leg standing protocol in Paper 3 was undertaken with the subjects freely positioning their foot on the floor on the force-plate without being constrained. Other research has constrained the foot position for participants in similar tasks by marking the force plate position such that all participants positioned their foot consistently (Plom et al., 2014; Turbanski et al., 2011). This approach is advantageous in that the variability in calculation of medial and lateral variables is removed. However, this methodology reduces any individual variability between participants in chosen standing position and may mean participants are in a position which does not represent their 'normal' or comfortable stance (McIlroy and Maki, 1997). Other methods include calculating a composite measure of the medial-lateral and anterior-posterior centre of pressure trajectory such that the position of the foot relative to the force plate does not influence variables (Prieto et al., 1996). In the current research the variables pertaining to centre of pressure direction were computed based on the position of foot and consequently the positioning of the foot on the force plate does not directly invalidate the medial- and lateral- calculations. It does, however, influence how these medial- and lateral- displacements relate to the body translations of the body centre of mass position. For example, if the foot is positioned 10° from progression in one condition and 15° from progression in another, subtle changes to the centre of mass position forward or backward will have different influences in these trajectories. A randomly chosen participant's foot placement for the single-leg standing task in the footwear conditions is presented in Figure 4.4. For this participant the foot positioning between conditions ranged from -1.7- 4.7°

from the anterior-posterior direction of the force plate (a range of 6.4°). The minimum angle was recorded in the Reebok EasyTone^{TM} condition, the maximum in the Skechers Shape-Up^{TM}. This is a relatively consistent magnitude when considering the other influencing factors and error in motion capture protocols (McGinley et al., 2009). However, this does mean that the centre of pressure variables compared included this angular deviation from the centre of mass projection directly forwards. Further research or analysis of this data or similar protocols or research should consider overcoming this limitation to the interpretation of results.

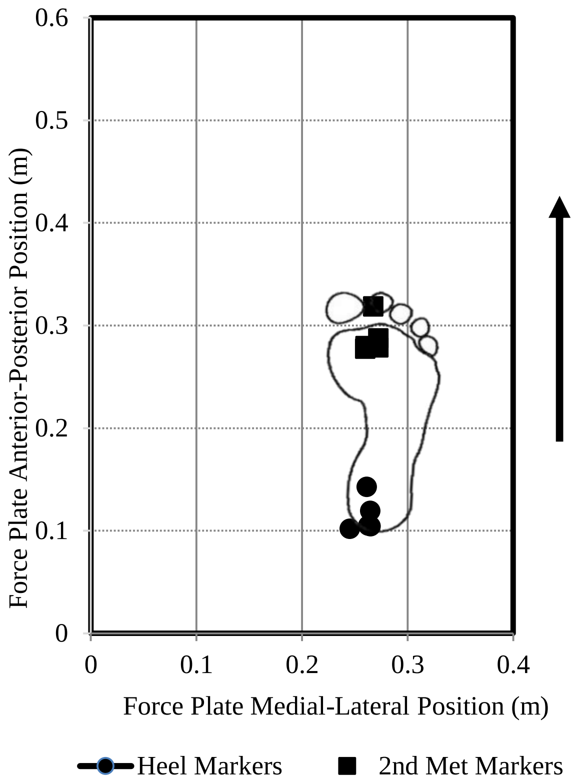


Figure 4.4 Foot placement on force plate for single-leg balance trials of randomly chosen participant

Where: heel markers from the five conditions are identified with a circular marker and markers from the second metatarsal with a square marker. For clarity, the foot overlay demonstrates an example position of the foot on the force plate and the arrow the direction in which the participant was facing.

4.2.3 "Gait Modifications"

The quantification of 3D-motion and plantar pressure when walking in flip-flops in adults is a progression of this research field, which at the time of the study (2010) included limited data.

The main methodological choices and novelties associated with Paper 5 were the design and use of a bespoke plantar pressure insole and the quantification of gripping under the hallux using the resulting data.

4.2.3.1 Bespoke Insole

For plantar pressure quantification in flip-flop style footwear a novel method of capturing data was required. A bespoke insole was designed and constructed with the assistance of Medilogic (T&T Medilogic, Gmbh, Germany) to measure plantar pressure in flip-flops without any data loss associated with cutting the insole to accommodate a toe-post (Paper 5, Figure 3.10, pg. 125). The insole was designed to cover a size U.K. 6 FitFlop Walkstar^{TM} and Havaiana^{TM} flip-flop footbed. The insole contained 150 surface resistive sensors (Figure 4.5) and was adapted with a small cut-away hole and slit from the medial edge to fit the toe-post. The insole had a larger area (≈190 cm^2) than the standard insoles (≈179 cm^2) to cover a greater plantar area than open footwear due to the larger surface area due to a wider last and lack of rand for upper attachment.

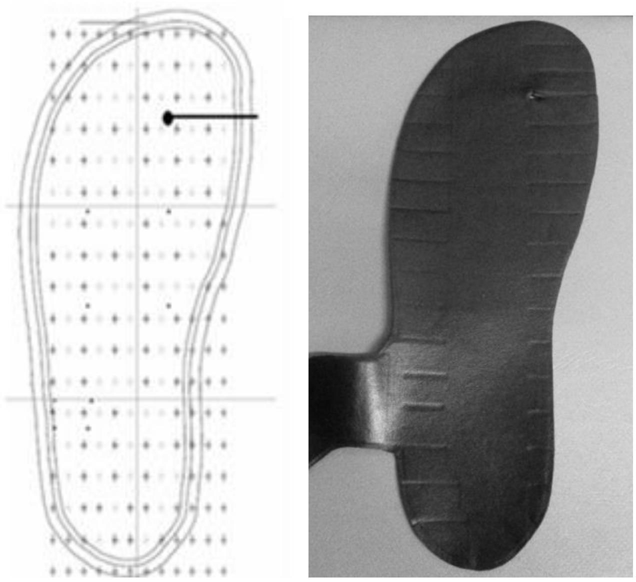


Figure 4.5 Bespoke insole schematic and photograph

Crosses identify single sensors within the sheet from which the insole was produced and the black line and circle highlight the region that was cut and re-wired to accommodate toe-post footwear.

The repeatability and validity of data from the bespoke insole was established prior to testing with a two-session repeat test of five participants walking in the insole over a force plate. The number of steps required for a stable mean was firstly investigated in the bespoke insole to determine a minimum step number required for the protocol. The subjects undertook repeated steps in the footwear and sequential estimation was undertaken to determine the step number where the cumulative mean fell between the total trial mean ± 0.25 of the total trial standard deviation (Hamill and McNiven, 1990). All centre of pressure variables were stable within nine steps for all subjects. Assessment of the peak pressure variables identified that all except the hallux pressures were stable within eight steps for all subjects. Mean and peak pressure under the hallux, in stance and swing, took up to 11 steps in two of the subjects to stabilise. These estimates of required steps for reliable pressure data are similar to eight steps for healthy participants walking on a treadmill (Kernozek et al., 1996) and 12 steps previously reported in neuropathic diabetic patients in covered footwear (Arts and Bus, 2011). It was ensured that all participants' data collection for the included research far exceeded the 11 steps estimated as a minimum requirement.

The repeatability of the data was considered utilising correlation coefficients and intra-class correlation coefficients (ICC) between two sessions walking in the footwear. Correlation coefficients ranged from r = 0.63-0.99. The lower correlations were in the average centre of pressure anterior-posterior velocity (r = 0.67) and the pressure time integral in the heel (r = 0.63). All other reported variables exceeded r = 0.82, with contact area in the midfoot being particularly consistent (r = 0.99). An example of the centre of pressure trajectories recorded from the two sessions is presented in Figure 4.6. ICCs between-session for the centre of pressure velocities in the medial-lateral direction were all ≥ 0.849, values exceeding 0.6 have been proposed as 'useful' (Chinn, 1990). The repeatability of these variables is relatively consistent with values reported for similar in-shoe pressure systems (Martínez-Nova et al., 2007; Murphy et al., 2005). All regional mean and peak pressures demonstrated ICCs greater than 0.7 (range ICC = 0.739-0.992). More specifically, the mean pressure in the hallux produced an ICC of 0.721 for the five subjects tested in the repeatability study. This exceeds the poor reliability (ICC = .14) previously reported by Boyd et al. (1997). Slightly lower repeatability was expected in the current protocol due to the toe-post footwear being tested. Compared to footwear with a restricting upper, placement of the insole can be slightly more erroneous and it was expected that insole and foot position within-the shoe may be more variable during walking and thus data less repeatable between steps and sessions than in a covered shoe.

The validity of the in-shoe pressure insole was established by comparison to force plate data. Correlations between force plate and insole centre of pressure trajectories and velocities produced ‘good’ to ‘strong’ correlations. Medial-lateral variables demonstrated weaker relationships (range: r = 0.67 and mean velocity: r = 0.50) than anterior-posterior variables (range: r = 0.67 and mean velocity: r = -0.84). The correlation between the ground reaction force and total force calculated from the insoles was high. However, a standard error of the estimate demonstrated a mean value of 39 N, demonstrating a substantial underestimation by the insoles. A conversion factor of *1.6 produced comparable peak loading values from both sources, consistent with findings using other resistive in-shoe measurement systems (Hennig *et al.*, 1996).

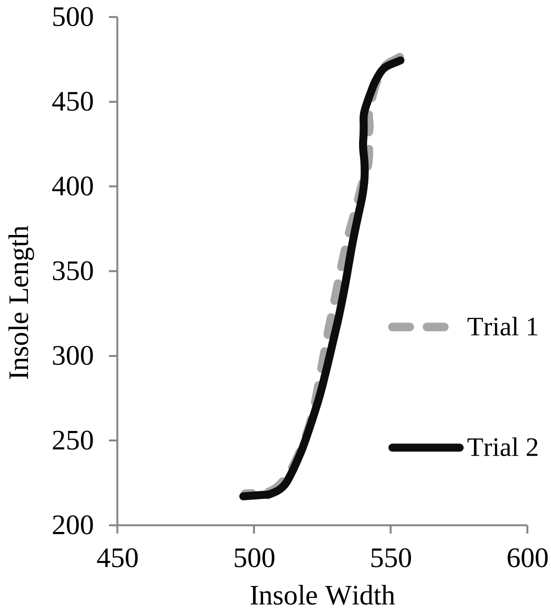


Figure 4.6 Example difference between centre of pressure trajectory in two trials.

The sensitivity of the Medilogic in-shoe pressure system was a limiting factor in the conclusions that can be extrapolated from Paper 5. Most notably, the advertised minimum range of the sensors in the system is 0.6 $N{\cdot}cm^{-2}$ (6 kPa). The data range reported for the plantar pressure under the hallux in swing in Paper 5 ranged from 0 to approximately 30 kPa. A large number of data points contributing to the mean value under the hallux presented in FitFlop^{TM} (0.05±0.50 kPa) and Havaiana^{TM} (0.36±0.62 kPa) were recorded below 6 kPa. The repeatability of this data however was established prior to testing. The use of single-sensor systems positioned under the hallux may provide a more sensitive measure to compare this variable in future as the sensitivity range of these devices is lower than the insoles utilised for testing. Another limitation to the research application was the data analysis and masking the sensors into regions for comparison. The use of a multi-sensor pressure insole system, however, may pose resolution advantages over the use of a discrete pressure measurement system where single-sensors are placed under specific anatomical regions of interest. Lange et al. (2009) conducted a similar study with a seven single sensor pressure system. The sensors were positioned under specific regions to relate to specific perceptions of comfort. These discrete sensors, however, may act as a foreign body in the shoe, potentially explaining the low correlations between reported comfort in the heel via questionnaire and the peak pressure measurements (lateral heel r = 0.057, medial heel r = -0.162) (Cavanagh et al., 1992).

4.2.4 "Comfort"

The assessment of footwear comfort as it is an essential feature of 'health and well-being' footwear. The combination of subjective and objective methodologies within one protocol was a consistent approach with other footwear biomechanics literature. The adjustment of the scale to be wearer- and footwear style- specific was a novel approach in comparison to previous literature.

4.2.4.1 Task

The comfort protocol testing was constrained by time, laboratory space and a large total number of subjects being tested (N = 65, with four to eight footwear conditions each, not all included in Paper 7). Furthermore, the protocol required a fixed walking velocity due to the collection and comparison of plantar pressure and tibial acceleration data. Hence, the decision was made to collect this data on a treadmill, consistent with other similar research (Hennig et al., 1996; Healy et al., 2012). The lower vertical heel velocity in treadmill walking (Lake and Robinson, 2005) may explain the minimal difference identified between the two conditions in peak tibial acceleration values and the thickness of the two footbeds was similar (Paper 7, Table 3.21, pg. 159). The thickness of viscoelastic material in the heel section was therefore similar and thus differences may not be expected due to consistent shock absorption properties (Paper 2; Whittle, 1999). Kinematic differences between treadmill and over-ground walking such as small changes in knee range of motion (Alton et al., 1998; Riley et al., 2007) and smaller ground reaction force maxima (Alton et al., 1998; White et al., 1998) mean that the findings from this study may require further consideration to be generalised to walking in a 'real life' situation. However, differences between walking on a treadmill and over-ground are reportedly small in healthy female participants familiar with treadmill walking (Alton et al., 1998), as these study participants were. Additionally, as the comparison was between two footwear styles with similar uppers and masses, it was assumed that the influence of the treadmill walking was relatively systematic for each footwear condition and therefore influenced each footwear condition plantar pressure, tibial acceleration and comfort result similarly. Treadmill walking is consistently used as an approach to collect comfort data on subjects, likely due to the ease of controlling walking velocity and duration of time spent in each shoe condition (Table 2.3). The included comfort protocol would have also benefited from the inclusion of dorsal pressure measurement as previously discussed (pg. 44), this was limited by a lack of a dorsal pressure measurement system at the University, however further work should incorporate this.

4.2.4.2 Plantar Pressure Data

The repeatability of the in-shoe pressure device was established for Paper 5 for the bespoke insoles; however, assuring the repeatability and validity of the standard insoles supplied with the device was essential for this publication. The insoles were loaded in the Emed TruBlu calibration device at eight known loads from a range of 30-400 kPa. An additional reading was taken with no applied load with the insole inserted into the device to ensure firstly, that the insole sensors recorded zero and secondly, that the device did not exert any unintended pressure pre-application of load. This device uses a bladder and air cylinder to load the insole to a chosen pressure between 50-400 kPa evenly over the insole surface (Figure 4.7).

Testing was undertaken in two sessions to assess both the validity and the repeatability of the insole in measuring applied pressure. From the first session, the maximum and minimum pressure values recorded from every sensor in the foot are presented (Figure 4.8) as well as the mean and standard deviation of the pressures applied (Figure 4.9).

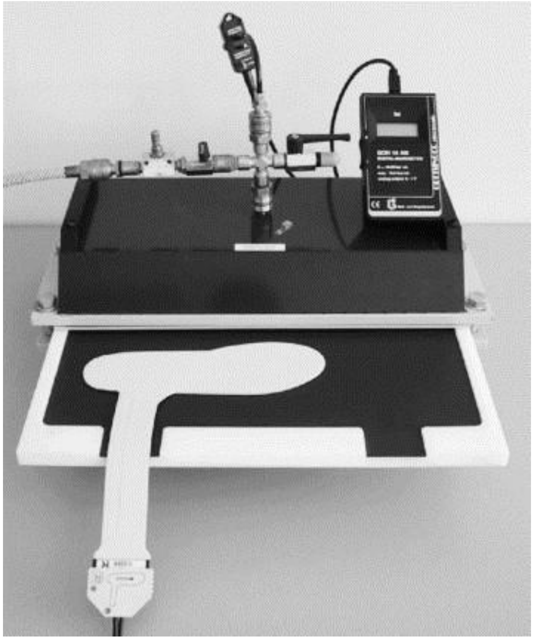


Figure 4.7 Novel TruBlue calibration device.
http://novel.de/old/productinfo/systems-pedar.htm

At the regions with lower plantar pressures such as the medial midfoot, the lateral midfoot and even the lateral metatarsal head, the error was lower. However, in regions with higher pressures it is evident that the maximum pressure values for single sensors are substantially inflated (Figure 4.8). Thus, the maximum and minimum pressure values varied compared to the actual pressure applied by up to a maximum value of 56%, which was recorded at 400kPa of application in a sensor in the medial heel border of the insole. However, when the sensors were grouped, as in this regional analysis approach, this reduced the maximum error across the entire insole to a maximum of 11.2% (mean 8.0%). The errors in measurements with the Medilogic insole were larger than those previously recorded in the Pedar system as well as the F-Scan system (Hsiao et al., 2002), but consistent with a recent comparison using different insoles with the same Medilogic system (Price et al., 2014).

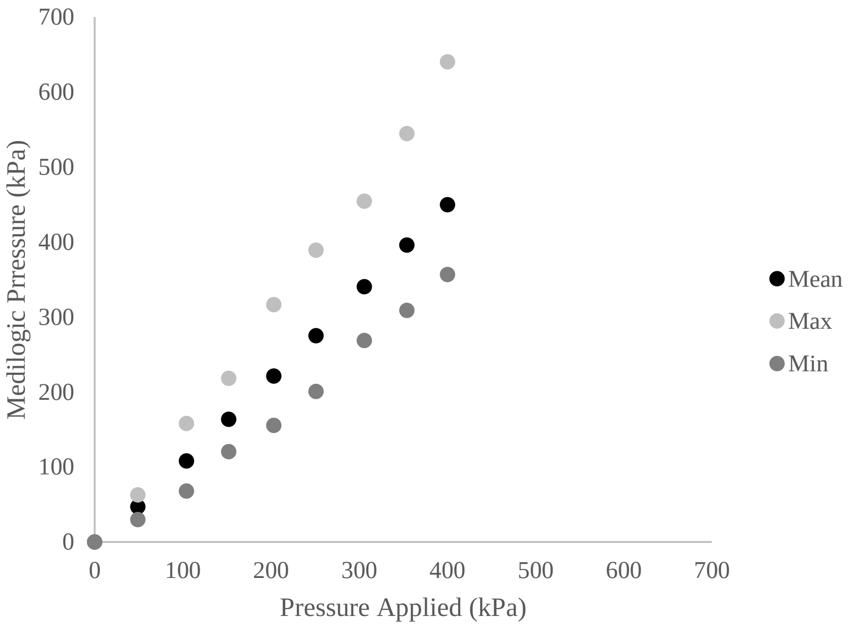


Figure 4.8 Range of pressure values from all sensors summed recorded in the first session

After a load of 49 kPa, all loads were overestimated by the insole. The application of 49 kPa produced a mean underestimation of applied pressure of 4.3%.

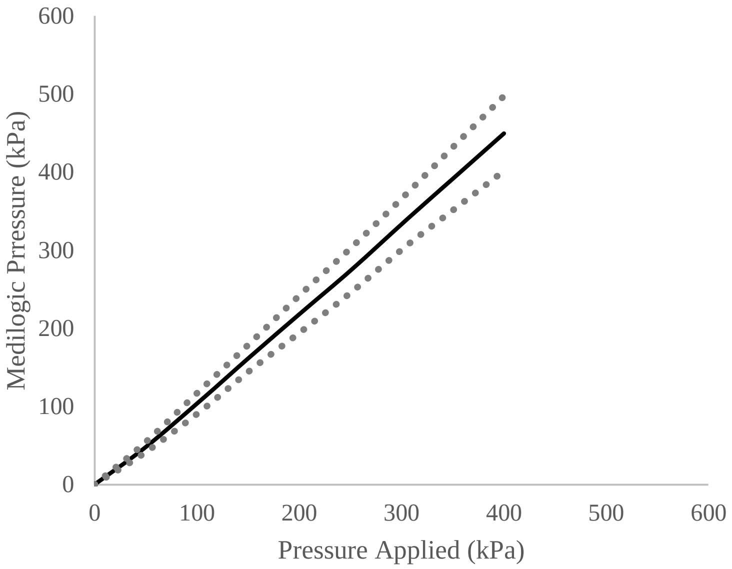


Figure 4.9 Mean and standard deviation of pressure values from all sensors summed recorded in the first session

The same pressure application protocol was undertaken three days after the original testing. The data from both test sessions is presented in Figure 4.10.

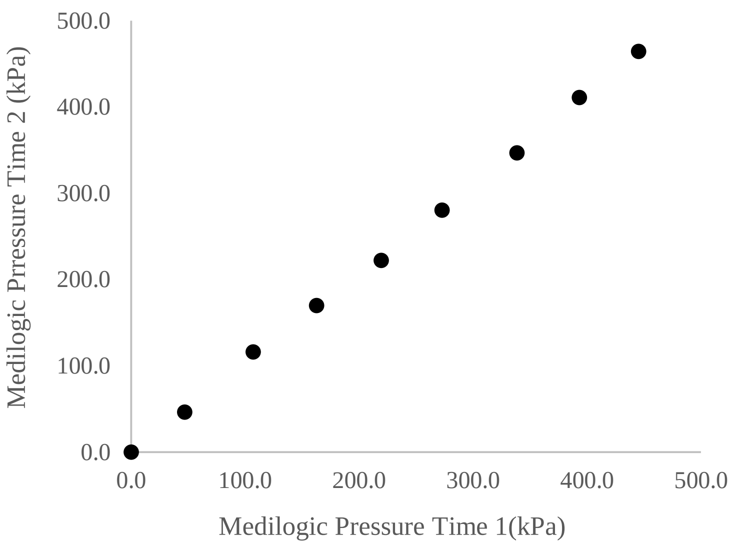


Figure 4.10 A comparison of the mean pressure values summed from all sensors recorded in the two session with the calibration device.

The average error at the load application of 251 kPa was an underestimation of 8.1% at time 2 from time 1. Fifty five of the 116 sensors within the insole demonstrated a difference of less than 5% between the two test sessions. Consistent with Price et al. (2014), the Medilogic insoles demonstrated high repeatability.

4.2.4.3 Comfort Scale

To conduct the subjective measure of comfort within this study, a comfort scale was developed by adapting a pre-existing well published and validated scale (Mündermann et al., 2002). The modified scale altered some terminology that was not deemed relevant to the footwear and participants being tested in this range of studies, reducing the focus of the VAS from running footwear and making it more general. (Paper 7, Figure 3.16, p 165). The repeatability of the newly developed scale was established by five subjects (mean±1 standard deviation: age 24.8±3.1 years, mass 62.8±3.0 kg and height 1.66±0.08 m) completing the questionnaire in three shoes (one twice each visit) twice in one day (4 hours apart) and again the next day. Repeatability was established with statistical analysis of the within-session, between-session and between-day scores for each subject and each aspect of the comfort scale.

Within-session comparisons were undertaken for the two tests of the control shoe by five subjects on three visits such that 15 difference scores were compared for each scale. Minimal changes in scores were recorded for the repeated footwear condition and some scales showed no mean difference within-session. The most repeatable aspects were the visual analogue scales relating to the "comfort around the heel" and "overall length comfort", which differed by a mean value of five (range 0-10) within-session from a maximum of 150 comfort points.

The least repeatable aspects of the comfort scale were "comfort under the foot arch" and "comfort of the toes". "Comfort of the toes" demonstrated a maximum difference of 56 and a mean difference of nine comfort points. "Comfort under the arch" differed by a mean value of eight and a maximum of 45. These large variations were generally explained by two of the subjects who demonstrated mean differences of greater than 10 points for the questionnaire visits. Mündermann et al. (2002) also reported varied responses and repeatability between participants and attributed this to participants having low foot sensitivity or being unfamiliar with, and thus unreliable completing, a VAS. The current within-session comfort scores should be viewed with caution as a baseline control shoe was not implemented. These scores would therefore be expected to be higher (the data more repeatable) if a consistent control condition was used prior to each test shoe (Mündermann et al., 2002). Between-session repeatability was determined by comparing the difference between sessions one and two for each shoe that followed the control condition (N = 2) and each participant (N = 5) for each scale (N = 10). Between-day repeatability mirrored this approach, however used sessions one and three. Comparisons considered mean differences, correlation coefficients and intra-class correlation coefficients.

The mean difference in scores between session one and two was 13 points, with a range of 0-72. Again, "length comfort" was the most consistent reported aspect with only one subject having a mean change in this score greater than ±7% (11 points). Mündermann et al (2002) also found this measure to be particularly stable, however their protocol utilised the same footwear with orthotic interventions such that the subjects were exposed to the same characteristic numerous times as shoe length was fixed throughout the protocol, it varied in the current research. Other scales were less stable, for example "comfort in the toes" demonstrated a mean difference of 17 points, with only one participant reporting comfort within 10% of the previous test for both shoes. The maximum change in "overall comfort" ratings between sessions was 38 points between session numbers one and three from the 150 point scale. This is substantially lower than evident in Mündermann et al (2002) between session numbers one and three for subject one for example who differed by ≈54 points in their reported scores for "overall comfort". Additionally, it is lower than some subjects demonstrate in their study for "overall comfort" in the recommended mean of session's four to six. The mean difference between sessions was consistent across the two footwear conditions tested, despite the comfort of one shoe being reported as substantially higher than the other by all participants in all sessions. This suggests that with the modified questionnaire the repeatability of the questionnaire was not affected by the comfort of the footwear at baseline and errors between sessions are systematic across shoes of different comfort levels. This contrasts Mündermann et al. (2002) findings where the comfort reported for the two extreme insoles (soft and hard) was more variable than the control insole. It may be that applying more conditions to the subjects would produce more variable responses, particularly at the extreme sensations and potentially due to respondent fatigue with a large number of conditions.

Correlations of participants subjective comfort scores were all significant ($p < .001$), with average r values of .780±.046, .690±.131and .793±.138 for sessions one v two, one v three and two v three respectively. An example of one subjects' correlations coefficients are presented in Table 4.6, other participants' data is plotted in Figure 4.11.

Table 4.6 Example correlation values for subject one for between-session and between-day questionnaire data.

	Subject 1 Session 1	Subject 1 Session 2	Subject 1 Session 3
Subject 1 Session 1		$r = 0.727$ $p < .001$	$r = 0.675$ $p < .001$
Subject 1 Session 2			$r = 0.887$ $p < .001$
Subject 1 Session 3			

Where: r = Pearson's product moment correlation coefficient and p = significance where significance was determined by $p < .05$.

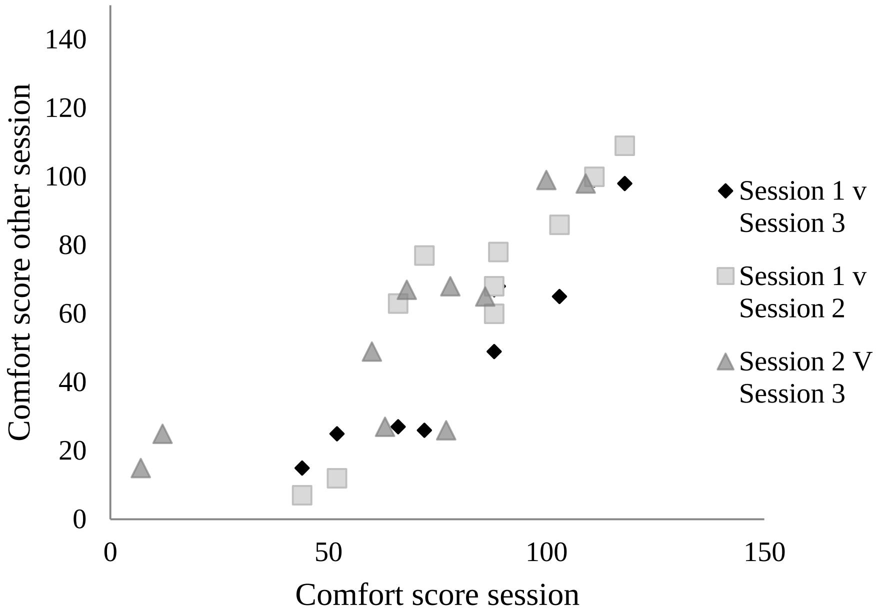


Figure 4.11 Example subject scatter plot of subjective comfort scores for each visual analogue scale on the comfort questionnaire.

All subjects demonstrating significant correlations between all sessions in the present study contrasts Mündermann et al. (2002) within session results, where correlations ranged from *r* = 0.108-0.952 and three of the nine subjects did not reach significance. Consistent with the present results, comfort score variability was highly subject-dependent. This may be attributable to variations in foot sensitivity or some subjects' lack of familiarity with a VAS (Mündermann et al., 2002).

The *r* value was calculated to compare to Mündermann et al. (2002). The intra-class correlation coefficient (ICC) was also calculated to determine how closely the subjective comfort scores resemble each other, not just how linear the relationship is between the two as determined by correlations (Halligan, 2002). The ICC data for all subjects and individual subjects' within-day is presented in Table 4.7. For each individual subject ICCs for the questionnaire (all scales) were significant, as were data sets for each scale when all subject data was grouped (N = 5). It is apparent that some aspects of the scale are more repeatable, particularly overall and length comfort (ICC > 0.815). Other aspects are less consistent across days, and subjects, such as arch comfort for which the ICC ranged from 0.160-0.806. All ICCs for all subjects combined demonstrated either 'moderate' (0.60-0.69), 'good' (0.70-0.79) or 'excellent' (≥ 0.80) agreement between sessions (Portney and Watkins, 2007).

Calculation of the minimal clinically important difference (MCID) was undertaken by Mills et al. (2010) to assess the repeatability of a similar questionnaire and determine an absolute difference that surpassed error and became meaningful. In the present work, a similar approach was undertaken using all significantly different pairwise comparisons from session one and two. Two methods were chosen, method one utilised all footwear tested for each subject (5 subject*4 shoes, N = 20) and the second method used only the same condition for each subject (N = 5). The data was compared using t-tests between session and only significantly different paired comparisons were utilised to calculate the MCID. Using either method, only the "Feeling under the foot arch" (ICC = 0.778) and "Comfort of toes" (ICC = 0.886) differed significantly. The standard error of the measurement was calculated for each utilising Equation 5, resulting in values of 6.9 and 14.1 mm respectively.

Equation 5 Standard error of the measurement (SE_m)

$$SE_m = s\sqrt{(1- r)}$$

Where s = mean standard deviation of session 1 and session 2, r = the reliability coefficient of the test i.e. intra-class correlation coefficient between session s 1 and 2.

Using Mills et al (2010) method, the minimum difference for a meaningful change in comfort is therefore 14.1 comfort points. This is similar to the previous authors' values of 9.6 from their mathematical method and 10.2 comfort points from asking participants to indicate on the scale what they perceive to be a meaningful change in comfort (Mills et al., 2010).

Table 4.7 Intra-class correlation coefficients for individual subject comfort scores between sessions on day one.

Subject	Overall Comfort	Overall Width Comfort	Overall Length Comfort	Cushioning	Around Heel	Upper	Arch Comfort	Toe Joint	Under Ball	Toes	All Scales
1	0.997**	0.702	0.999**	0.007	0.067	0.923*	0.348	0.86	0.824	0.804	0.812**
2	0.918*	0.898	0.816**	0.752	0.022	0.752	-0.315	0.061	0.071	0.877	0.530*
3	0.866*	0.678	0.872**	0.859**	0.676	0.514	0.160	0.717	0.472	0.546	0.790**
4	0.919*	0.966*	0.983**	0.737	0.993**	0.994*	0.806	0.862	0.794	0.902*	0.908**
5	0.963*	0.403	0.836	NA	0.936*	0.618	-1.121	0.868	0.459	0.693	0.789**
All combined	0.960**	0.810**	0.967**	0.823*	0.900*	0.924**	0.778*	0.826**	0.818**	0.886**	NA

Note: Subject 5 missed one score for cushioning so all these scale results for this participant have been excluded. * = $p < .05$, ** = $p < .01$.

4.2.4.4 Pressure Variables

The pressure variables selected for this research included both traditionally reported peak pressures (calculated from a single-sensor at a single point in time) and regional pressures (calculated by grouping the sensors in an anatomical region). The traditionally reported peak pressure variables are based on the threshold for ulceration in diabetic patients. Plantar pressure systems were originally utilised in diabetes research to quantify plantar pressures and identify areas of high ulceration risk (Cavanagh and Ulbrecht, 1994). Thus individual areas (or single-sensors) of high pressures are relevant and need to be recorded in order for an intervention to relieve the high pressure and reduce ulceration risk. Regional pressures were calculated in the current research by grouping sensors into anatomically relevant regions (as would usually be conducted in pressure analysis). This was undertaken as it was presumed that sensation for the anatomical region that participants were being asked VAS for were based on the sensation for the entire region as opposed to a single sensor within the region. This process had the advantage of reducing the influence of the erroneous nature of some of the sensors in the Medilogic system on peak pressure values (Price et al., 2014). This methodology for treating pressure variable warrants further exploration and may be more appropriate for future work considering subjective measures of footwear comfort within protocols which do not require peak pressure variables to establish clinical risk, such as in diabetes research.

4.3 Research Findings

4.3.1 "Shock Absorption"

As evident from the previous literature review, research work on modifying impact characteristics with footwear focuses on running footwear. The addition of modern data providing peak acceleration data in walking footwear adds value (Paper 1), as does a range of thickness and hardness midsoles tested in walking situations (Paper 2). Additionally, the work undertaken in Papers 1&2 provided FitFlop ltd. with shock absorption data for their product and potential developments, satisfying the primary objective of this research. The measurement of impact in modern walking footwear is novel as most published research is on running footwear (Hamill et al., 2011; Nigg et al., 1987; TenBroek et al., 2013) and that which exists on walking footwear is dated (Light et al., 1980). Walking footwear is worn daily by the general population and therefore consideration of the shock absorption properties is important for comfort and the minimisation of any potential symptoms (Voloshin and Wosk, 1982; Whittle et al., 1994). The impact quantification aspects of the thesis demonstrated that ‘health and well-being’ footwear can meet the user requirement of absorbing shock in walking and thus reduce loading on the lower limbs. Paper 1 identified that a ‘health’ version of a sandal reduces the occurrence of a heel-strike transient when compared to a flip-flop and the results from the ASTM mechanical testing protocol identified an increased shock absorption capability if the external energy applied to the two items of footwear is equal (Paper 7, Table 3.19, pg. 155). Further exploration of the meaning of the heel-strike transient magnitude is required in terms of comfort (Whittle et al., 1994) and injury or symptom alleviation (Voloshin and Wosk, 1982) in walking to contextualise findings.

4.3.1.1 Drop-Test Protocol

The additional aspect to the research developed a novel methodology for assessing impact attenuation characteristics of walking footwear, including ‘health and well-being’ footwear (Paper 1). This was through a modified protocol from the standard ASTM test, which was footwear and population specific and could be utilised within ‘health and well-being’ footwear companies to inform their product development process. This extension to the research included the manipulation of a commonly reported mechanical testing method to more realistically test walking footwear. This is a novel approach to footwear comparisons and the pilot work promotes other authors to, in future, consider the relevance of the testing methodologies and standards they are applying to footwear. Although the methodology employed has some evident limitations (Paper 1), it was tested and demonstrated to be repeatable and acts as a starting point for the continuation of the future development of methodologies for quantifying shock absorption in walking footwear. The second paper applied this novel protocol to quantify the heel-ground impact in walking (and the drop-test replicating walking) in modified footbeds with different hardness and thickness. This demonstrates the potential application of such a protocol in footwear research and development within ‘health and well-being’ footwear to quantify data related to user expectations or marketing claims of this footwear category. The findings from these two papers have direct application to footwear manufacturers in that the results demonstrate that footwear testing methods should be gait style specific to accommodate the differences in impact conditions in walking compared to running. Additionally, that footwear testing methods should be footwear style specific to accommodate different gait kinematics evident in different footwear upper styles, consistent with objectives two and three of this research. These inferences span shock absorption testing in addition to other footwear testing protocols such as footwear cushioning and product longevity. Also, inferring the findings from this paper suggest that the effective mass of the wearer will influence significantly the mass applied to produce realistic impact characteristics and, consequently population-specific modifications may be required. This factor poses a greater influence to walking or 'health and well-being' where a greater diversity of wearers are expected (such as adults who are obese) as opposed to running footwear where the population characteristics (such as body mass and limb inertial properties) may be more uniform. This highlights the importance of considering defining the final wearer when undertaking research related to objective four of this thesis and the exploration of footwear biomechanics in 'health and well-being' footwear.

Despite the most common impact test in footwear literature being the ASTM protocol designed to assess shock absorption properties of running shoes (Section 2.2.3, pg. 18-22), some adapted mechanical protocols are evident in literature. Researchers at Southampton Solent University developed a device replicating the ASTM protocol to test the cushioning properties of athletic socks (Blackmore et al., 2013). Additionally, Schwanitz et al. (2010), developed a Hydraulic Impact Test to more closely resemble the contact force in running for durability tests. Other bespoke methodologies and devices also attempt to replicate the impact evident in running to assess energy absorption in footwear insoles (Chiu et al., 2001; Chiu, 2005). Research studies which attempt to replicate the energy apparent in walking examine the shock absorption properties of the human heel pad (Jørgensen and Bojsen-Møller, 1989; Jørgensen and Ekstrand, 1988). To the author's knowledge, no other bespoke drop-test protocols are evident in the research literature to quantify shock absorption properties of walking footwear.

The adapted methodology was more representative of walking data than the ASTM protocol, emphasising the importance of the second and third objective of this thesis concerning modifying running footwear test protocols to quantify characteristics of walking footwear. Despite this, the protocol results do not match the human walking results (Table 4.8) and the validity of the proposed methodology in Papers 1&2 could be further increased as previously discussed, particularly by more closely replicating the conditions of initial contact.

Table 4.8 Methodology comparison for drop-test protocols from Paper 1

Method	Peak Acc. ($m \cdot s^{-2}$)	time Peak Acc. (ms)	Peak Force (N)	time Peak Force (ms)
Δ ASTM	80.7-302 ($p < .001$)	7.2-13.3 ($p < .001$)	416.6-2411.9 ($p < .001$)	5.3-14.6 ($p < .001$)
Δ Adapted	0.6-10.5 ($p < .001$)	3.4-19.7 ($p = .771$)	57.6-151.2 ($p = .786$)	1.1-18.9 ($p = .001$)

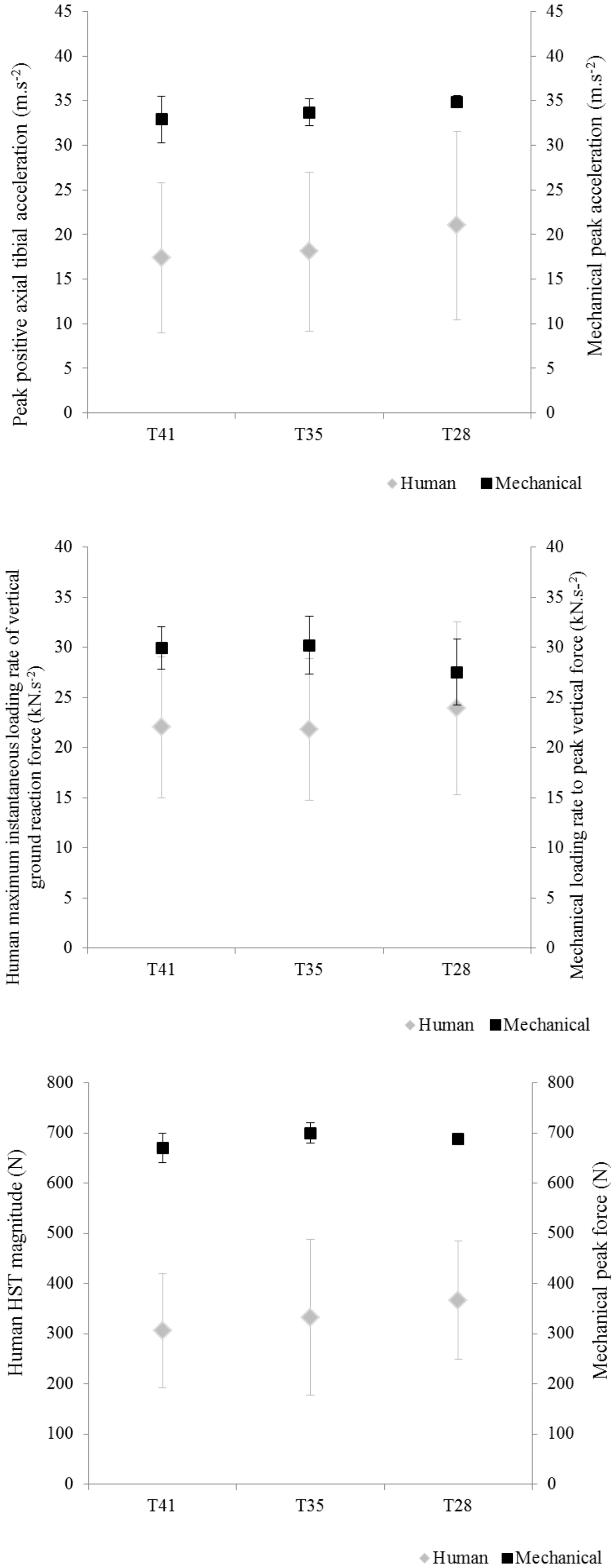
Where: Δ denotes a difference from the human data, and Acc = Acceleration.

As evident (Table 4.8), the peak acceleration data and the time of the peak force variable differed significantly using the modified protocol from the human data in Paper 1. Furthermore, correlations between the peak force (thickness $r = .532$, $p = .643$, hardness $r = -.408$, $p = .592$), force loading rate (thickness $r = -1$ $p = .010$, hardness $r = .916$, $p = .084$) and peak acceleration (thickness $r = .975$ $p = .172$, hardness $r = .672$, $p = .327$) from the drop-test and human walking in Paper 2 were mostly not significant (Figure 4.12 & 4.13). Hennig et al. (1993) compared peak axial tibial accelerations in differing rearfoot constructions of running shoes to peak acceleration scores with a missile replicating the ASTM method. The study identified a low ($r = 0.26$), non-significant correlation with in vivo results from 19 shoes and 27 participants. T-tests between the mechanical and human data from Paper 2 identified that the peak tibial acceleration and force loading rate differed significantly between the mechanical and human methods. When testing the hardness conditions, the mechanical protocol overestimated peak axial tibial acceleration by 9.4 $m \cdot s^{-2}$ ($p < .001$) and underestimated the maximum instantaneous loading rate of the ground reaction force by 9.1 $kN \cdot s^{-2}$ ($p < .001$). The mechanical variables better replicated the patterns evident in the human results in the hardness variations than the thickness (Figures 4.12&4.13), despite the hardness variations inducing kinematic differences in walking (Paper 2, Table 3.6, pg. 83). The comparative values for the thickness data were overestimations of both peak acceleration (16.4 $m \cdot s^{-2}$, $p < .001$) and loading rate (6.0 $kN \cdot s^{-2}$, $p = .01$). This is potentially due to increased error induced in the mechanical protocol when adjustments were manually made for thickness, the hardness conditions were all tested from exactly the same drop height. Differences between the two protocols are likely due to the variance in time in the two impacts, human peak acceleration occurred within 40 milliseconds of impact, in the mechanical protocol peak acceleration occurred earlier, within 33 ms of impact. The missile in the mechanical situation is a rigid mass, whereas the human limb is non-rigid and will attenuate the impact in soft tissue etc. The overestimation of loading rates in the mechanical protocol has significant implications for the testing of shock absorption properties, despite the near-linear relationships identified by the correlation analysis. The rate-dependent of the nature of the shock absorption properties of viscoelastic materials means that it is essential that the loading characteristic mimic that which will be evident in the ‘real life’ use of the footwear in walking (Whittle, 1999).

Numerous limitations exist in the methodology (Paper 1), which likely account for these differences. In particular automating the device would be expected to increase the repeatability of the values and thus reduce the standard deviation and minimal detectable change. To be an applicable and worthwhile tool for the footwear technician in order to be fully integrated into the research and development process of a company, this methodology thus needs some development and further validation to ensure results are more representative of the mean human results for which the footwear was designed.

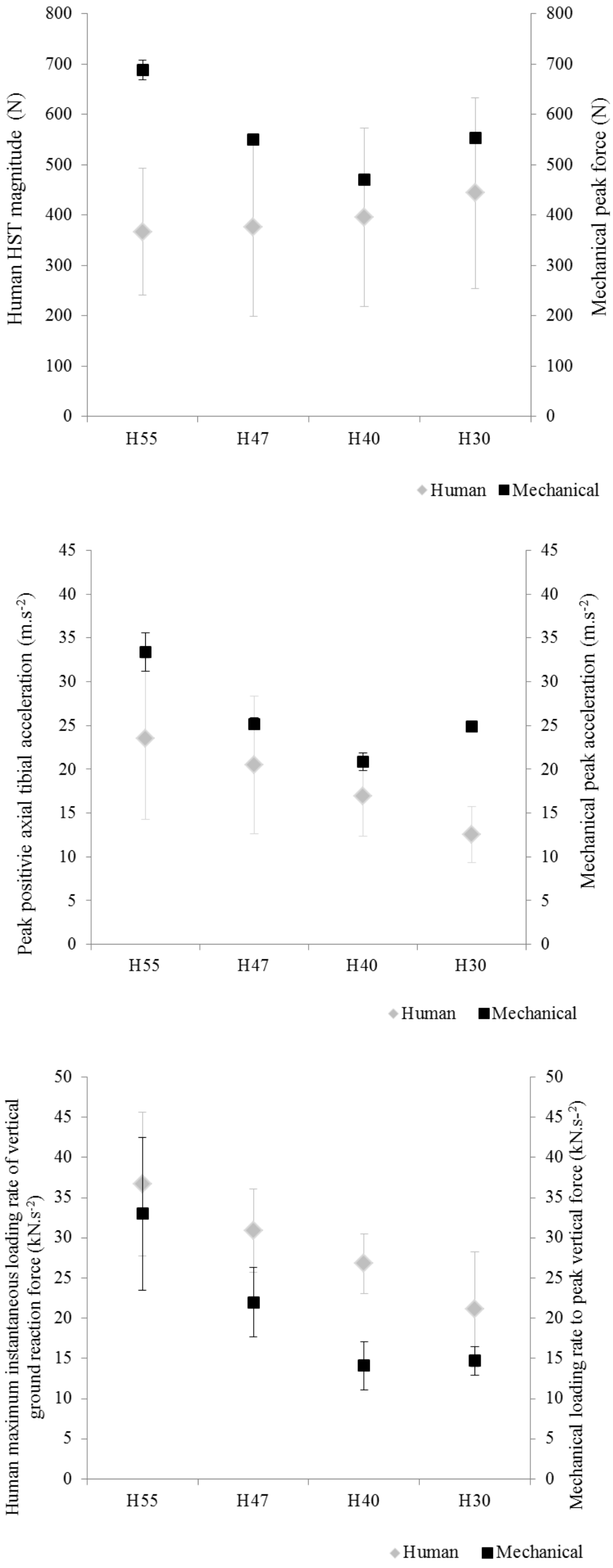
The statistical analysis of the force variables, from the work related to the biomechanical concept of increasing shock absorption (Papers 1&2), was not undertaken due to the small and inconsistent population numbers for this data. Calculation of the maximum instantaneous loading rate may have been more useful from this perspective, however, this does not precisely characterise the heel-strike transient as desired in this work. The nature of the quantification of variables only from heel-strike transients means that the data is not interpreted from all subjects in Papers 1&2 and this limits the subject number. As aforementioned, previous research focuses on running footwear where a transient is consistently evident and this methodological choice is not relevant (pg. 13-15). Additionally, loading rate was included in Paper 2, reducing these limitations to the comparison while maintaining the original variables in Paper 1. The inclusion of loading rate is particularly important for viscoelastic material assessment (Paper 2, pg. 84-87).

Figure 4.12 Scatter plots for mean acceleration, force and loading rate variables resulting from mechanical and human test methods for thickness variations.



Where: T41=41 mm, T34=34 mm and T28=28 mm and error bars indicate 1 standard deviation.

Figure 4.23 Scatter plots for mean acceleration, force and loading rate variables resulting from mechanical and human test methods for hardness variations.



Where: H55=55 Shore A, H47=47 Shore A, H40=40 Shore A and H30=30 Shore A and error bars indicate 1 standard deviation.

4.3.1.2 Other Findings

Paper 1 highlighted that vertical heel velocities towards the floor varied in different footwear styles and this therefore reduces the generalizability of the findings in Paper 2. The vertical heel velocity at contact with the floor recorded in the human testing, and then replicated in the drop-test protocol, was specific to the flip-flop upper for which the data was collected. Consideration of the influence of the thickness and hardness of the EVA footbed would require further exploration if the footbed was to be imbedded in a covered shoe. The generalizability of the results is limited by the dimensions of the footbeds tested, which are thicker than may be utilised in standard walking footwear (28-41 mm, Paper 2, Table 3.3, pg. 77). However, these dimensions were relevant to the external research project and ‘health and wellbeing’ footwear in general and related to the original FitFlop™ product (41 mm); and potential future products. This justification is consistent with the hardness of EVA tested, which represented a conceivable range about the standard FitFlop™ footbed. The dissemination and value of the research undertaken to the FitFlop™ product cycle and is evident by the new FitFlop™ products now available on the market which include altered dimensions. The work was considered by the footwear technology and design team and due to the minimal evident difference in the shock absorption capability from 41 mm to 35 mm (Paper 2, pg. 82) some new products have been manufactured on a slimmer midsole (http://www.fitflop.co.uk/womens/ballerinas/). This demonstrates a direct application of the work undertaken into the company’s research and development process, which is an unpublished form of research dissemination and integral to the primary objective of this research. This integration of research into footwear construction and development would be a positive step for the field moving forward as currently research can generally appear relatively independent from manufacturers and companies product development processes.

In addition to the shock absorption properties of the footwear, altering the hardness and thickness of the footbed will influence other variables for wearers and these aspects should be investigated as footbed properties are altered in the design process. Paper 2 identified that hardness changes alter the velocity of the heel toward the floor at impact, providing useful data to compare to the vast array of running research considering this topic, consistent with the second objective of this research. Further exploration of the influence of footbed changes to kinematics after heel-contact is essential, particularly considering findings from running literature highlight an increase in frontal plane motion of the foot during midstance with softer shoes (De Wit et al., 1995). It is likely that softer and thicker footwear will be more comfortable for wearers and more effectively alleviate perception of impact in walking (Milani et al., 1997; Sterzing et al., 2013). However there is no measure of clinically meaningful difference in this study, it may be that even the highest mean peak positive vertical tibial acceleration (23.5 $m \cdot s^{-2}$, Paper 2, Table 3.6, pg. 83) is not detrimental to the average wearers health or comfort during walking and this should be established. A meaningful threshold for comfort, or discomfort, in walking impacts would provide useful information when considering the biomechanical influence of ‘health and well-being’ footwear in-line with objective four of the current research.

Within Papers 1&2 data relating to the occurrence of heel-strike transient in walking has been presented for a range of different footwear upper and midsoles. In the differing footwear conditions the heel-strike transient occurred in between 6.9% (in trainer) and 98.5% (in barefoot) of the trials (unpublished data from Paper 1). The heel-strike transients identified in the trainer occurred closer to heel impact than in other footwear conditions, these were of small magnitudes and caused by only two participants, potentially anomalies due to footstrike pattern combining with the heel-flare of the shoe (Whittle, 1997). In the varying hardness (6.7-71.7%) and thickness (43.3-51.7%) conditions tested in Paper 2 occurrence of heel-strike transient spanned similar ranges. The heel-strike transient in the vertical ground reaction force was an evident characteristic in all subjects walking barefoot and in 35% of all steps in the study. This was a comparable incidence and between subject variability to that reported in shod walking at a range of walking velocities in similar protocols (McCaw et al., 2000; Verdini et al., 2006). Verdini et al. (2006) reported heel-strike transients in a total of 89.3% of the trials recorded in barefoot walking at a self-selected cadence through a definition which involved frequency domain analysis and the identification in both the vertical and anterior-posterior components of the ground reaction force. The magnitude of the heel-strike transient is determined by the rate at which the momentum of the foot changes and thus a combination of individual subject factors in addition to footwear and surface factors (Whittle, 1999). The individual subject factors which determine the magnitude of this feature are those which control the effective mass and velocity of the foot at initial-contact; thus, the lower limb kinematics and muscular control (Jefferson et al., 1990; Whittle, 1997). Specifically, appropriately timed activation of the quadriceps to control the deceleration of the lower limb has been identified in patients who do not demonstrate a severe heel-strike transient (Jefferson et al., 1990; Verdini et al., 2006). As aforementioned, these individual subject differences have implicated the magnitude of the outcome variables and therefore the drop-test protocol within this thesis. The footwear features affecting the presence or not of the heel-strike transient, aside from those which would have influenced touchdown kinematics, are the shape of the footbed (thickness) and the material properties of the footbed (elasticity and viscosity) (Whittle, 1997). The foot-sole-ground angle is influential in that it determines position and therefore the functional shape of the shoe under the foot during impact with the ground. Factors such as thickness, shape, material properties and perceived comfort will all combine to affect gait kinematics and hence shock attenuation. This means that any measured differences in force and tibial acceleration cannot be attributed only to footwear material influence.

4.3.1.3 Summary

The measurement of impact in walking footwear is necessary as most published research is on running footwear (e.g. Hamill et al., 2011; Nigg et al., 1987) or reports walking footwear results which cannot be related to modern commercially available footwear (e.g. Light et al., 1980). The research undertaken identified that modifying footwear thickness did not alter the velocity with which the ground was impacted, however the hardness modifications did. This knowledge should be considered by footwear technologists as they make design decisions for existing or future products. Additionally, this phase of the research included the manipulation of a commonly-reported mechanical testing method to better suit the footwear conditions being compared. This is a novel approach to footwear comparisons and the pilot work promotes other authors to, in future, consider the relevance of the mechanical testing they are undertaking on the footwear. Although the methodology employed has some evident and discussed limitations, it was tested and demonstrated to be repeatable. It acts as a starting point for the continuation of the development of methodologies for quantifying shock absorption and other variables in walking footwear in the future and the development of the work related to objectives two and three of this thesis.

4.3.2 "Instability"

Papers 3&4 attempt to resolve some highlighted weaknesses of previous research considering unstable footwear by providing relevant measures, in appropriate situations, on applicable users. These papers addressed objectives one and four of the research, providing data on the influence of FitFlop^{TM} footwear on a wearer and also quantifying variables related to footwear biomechanics concepts relating to 'health and well-being' footwear, in this case whether the footwear is unstable. From these two studies it can be concluded that the instability induced by unstable footwear cannot be generalised and is specific to the footwear style, probably most notably the outsole shape. The different designs compared affect different aspects of stability at different points of the gait cycle (Table 4.9). Therefore, it is apparent that the instability induced by unstable sandals is variable and design-specific.

Since the conception of the research for Paper 3&4, a vast array of literature pertaining to unstable footwear has been published. This is concurrent with an increasing popularity of unstable or "toning" footwear from 2009-2013, particularly in the U.K. and U.S.A.. The body of knowledge considering unstable footwear has increased and now includes papers on FitFlop™ (Burgess and Swinton, 2012), Reebok EasyTone™ (Horsak and Baca, 2013), Reflex Control Schuh™ (Turbanski et al., 2011) and Scholl Starlit™ (Forghany et al., 2014). Despite these additions, the majority of the peer-reviewed literature in this category remains focused on MBT™ footwear or anterior-posterior rocker shoe technology. The extensive research existing on MBT™ was recently evident by an entire volume of the Footwear Science journal being almost entirely dedicated to research concerning the footwear (Volume 4, Issue 2, 2012). In order to address the limitations previously highlighted in literature relating to conditions tested, tasks and populations relating to unstable footwear, characteristics of studies will be addressed for literature that had been published since 2011, including Papers 3&4 of the current research.

Table 4.9 "Induce Instability" paper findings relating to key variables.

Variables	Paper 3 (Standing)	Paper 4 (Gait)
Centre of Pressure	Anterior-posterior range greater in MBT than all other conditions (+11-15.1 mm, *p* = .008-.045). No differences in total path length, medial-lateral range or velocities between conditions.	Anterior-posterior range in MBT™ less than SK (-8.9 mm, *p* = .001) and CON (-10 mm, *p* = .004). Medial-lateral range greater in RE than FF (+5.8 mm, *p* = .030). No difference in velocities between conditions.
Kinematics	ROM in MBT greater in the sagittal plane at the ankle than all other conditions (+4-5°, *p* > .001). ROM in RE greater in the frontal plane at the ankle than CON (6.4°, *p* > .001) and SK (5.9°, *p* = .005).	ROM across stance did not differ between conditions for ankle (sagittal and frontal planes), knee and hip (sagittal plane).
Electromyography	SK, FF and MBT greater RMS for gastrocnemius than CON during standing (+27-35%, *p* < .05). MBT greater RMS for soleus than CON (18%, *p* < .05) and FF (14%, *p* < .05) during standing. RE (18%, *p* < .05) and MB (10%, *p* < .05) greater than FF; and MB (13%, *p* < .05) greater than CON for rectus femoris during standing.	MB decreased median tibialis anterior (-46%, *p* = .005) RMS and increased peroneus longus (32%, *p* = .015-.020) and gastrocnemius (69%, *p* = .010-.020) activation at loading response compared to other conditions (values are CON). FF (-24%, *p* = .005) and SK (-17.4%, *p* = .005) decreased gastrocnemius and soleus activation during mid-stance compared to CON and MB (values are CON). FF (46%, *p* = .025), RE (33%, *p* = .025) and SK (51%, *p* = .010) increased peroneus longus activation at pre-swing compared to CON.

Where MBT = Masai Barefoot Technology™, RE = Reebok EasyTone™, SK = Skechers Tone-Ups™, FF = FitFlop Walkstar™, CON = Earth Kalso™, RMS = Root Mean Square and ROM = Range of motion.

4.3.2.1 Conditions

As noted above, the focus of research literature pertaining to unstable footwear remained MBT™ footwear (Branthwaite et al., 2013; Federolf et al., 2011; Landry et al., 2012; Taniguchi et al., 2012). This emphasis may be due to the availability of the footwear already within institutions, the resemblance of the shoe to clinical anterior-posterior rocker shoe footwear, or simply a product of the duration between study conception and paper publishing in the peer-review process. The majority of this research arises from the University of Calgary, or affiliated authors, who acknowledge links with MBT™ and associated foundations within their papers. It is unfortunate however, that the characteristics and influence of other unstable footwear designs has been largely overlooked. Research literature does consider other outsole styles and technologies, either compared to stable control shoes (Burgess and Swinton, 2012) or to MBT™ footwear (Turbanski et al., 2011); nonetheless these papers are in the minority. An inherent strength of Papers 3&4, in that they considered alternative footwear styles alongside MBT™ footwear providing data to investigate the biomechanical concept of instability in this footwear category. Descriptive studies comparing MBT™ footwear during walking are now relatively exhaustive for asymptomatic wearers and future consideration should encompass other styles or technologies in order to specifically address changes in footwear design features leading to specific unstable outcomes in populations. The alternatives may induce instability which is more suitable for some wearers, or simply more accessible to the population due to the lower retail cost (e.g. for a leather upper flip-flop style shoe: FitFlop™ Lulu = £50, MBT™ Kamili = £153).

The studies herein provide the only data quantifying instability in numerous unstable sandals as opposed to covered upper unstable shoes. The sandal upper may influence the stability further by not constraining the foot or providing additional support and therefore this data may have reduced external validity and inference to covered styles even with consistent outsole features. However, this data might be particularly relevant for wearers and clinicians in warmer climates where open-footwear might be the primary footwear choice throughout the year. In studies utilising MBT™ footwear, some authors utilise sandal style footwear due to the advantages it poses for marker placement and experimental setup as opposed to for specifically quantifying variables in open footwear (Cox et al., 2010; Roberts et al., 2011). Buchecker et al. (2012) investigated contrasts evident in a wearer when walking in different styles of MBT™ footwear. Their findings suggested that specific variations in MBT™ sole construction criteria differently challenged the postural control system as contrasting responses were evident in centre of pressure data and self-reported perceived instability. Similarly, Gardner et al. (2014) reported differences in ground reaction force variables and anterior-posterior centre of pressure displacement between two styles of Active Balance™ unstable shoes. However the data was compared from two different studies utilising different participants and different control shoes. Despite these methodological peculiarities, this research embodies a more recent trend, which disregards the apparent assumption in some earlier literature (Pocari et al., 2010), and comments about the generalizability of unstable footwear studies across the category (Nigg, et al., 2012). This being that unstable shoes influence wearers consistently, despite the differing outsole features. It is evident, from the results of the papers herein and other research on custom-modified footwear (Hömme et al., 2012), that diverse shoe modifications induce different instability and this should be inferred to footwear design for specific populations.

Papers 3&4 of the current research provide data directly associated with retail names such that wearers and clinicians alike can be informed and use the study results to direct their recommendations or buying behaviour. This approach has recently been replicated by Plom et al. (2014), utilising a similar protocol and footwear conditions, although testing covered versions of the unstable footwear. Other authors have attempted to isolate the influence of altering specific footwear aspects such as outsole wasting on stability of wearers through custom-modified footwear, which despite not being directly applicable to a wearer, can supplement valuable information into the footwear design process (Hömme et al., 2012). Therefore an emphasis of this research in future would be to disseminate it into footwear design processes and ultimately influence the footwear that is available in retail to a consumer. This approach enabled comparison of medial-lateral to anterior-posterior outsole or midsole features and their effects (Hömme et al., 2012). Attempting to isolate independent variables enables the specific influence of one factor to be determined as opposed to differences exiting in upper material, stiffness, mass and last geometry. For example, the mass of the footwear has previously been proposed as a reason for an evident increase in tibialis anterior muscle activation evident in the footwear (Romkes et al., 2006) or increased energy expenditure (Santo et al., 2012; Thuesen and Lindahl, 2009). Results from investigations of this theory vary in their findings. Forghany et al, (2014) matched the mass of the control footwear when reviewing an anterior-posterior rocker shoe. Allometric scaling comparisons for mass identified increased oxygen consumption in MBT™ (86.6 $ml \cdot min \cdot kg^{-1}$) compared to a trainer (82.6 $ml \cdot min \cdot kg^{-1}$) (Gjøvaag et al., 2011). However comparisons between control shoes with equivalent mass to MBT™ identify no significant difference in energy expenditure or muscle activation (Forghany et al., 2014; Santo et al., 2012). Different situations in these studies may account for these contradictory findings. Gjøvaag et al. (2011) included fast uphill walking in their treadmill protocol, which is likely more challenging compared to flat walking on a laboratory floor (Forghany et al., 2014; Santo et al., 2012) in a rocker-soled shoe. In studies where the research design does not enable the bespoke production of a control shoe there is an increasing trend to report further information regarding the footwear, which benefits the interpretation and application of results. For example, reporting the mass of footwear conditions is common (Paper 6; Buchecker et al., 2012; Sacco et al., 2012) and provides additional information to generalise results. Another footwear characteristic which warrants isolation and comparison includes determining the influence of the height of centre of mass (Stöggl et al., 2010). MBT™ footwear, and most unstable footwear conditions, has soles with increased depth, it may be that raising the height of the centre of mass reduces the stability of the body independent of any other specific outsole features, consistent with wider literature which demonstrates a reduced stability with increased heel height (Lee et al., 2001; Nag et al., 2011). This concept could be investigated with an independent variable of sole-depth modified in otherwise matching footwear conditions.

4.3.2.2 Participants

The majority of study populations continue to be younger, physically active, healthy volunteers. This limits the outcome measures of studies because, as highlighted by Burgess and Swinton (2012), participants may have required a greater input of instability to produce detectable outcomes compared to the 'health and well-being' categories potential footwear wearers who are likely to be an older generation. This makes study designs less sensitive in nature to any unstable features of the footwear conditions being applied. This may explain the lack of difference in muscle activation compared to a stable shoe in numerous recent studies. The papers in the current research attempted to select participants who were older and were recruited outside the sports science department. Hence, the participants reflected better the wearer of the shoes (Papers 3&4). Other authors have diversified the populations being tested by considering symptomatic patients wearing MBT™ footwear such as participants with lower back pain (Nigg et al., 2009), in addition to specific population groups such as males who are overweight (Buchecker et al., 2010).

At the outset of this research, the focus of research literature quantifying unstable footwear influences was on female participants. More recently the diversification of study populations has included male participants (Stöggl et al., 2010; Taniguchi et al., 2012; Zhang et al., 2012). Some research has grouped male and female participants as one study populations, however differences between the influence of unstable footwear on male and female wearers would suggest that genders should be separated. A comparative study by Nigg et al. (2010) identified an increased centre of pressure anterior-posterior excursion in the female (45.7±19.0 cm) compared to male (39.2±11.8 cm) participants during bipedal standing. The increased instability documented in the females induced by unstable footwear suggests that males may be less influenced by these footwear styles than their counterparts. This may be a function of the increased centre of mass height in MBT^{TM} being greater relative to the standing position in females due to their reduced height. Furthermore, it alludes to the requirements for companies to consider gender differences when designing the unstable aspects of their footwear, similar to the gender-specific approaches evident in female-specific footwear for athletic activities (Krauss et al., 2010). The current focus of this footwear category, however, is female wearers.

4.3.2.3 Tasks

The majority of studies published since the conception of the studies included utilise similar protocols where participants stand (Paper 5; Buchecker et al., 2012; Germano et al., 2012; Horsak and Baca, 2013) or walk (Paper 6; Horsak and Baca, 2013; Taniguchi et al., 2012) in the footwear conditions. The walking tasks vary in their format, with some authors selecting treadmill walking (Burgess and Swinton, 2012), others specific walking tests (Forghany et al, 2014) and standard walking in the laboratory at self-selected (Forghany et al., 2014; Horsak and Baca, 2013a) or prescribed (Gardner et al., 2014; Nigg et al., 2010) walking velocities. The diverse range of tasks now captured in unstable footwear give a clearer representation of how the footwear may influence wearers in their daily lives. Kinematic adaptations to wearing the MBT^{TM} shoe in walking are apparent when wearing them immediately with MBT^{TM} training (Taniguchi et al., 2012) and without (Sacco et al., 2012), which impacts on the usability of the footwear. It is evident that the most valuable future research will combine a relevant population undertaking relevant tasks to clarify the influence of unstable footwear.

4.3.2.4 Variables and Findings

Research studies pertaining to walking in unstable footwear exist in great numbers with a variety of protocol and variables utilised to compare muscle activity intensity and magnitude. In these approaches, the participants included and the shoes compared mediate the outcome of research and make drawing conclusions across the literature rather perilous.

Electromyography data has consistently been quantified in unstable footwear and, consistent with the current research, focuses on the larger more accessible muscles of the lower limb such as the medial gastrocnemius, rectus femoris and biceps femoris (Branthwaite et al., 2013; Elkjær et al., 2011; Germano et al., 2012). Results and conclusions from single-leg standing research tend to demonstrate no, or limited, increase in muscle activation in the rectus femoris, vastus medialis and lateralis, biceps femoris and gluteus maximus in a range of unstable shoe models including MBT™, Reebok EasyTone™ and Skechers Shape-Ups™ (Paper 3; Germano et al., 2012). Increased muscle activation has been reported in the gastrocnemius in standing in the range of 6-38% in unstable shoes (Paper 3; Sousa et al., 2012). This is smaller than the response previously reported by (Nigg et al., 2006b) for the tibialis anterior (70±85%, $p < .05$), and for the gastrocnemius muscle (38±41%, $p > .05$) in bi-lateral standing in MBT™ footwear, which as noted in Paper 3, is more extreme profile design. Increased calf (gastrocnemius and soleus) muscle activity in unstable shoes was recorded in studies utilising older, and potentially less physically active, participants (mean age 36.6±7.7 years; Sousa et al., 2012, 29±6.7 years; Paper 3) compared to younger participant studies (22.4±2.2 years; Germano et al., 2012). This supports the concept that the protocol must challenge participants to effectively differentiate between footwear conditions; therefore they must be representative of end-users or wearers. It is not clear whether Germano et al (2012) utilised MBT™ shoes as a condition as the footwear conditions included are not defined in the paper; it may be that the four unstable shoes tested had more subtle outsole designs. Increased activation in the gastrocnemius and soleus muscles in MBT™ was evident in single-leg standing compared to a control shoe (Paper 3) and barefoot (Sousa et al., 2010). However, as aforementioned the control shoe in Paper 3 may be influential when interpreting results. In particular the decreased gastrocnemius and soleus activation during gait in FitFlop™ and Skechers™ may be due to the 5° dorsiflexion in the control shoe (Paper 3, pg. 102; Paper 4, pg. 116).

In standing, concurrent with the limited significant increases in muscle activation, is minimal range increases in centre of pressure deviations in unstable shoes (73.7-79.0 cm Germano et al., 2012; 124.5-132.2 cm Paper 3; 35.8-38.8 cm Turbanski et al., 2011) compared to control conditions (90.0 cm Germano et al., 2012; 121.9 cm Paper 3; 36.2 cm Turbanski et al., 2011). These are smaller differences than identified in previous research on MBT™ footwear ranges in bipedal standing compared to the subjects own stable shoe (103% medial-lateral, 105%

anterior-posterior; Landry et al., 2010) and compared to a running shoe (104% medial-lateral, 54% anterior-posterior; Nigg et al., 2006). More recently, Plom et al. (2014) identified significant differences in the centre of pressure total excursion when participants stood in single- and double-limb stance in Reebok Easy-Tone™, FitFlop SuperT™ and Skechers Shape-Ups™. Centre of pressure variables were larger in the Skechers™ condition, for example the anterior-posterior cumulative displacement in double-limb stance (185.1±33.50) exceeded FitFlop SuperT™ (147.1±22.2, $p < .001$), Reebok EasyTone™ (147.9±25.3, $p < .001$) and barefoot (145.3±15.8, $p < .001$) conditions. No additional data was collected alongside the centre of pressure data within this study and the total path lengths are long for the 10 second trial length that was reported. These Skechers™ shoes are not comparable to those utilised in Papers 3&4 and more closely resemble an anterior-posterior rocker shoe design similar to MBT™, which rationalises why the results are more similar to these findings.

In gait, the most consistently reported result from the quantification of muscle activity is that tibialis anterior activation reduces in early stance in MBT™ (Nigg et al., 2006; Sacco et al., 2012; Romkes et al., 2006) and other anterior-posterior rocker-soled style unstable footwear (Zhang et al., 2012; Forghany et al., 2012). This is in terms of intensity, IEMG, average and maximum activation when compared to a flat control shoe either provided for the study or the participants own. Findings for the same muscle compared to a mass-matched shoe are contradictory however, reporting reduced average and RMS EMG (Santo et al., 2012) or no difference in IEMG (Forghany et al., 2012). This response of reduced tibialis anterior activation is not evident in walking in FitFlop Walkstar™ (Paper 4, Figure 3.7, Table 3.13, pg. 113-114) or Reebok EasyTone™ (Horsak and Baca, 2013a) footwear, which do not incorporate large anterior-posterior rocker soles as part of their design. However, this pattern of reduced tibialis anterior activation is consistent with a reduced dorsiflexion moment in early stance, evident in some research considering MBT™ (Forghany et al., 2013; Zhang et al., 2012) and other roll-over footwear (Forghany et al., 2013).

Aside from the tibialis anterior, muscle activation variations are less consistent in alternative lower limb muscles. No significant differences have been reported in wavelet intensities of lower limb muscles wearing MBT™ (Nigg et al., 2006), in mean RMS EMG wearing FitFlop™ compared to controls (Burgess and Swinton, 2012), or in peak linear envelope magnitude or IEMG wearing MBT™ (Sacco et al., 2012). Similarly, Elkjær et al. (2011), identified no significant differences in treadmill walking in peak linear envelope magnitude or IEMG of gastrocnemius lateralis, biceps femoris and gluteus maximus in ten male participants walking in Reebok EasyTone^{TM} (age 24.5±3.8 years, B.M.I. 24.03±1.09 kg·m^{-2}). However, this data was quantified across the whole of the stance phase meaning that increases and decreases at different phases in the gait cycle in different footwear designs would have been eliminated from analysis. Contrasting these findings, Forghany et al (2013) identified increased maximum IEMG in MBT^{TM} and the Scholl^{TM} rollover footwear in both soleus (+13% MBT^{TM} and +8% Scholl^{TM}, $p < .05$) and gastrocnemius (+8% MBT^{TM} and +6% Scholl^{TM}, $p > .05$) muscles in the stance phase. Nevertheless, this is consistent with the 'trend' reported by Nigg et al. (2006). Romkes et al., (2006) data appeared more sensitive with significant increases in activation in initial stance in gastrocnemius and decreased rectus femoris IEMG in MBT^{TM}. The analysis process implemented sub-divided stance into 16 phases, which may be more sensitive to differences between conditions and has the benefit of potentially alluding to the features of the footwear which may be inducing unstable responses. Furthermore, a similar approach in Paper 4 identified alterations to activity of muscles acting in the sagittal plane to stabilise in MBT^{TM} and in the frontal plane in the other designs of unstable footwear (Paper 4, Figure 3.7, Table 3.13, pg. 113-114). Opposing this speculated increased sensitivity to differences when separating stance into phases, Horsak and Baca (2013) found no difference in mean muscle activation over any sub-phases in gait in the Reebok EasyTone^{TM} footwear compared to the participants own footwear. These findings, again, affirm that unstable footwear induces differences at different phases in stance based on the specific design of the footwear being tested, the task undertaken and the participants. In particular, the male participants (N = 7) in this study Horsak and Baca (2013) may explain the contrasting outcomes in comparison to Paper 4 as male participants appear to be less influenced by the perturbations induced by unstable footwear (Nigg et al., 2010).

In addition to the variables calculated, the muscles selected for analysis may explain the lack of significant difference identified between unstable footwear and stable controls in some research. As aforementioned, Burgess and Swinton (2012) found no difference in lower limb muscle activation in medial gastrocnemius, biceps femoris, rectus femoris, tibialis anterior or gluteus maximus in stance single-leg standing, walking, treadmill walking, zig-zag walking, or stair climbing in FitFlop^{TM} compared to barefoot or a flip-flop. The muscles selected for analysis in papers pertaining to the theme of unstable footwear are commonly variations on those above (Buchecker et al., 2012; Forghany et al., 2014; Horsak and Baca, 2013). It may however be that these muscles are not the muscles utilised to control any instability induced by the footwear characteristics. Landry et al. (2010) proposed that muscles closer to the joint axis are stimulated to increase activation by MBT^{TM}. This is consistent with findings in Paper 3 that peroneal activity increased at pre-swing in three of the unstable shoes tested (Paper 4, Table 3.13, pg. 114). Data for this muscle is unfortunately not reported in any other studies considering instability in the FitFlop^{TM} (Burgess and Swinton, 2012; Porcari et al., 2009) and the finding is not repeated in a study which quantified muscle-activity in pre-swing in Reebok Easytone^{TM} on a mixed gender cohort (Horsak and Baca, 2013). As aforementioned, Landry et al (2010) validated the functioning of MBT^{TM} on increasing activity in smaller lower limb muscles by 50-800% in bipedal stance. Significant increases were evident in the extensor digitorum longus (≈550%) and peroneus longus (≈100%). Subsequent research should continue to explore the influence of unstable footwear features on more intrinsic musculature of the lower limb. However, appropriate presentation of data should be considered as percentage difference can be misleading if values are minimal, likewise the clinical influence of such changes should be evidenced.

Instability footwear literature lacks the quantification of muscle activation timing variables, which are not normalised to stance time. This is, in-part, explained by the dynamic and static protocols often being undertaken concurrently and static protocols having fixed timings and the duration of activation being consistent. However, even dynamic protocols conducted independent of static protocols focus on quantifying muscle activation magnitudes as opposed to durations (Burgess and Swinton, 2012; Germano et al., 2012). Duration of muscle activation has been considered in studies of orthotics (Tomaro and Burdett, 1993) and footwear (Li and Hong, 2007). It may be that instability footwear maintains a consistent magnitude when compared to standard footwear, but prolongs the duration for which the muscle is active. The use of the IEMG signal over stance (not normalised) would provide a combination of magnitude and duration of muscle activation (such as in Paper 4). However if the magnitude decreased in one area and increased in another, results may be equal when graphs denote contrasting patterns (e.g. Nigg et al., 2006). Separating stance into phases enables comparisons across time-points and may more effectively relate back to specific footwear design features and their function during the gait cycle, thus enabling function in these footwear styles to be clarified and adds knowledge which could be integrated into informing the design process.

Kinetic analysis of walking in MBT™ footwear has been undertaken more extensively in recent years. Various authors have reported an increased loading rate to the first peak of the ground reaction force and the magnitude of the first peak when walking in MBT™ compared to barefoot or control shoes (Sacco et al., 2012; Taniguchi et al., 2012; Zhang et al., 2012). Vernon et al. (2004), however, report a reduction in transient peaks in the MBT™ condition compared to a normal shoe. The aforementioned studies considered the loading rate to the loading peak of the ground reaction force (rate of weight acceptance), not the maximum instantaneous loading rate, which is more likely to aptly reflect the transient features identified by Vernon et al. (2004). The use of tibial mounted accelerometers to specifically assess the shock absorption properties of unstable footwear is rare in literature. However, Hömme et al. (2012) identified no significant difference in peak tibial acceleration magnitude or time of occurrence in their bespoke anterior-posterior unstable, medial-lateral unstable and control conditions. The anterior-posterior outsole shape of this footwear was similar to MBT™, however it lacked the heel element integral to the footwear and thus, unfortunately, results cannot be directly extrapolated to the commercially available shoe. The reduction in transients may, in part, explain the benefits reported in clinical groups wearing MBT™ (Collins and Whittle, 1989; Wosk and Voloshin, 1981). It is likely a result of the thickness of the footwear and the contact position being midfoot result in this apparent reduction in loading at contact with the floor (Paper 2; Vernon et al., 2004). The presence and material construction of the heel element would influence the shock absorption at heel contact in MBT™ footwear specifically.

4.3.2.5 Summary

Despite conflicting study results, it is evident that unstable footwear makes subtle changes to muscle activation, which result in large intra-subject responses (Nigg et al., 2012) dependent on both activity and footwear worn (Papers 3&4; Germano et al., 2012). As previously discussed, findings cannot be generalised across all unstable footwear models from the same company (Buchecker et al., 2012; Gardner et al., 2014) or all unstable shoes (Paper 3&4; Germano et al, 2012). Although, considering objective four of this research, unstable footwear can induce specific-postural response in wearers as a response to instability. Compared to other unstable footwear designs, anterior-posterior rocker-soled shoes have the most exaggerated response in terms of centre of pressure excursion and muscle activation, particularly in the sagittal plane (Paper 3&4). This is as expected with the large unstable profile of the design in comparison to some of the more subtle design features in alternative unstable shoes. To deliver the most effective intervention, fully characterising instability from these shoes on relevant wearers in appropriate situations is essential. In future research this may be most successful through consideration of individual-subject differences (Branthwaite et al., 2013; Nigg et al., 2012).

4.3.3 "Gait Modifications"

It has long been anecdotally reported that flip-flop or toe-post footwear is detrimental to a "normal" gait style and results in substantial modifications in swing and stance, which are generally deemed to be detrimental (pg. 39-43). Despite this wide-belief, the use of open footwear is prevalent, particularly in warmer climates and in the summer months in the UK. Therefore, a footwear version which maintains key design features, but reduces gait modifications would be beneficial to wearers. The literature review conducted at the conception of these studies identified limited published research considering toe-post footwear, despite an array of anecdotal testimonies. In the interim, three international research groups have been adding to this body of work; Zhang et al., Chard et al., and Shroyer et al..

The aim of reducing gait modifications while walking in toe-post footwear was tested by holistically comparing gait in a standard flip-flop to FitFlop^{TM} footwear. This was conducted across two papers, one of which focused on plantar pressures acting on the foot sole and the second which compared ankle motion and lower limb muscle activation in the two footwear styles and to barefoot. The second paper thus provided FitFlop ltd with comparison data to walking in a standard flip-flop, in accordance with the primary objective of this research. When comparing the data between the flip-flop and the FitFlop^{TM} alternative the papers included above demonstrated that (Table 4.10):

- The magnitude of loading at touchdown can be reduced with a thicker and softer footbed: "lack of protection of sole"
- The eversion in mid-stance can be reduced with a profiled footbed and thicker strap that fits across the midfoot: "control frontal plane motion"
- Gripping in swing can be reduced in both magnitude and duration: "gripping".

These findings contribute to the information this research provides concerning the biomechanics of wearers in 'health and well-being' footwear. It potentially demonstrates that ‘health and well-being flip-flops could less detrimental to their lower limb health than a standard flip-flop; however long-term assessments and longitudinal studies are required.

Table 4.10 "Gait Modifications" paper findings: FitFlop compared to flip-flop.

Variables	Paper 5 (Pressure)	Paper 6 (Gait)
Reduced magnitude of loading	3.6% (p = .010) lower peak pressure in the heel.	19% reduction in maximum instantaneous loading rate ($p < .001$).
Reduced frontal plane motion	19.9% (p = .001) increase in midfoot contact area.	0.9±1.7° (p = .008) reduction peak ankle eversion in stance.
Reduced gripping in swing	Shorter duration (.094±.210, .002±.092 s, p = .001) and lower magnitude (.36±.62, .05±.50 kPa, p =.020) of pressure.	Increased dorsiflexion throughout swing compared to flip-flop (e.g. maximum dorsiflexion swing FitFlop 8.5±3.4°, flip-flop 7.6±2.6°, p =.050).

The work of the aforementioned authors to quantify gait modifications while walking in toe-post footwear will be combined with the results and conclusions from the present work comparing toe-post walking to barefoot and a potential alternative to re-address the original detrimental aspects of toe-post footwear.

4.3.3.1 Lack of Protection of the Sole

In-shoe plantar pressure, ground reaction force loading or peak tibial acceleration data provide insights into the shock absorption or pressure alleviating properties of footwear midsoles. Zhang et al. (2013) compared the loading rate of the ground reaction force in barefoot, open-toed sandals, flip-flops and a running shoe. The loading rate to the first peak of the ground reaction force was higher in the barefoot (7.96±1.79 $BW{\cdot}s^{-1}$), sandal (7.22±1.54 $BW{\cdot}s^{-1}$) and flip-flop (7.52±2.61 $BW{\cdot}s^{-1}$) conditions compared to the running shoe (5.69±0.41 $BW{\cdot}s^{-1}$) and the barefoot compared to the sandal. The minimal difference between the flip-flop, sandal and barefoot conditions questions the proposal by Carl and Barrett (2008) that a flip-flop can reduce the loading rate compared to barefoot. They identified a lower peak pressure in the heel region of the flip-flop compared to when participants walked in a sock with the pressure insole, which is likely an ineffective measure. Zhang et al. (2013) results should, however, be interpreted with caution as the variable compared was the average loading rate to the peak ground reaction force, not the maximum instantaneous loading rate before the peak ground reaction force, which may have been more informative. Despite this inconsistency the authors made conclusions relating the data to the shock absorption properties of footwear and this reaffirms the point from the earlier literature review that the variables selected and calculated are not always the most relevant or valid to address study hypotheses. Papers 5&6 of the current work identified characteristics which point to the flip-flop protecting the body from loading at touch-down and that the FitFlop^{TM} alternative may enhance this. When considering the plantar surface and reducing pressure and increasing contact area, the peak pressures were 16.0% (p = .218) and 9.6% (p = .073) higher in the standard flip-flop in the hallux and first metatarsal-phalangeal joint respectively. Furthermore, contact area increased by 7.3% in the FitFlop^{TM} condition (Paper 5, pg. 126). These variables highlight a higher contact area and lower localised pressures, which have been related to increased comfort in footwear in asymptomatic patients in addition to reduced risk in symptomatic patients (Cavanagh et al., 1992; Che et al., 1994). This relationship with comfort was later explored with the same footbed in Paper 7. Unfortunately the plantar pressure data is unable to be compared to the Carl and Barrett (2008) paper as they did not present the values of their data. The peak pressures from the current research appear slightly low compared to other research studies utilising similar walking velocities (Perry et al., 1995; Yung-Hui and Wei-Hsien, 2005), which may be explained by the use of the Medilogic in shoe pressure system, which was previously reviewed in this discussion (pg. 199-203) and is further explored in a recent publication (Price et al., 2014). Any future work undertaken using this system will utilise a calibration factor as previously recommended with a resistive insole system (Mueller and Strube, 1996).

In addition to redistributing plantar pressures, the FitFlop^{TM} (21.7±5.4 $BW \cdot s^{-1}$) condition resulted in a 19% (p < .001) reduction in loading rate of the ground reaction force compared to flip-flop (26.7±5.6 $BW \cdot s^{-1}$), although both provided significant reductions compared to barefoot walking (41.4±22.9 $BW \cdot s^{-1}$). This was consistent with findings from Paper 1, where the heel-strike transient occurred in 2.5% less trials and the average magnitude of the feature was lower when walking in a FitFlop™ (299.7±126.1 N) compared to a flip-flop (370.6±122.2 N) (unpublished data from Paper 1 data collection). The large magnitude difference between these loading rate values and those reported by Zhang et al. (2013) (5.69-7.96 BW·s^{-1}) in similar conditions conveys the value in calculating a maximum instantaneous loading rate when considering shock absorption characteristics of footwear. These studies, as expected, highlight that toe-post footwear provides some shock absorbing material beneath the heel and, as a result, reduces the loading applied to the body at heel-strike. It is apparent that modifications to the design of flip-flops in the form of thicker footbeds and modified EVA densities can further reduce this loading and reduce pressures acting on the plantar surface of the foot. Despite these alterations being potentially beneficial to wearers of 'health and well-being' footwear, they do not reduce the risk of puncture wounds and toe-stubbing as previously raised and highlighted due to the open nature of the footwear (American Podiatric Medical Association, 2012).

4.3.3.2 Does Not Control Frontal Plane Foot Motion

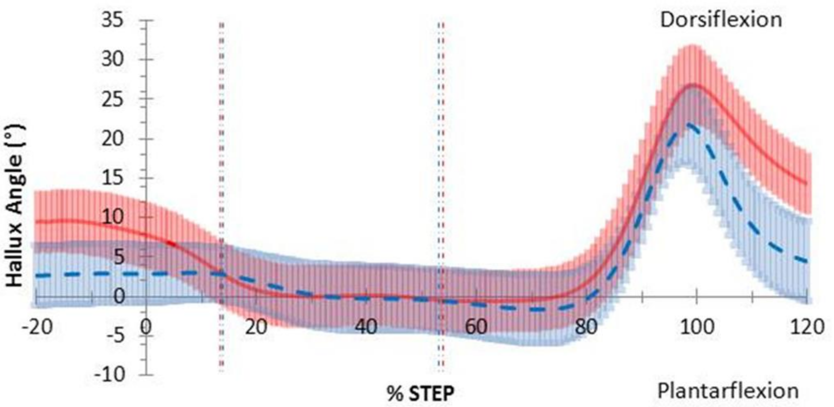
Specific studies on flip-flops are more limited than general barefoot/shod comparisons; however research on both children and adults now exits in this footwear. Chard et al. (2011) identified a trend towards a more dorsiflexed, everted and abducted midfoot in walking; however no significant differences in comparison to barefoot walking were identified. In further work, thirteen children (aged 8-13 years) were tested in total with an increased ankle dorsiflexion during initial-contact and increased midfoot plantarflexion during late-stance when walking. No significant differences were identified in eversion during midstance or range of motion in the frontal plane during midstance in walking with the multi-segment foot model (Chard et al., 2013). Frontal plane motion at the ankle decreased by 0.5º in the flip-flop and increased by 0.4º in the midfoot. Transverse plane motion also did not differ significantly at the ankle or midfoot in the children tested during walking (Chard et al, 2013). Contrasting this, Shroyer (2009), in his earlier work, identified reduced eversion during midstance and peak eversion angle in a flip-flop (-3.4±3.1°, -5.8±4.3°, both p < .001) compared to barefoot (-4.0±3.5°, -7.1±4.5°) in adults. Shroyer (2009) attributed this decrease in eversion in flip-flops to the y-strap of the flip-flop potentially limiting eversion of the hindfoot due to it running between the hallux and second phalangeal to both the medial and lateral side of the foot. Another mechanism proposed was that the reduced eversion is a result of the participants adapting their centre of pressure trajectory to reduce medial shifts of the foot due to the foot not being secured in the shoe with an upper. These multi-segment studies on flip-flop gait demonstrate conflicting results, potentially due to the contrast in adult and children's gait interaction with footwear, segment definitions, or variables chosen to quantify eversion. Shroyer and Weimar (2010) utilised eversion at midstance and the peak eversion value. In comparison Chard and Smith (2011) utilised a potentially more sensitive and holistic approach and compared the mean values across time in stance for four phases. (Shroyer, 2009) identified that eversion in four modified flip-flops, one with a medial arch support, mirrored that evident in the barefoot trials; attributing these findings to the population being defined as "normal arch" and the aforementioned lateral shift of the centre of pressure due to the foot not being fixed in the footwear. Despite reporting significant differences (between flip-flop and barefoot), the absolute difference in eversion experienced ranged by less than a degree between conditions in Shroyer's work (Shroyer, 2009), similar to Paper 6 (mean difference 0.9°, p = .008). The magnitude of difference is similar to that recorded by Chard et al. (2013) during midstance in walking between barefoot (3.6±1.8) and flip-flops (2.2±5.0, p = .231) and it may be that the lower subject number (N = 13) in this study meant that it lacked statistical power. Otherwise this study may have reported a similar significant difference, however a post-hoc power calculation identified a subject number of 12 was sufficient to achieve a significant difference with an alpha level at 0.05, power set at 0.8 and the effect size 0.62 (Chard et al., 2013). Furthermore, there is a large range in individual participant responses denoted by a standard deviation of 5° (Chard et al., 2013). As previously discussed, and noted in Paper 6, the meaningfulness of a reduced range of frontal plane motion of approximately 1° is yet to be established.

In contrast to using a multi-segment foot model, Zhang et al. (2013) reported data from a single-segment foot, which is consistent with Paper 6. Despite not being included in the papers within this thesis, the current work required the comparison of covered footwear in addition to the flip-flop, FitFlop™ and barefoot conditions, similar to the conditions compared by Zhang et al. (2013), and therefore employed a single-segment foot model. Zhang et al (2013) identified no significant differences between barefoot (-4.9±1.5°), flip-flop (-5.4±2.3°) or trainers (-6.5±3.1°) for ankle eversion range of motion, similar to Paper 6, which found significant differences in peak eversion in stance only between the FitFlop™ (-3.5±2.2°) condition and flip-flop (-4.4±1.9°) and barefoot (-4.3±2.1°), not between the latter. Differences in eversion may not be expected in a flip-flop condition compared to barefoot as the shoe is flat, has a flexible sole and no real upper, foot motion remains unchanged during stance (Wolf et al., 2008). The FitFlop™ data highlights, however, that the inclusion of a profiled footbed and more substantial upper can limit this range of motion. Consistent with this finding is a median increase of 20% of mid-foot contact area in FitFlop™ compared to flip-flop (Paper 5, pg. 126), again suggesting that the footbed is contacting the medial arch of the foot. This "support" for the medial arch in FitFlop™ thus removes one of the main criticisms of toe-post footwear, while maintaining the open style which gives the footwear its inherent functionality. This may be potentially beneficial to wearers as excessive frontal plane motion of the ankle has been repeatedly linked to overuse injuries (McClay, 2000; Mündermann et al., 2003; Willems et al., 2006). Nevertheless, prospective studies should quantify whether this benefit is apparent through interventions and whether this magnitude constitutes 'excessive' frontal plane motion.

4.3.3.3 Requires Gripping in Swing

A further criticism of the footwear style is that the footwear must be held on during gait as it has insufficient upper or straps to remain secure on the foot. Authors have explored this relationship with 3D motion analysis in stance to test the hypothesis that for the shoes to stay on in stance the wearer grips with the toes. Chard et al. (2013) identified no significant differences in mean hallux angle in barefoot compared to flip-flop footwear throughout stance, however hallux sagittal plane position was less dorsiflexed at -10% of stance (6.5º, p = .005), at heel-strike (4.9º, p = .031) and at 110% of stance (10.7º, p = .001) in children walking. A similar range of hallux flexion has been identified in adults and children when walking barefoot (Wolf et al., 2008). Contrasting this, Wolf et al. (2008) compared hallux motion in covered footwear, identifying increased dorsiflexion in shod walking, with a reduced range of motion of 11.4º (p < .001). Bojsen-Møller and Lamoreux (2009) also identified reduced hallux dorsiflexion at heel contact and at push off when walking in both flexible and stiff footwear compared to barefoot. This is consistent with other studies comparing barefoot and covered shod walking (Bishop et al., 2011). It is likely that these reductions in hallux sagittal plane motion in covered footwear are due to the upper constraining the motion of the toe. Bishop et al. (2011) compared a modified shoe with a cutaway for a hallux marker to the intact shoe and identified a 9.4º increase in hallux range of motion. It is therefore evident that hallux motion in a flip-flop may be more representative of barefoot kinematics than a covered shoe. However, evident within Figure 5 in Chard et al. (2013), some plantarflexion following heel-rise in the toe-post footwear was recorded (Figure

4.14). Additionally, dorsiflexion reduced at toe-off when compared to barefoot, although the mean values across four stance phases reported in the paper report no significant differences. This may be indicative of the “gripping” proposed to control relative foot and shoe motion when walking in toe-post footwear with no secured back (Carl and Barrett, 2008; Shroyer and Weimar, 2010). The reporting of hallux motion over the gait cycle may have provided better insight as to whether the hallux “grips” the footbed in swing as opposed to stance and should be investigated in future research. The enclosed paper (Paper 5, pg. 129) provides the only known data quantifying gripping with the hallux utilising in-shoe plantar pressure data. This mechanism was evident in swing (and was of a lesser magnitude and shorted duration in the FitFlop™), however, as there was no closed shoe or barefoot comparison, it cannot be confirmed that this is a feature of walking in toe-post footwear alone. With the work undertaken so far in this field differences in hallux motion in barefoot, toe-post footwear and covered footwear cannot be attributed to covered uppers alone restricting hallux motion as opposed to the footwear having a back-strap and therefore needing less control from the hallux to grip. It is likely that reported differences are an interaction of these two factors and further research is required to isolate features and potentially adapt footwear to remove or enhance this motion. Further electromyographical examination of the toe flexors would assist in this investigation.



Sagittal plane hallux motion in walking gait. Mean joint angles for barefoot (red) including 95% confidence intervals (red shading) and thong (blue dashed) including 95% confidence intervals (blue shading), including 20% before and 20% after stance (y-axis), while walking. Events foot-flat and heel-rise represented by the vertical red (barefoot) and blue (thong) dashed lines.
Chard *et al. Journal of Foot and Ankle Research* 2013 **6**:8 doi:10.1186/1757-1146-6-8

Figure 4.34 Sagittal plane hallux motion (Figure 5 from Chard et al., 2013, with permission).

Despite minimal differences identified in frontal and transverse plane motion of the foot in flip-flops, examination of the sagittal plane motion has identified significant differences between conditions (Chard et al., 2013; Shroyer, 2009). In adults walking, the flip-flop reduced dorsiflexion in swing (10.5±4.9°) compared to barefoot (11.6±4.9°, $p < .001$) (Shroyer, 2009). In the current research contrasting findings were identified with an increase in maximum dorsiflexion in swing in both flip-flop (7.6±2.6°) and FitFlop™ (8.5±3.4°) compared to barefoot (6.7±2.6°) (Paper 5, Figure 3.14, pg. 143). Shroyer et al. (2009) identified increased plantarflexion in swing when participants wore flip-flops compared to trainers, attributing their finding to contraction of the toe flexors to hold the flip-flop, creating a plantar-flexor moment at the ankle. As evident in Paper 5 (Figure 3.12, pg. 129), FitFlop™ would likely reduce this moment due to a decrease in duration and magnitude of pressure application under the hallux. Thus, if this hypothesised mechanism is correct, this could rationalise the contrasting findings in FitFlop™, where greater dorsiflexion is enabled due to a reduced plantarflexor moment. Additionally, the increased mass and thicker sole of the FitFlop™ footwear (Paper 5, Table 3.14, pg. 122) may illicit greater ground clearance than the flip-flop condition, thus a greater dorsiflexion in swing. The potential benefits of the wider straps fitting higher up the foot and maintaining the footwear contact with the heel, such that sagittal plane ankle angle would not require modification, needs to be examined in footwear with similar ground clearance characteristics (i.e. midsole thickness) such that decisions could be informed for design priority.

4.3.3.4 Summary

The included research provides valuable information regarding the nature of walking in toe-post footwear, and in particular provides the first presentation of plantar pressure data in both a flip-flop and FitFlop™. The Papers thus satisfy objectives one and four of this research, providing data considering FitFlop™ footwear and, specifically investigating biomechanics concepts related to ‘health and well-being’ footwear. The literature for flip-flops would benefit from the addition of empirical studies which define the mechanisms for injuries and pathologies which have been attributed to the footwear style. The nature of the papers, highlighting aspects related to comfort (e.g. pressure under the first metatarsal head) and potential detrimental behaviours (e.g. loading rate at heel-strike), is particularly relevant as they provide specific variables relating to specific hypothesis as opposed to descriptive studies. These also relate to the actual biomechanical function of the footwear and provide quantitative assessment of the influence of the footwear on wearers gait biomechanics. The kinematic analysis in Paper 6 provides a starting point comparing 3D motion in walking in adults in flip-flops. The conclusions that can be drawn from the data, however, are limited by the lack of multi-segment foot model implementation. Multi-segment foot motion is the next step for toe-post research and has been undertaken by Chard et al. (2013) on children walking and jogging in flip-flops and should be progressed, particularly relevant are models that reliably and validly quantify:

- Foot spread: the medial arch length and forefoot width (Wolf, 2008)
- Hallux motion modifications due to confinement (Bishop et al., 2011) and to hold flip-flops in place (Carl and Barrett, 2008)
- Calcaneus versus shank (eversion, overpronation) (Shroyer et al., 2010)
- Flexion at the metatarsal-phalangeal joint at toe-off (Thewlis et al., n.d.)

Further work along the same research themes should implement this.

4.3.4 "Comfort"

The primary aim of this research study was as a product comparison for the research contract with FitFlop ltd, relating to objective one of this research. This was as a benchmarking process for internal product development purposes which aimed to compare two of the new FitFlop^{TM} products to best-selling alternatives within similar footwear ranges. The product comparison aimed to quantify comfort within the footwear while being worn by relevant wearers who were more representative of wearers of the product than may have been utilised in previous research. A protocol involving subjective and objective measures was implemented on three FitFlop^{TM} products and relevant comparisons in two separate data collection protocols, one of which was written for publication (Paper 7). This protocol was drawn from running literature to provide data relating to walking footwear and thus contribute to the second and third objectives within this work.

4.3.4.1 Emphasis

Comfort is an essential feature of 'health and well-being' footwear and therefore this assessment was a relevant addition to the research field relating to the influence of this footwear category on the biomechanics of wearers and objective four of this research. Moreover, this paper extended and combined aspects of the other work within the publications within this thesis to provide variables to quantify comfort (tibial acceleration, plantar pressures and a comfort questionnaire) and define characteristics of the footwear (e.g. peak acceleration from a modified drop-test). The work combined subjective and objective comfort measures, which was consistent with approaches evident in empirical literature (Arezes et al., 2013; Che et al., 1994; Dinato et al., n.d.; Yung-Hui and Wei-Hsien, 2005).

The use of a modified questionnaire to better suit the characteristics of the population and footwear being tested, however, was an advancement compared to most protocols, which implement the standard (Mündermann et al., 2002) questionnaire to assess all footwear styles (Dinato et al., n.d.; Hong et al., 2005). The manipulation of the questionnaire data from the consistently used scale in literature was a novel-aspect of the work undertaken as part of this research. Previous authors have repeatedly reported the use of the Mündermann et al. (2002) scale for situations and methods for which it has not been designed or validated (Table 2.3, pg. 48-49). The manipulation of specific aspects of the scale to better suit the population and footwear being assessed was essential to obtain useful and valid information pertaining to comfort from test sessions. However, future use and development of comfort scales should look to undertake further, and more extensive, validation of a modified scales reliability and constructs.

4.3.4.2 Comfort Scale

Since the conception of the research (June 2012) a wide range of comfort research has been undertaken and published considering walking footwear, which has followed the previously discussed style with implementation of VAS (pg. 45-51) to quantify subjective comfort alongside objective measures (Ceccaldi and Janin, 2014; Dinato et al., n.d.; Zhang and Li, 2014). As with Paper 7, the development of a new questionnaire is outside the scope of footwear biomechanics research in-terms of time, expertise and resources available. Thus the current study altered the comfort scale developed for running by Mündermann et al. (2002) to be more specific to walking, consistent with the third aim of this body of work. As protocols incorporate existing scales they concurrently integrate the inherent weaknesses associated with them. The reported perceptions of comfort are due to multi-factorial aspects relating to both the footwear (stiffness, compression, energy dissipation, deceleration, style, temperature, volume etc.) and the wearer (sensitivity, foot size, expectations, aesthetics, familiarity, assumptions, activity etc.), which then interact with the scale (instructions, length, clarity, comprehension, format, definition/anchor words etc.). Evidently, if the scale does not match what the wearer perceives or internally defines as comfort then this perception or sensation will be misinterpreted. For example, “not comfortable at all” may be quite ambiguous as an anchor phrase and may mean ‘zero comfort’ or ‘maximum discomfort’ depending on the theory of comfort employed (De Looze et al., 2003; Helander and Zhang, 1997; Vink and Hallbeck, 2012). No participants in the current study questioned this aspect of the scale design, however, and the repeated measures nature of the footwear comparison means that any influence of misinterpretation or misrepresentation was likely consistent across the footwear styles compared. This ambiguity however would influence the relationship between objective and subjective variables; hence the 'difference' scores were compared between conditions as opposed to absolute scores from the objective and subjective sources (Paper 7, pg. 156). Knowledge evident in the ergonomics and product performance fields could increase the validity of concepts in comfort questionnaires utilised for footwear assessment in footwear biomechanics literature, for example consideration of the multifactorial model presented by (Vink and Hallbeck, 2012). This would also be beneficial for consideration within footwear companies to ensure that product creation and development aligns with wearers subjective requirements (Sterzing et al., 2012)

The participant response to the questionnaire within the study produced varying outcomes. Responses to the scales relating to "feeling under the foot arch" and "comfort around the toe joints" resulted in a large range in response from participants (Paper 7, Table 3.20, pg. 158). These were unfortunately the novel scales relating to the footwear specific in this study that were altered from the Mündermann et al. (2002) study. This could be indicative of the participants not understanding the concepts that were being assessed or that the sensation did differ widely between participants. The "feeling under the foot arch" was stable in the repeatability study and thus the latter is a reasonable explanation (pg. 197-202). Higher plantar pressure below the medial arch have been identified in more comfortable footwear due to it representing an increased contact area (pg. 53-54) and this aspect has been identified as an important factor in overall footwear comfort considering arch comfort, cushioning and height (Mills et al., 2010). Consequently a valid subjective measure of this sensation may be useful in the future to discriminate between more and less comfortable footwear. Participant variability in comfort questionnaires is high (Mills et al., 2011; Mündermann et al., 2002, 2001; Worobets et al., 2009). Thus, data treatment may be a strategy to manage this occurrence as sensations and perceptions will always vary. Presenting data as an average difference from a control shoe value may reduce standard deviations, thus increasing the power of comparisons (Mündermann et al., 2002) or reporting data normalised to participants own maximum and minimum ratings (Jordan et al., 1997; Witana et al., 2009). Additionally, as identified, multiple visits and ratings may increase the validity of this data (Mündermann et al., 2002).

4.3.4.3 Interpretation

Due to the inconsistent and seemingly unreliable nature of footwear comfort, the approach taken by many researchers is to combine subjective and objective measures to quantify footwear comfort or perceptions of cushioning (Dinato et al., n.d.; Goonetilleke, 1999; Sterzing et al., 2013). The relationships between the regional objective and subjective variables measured in the current study were weak (r = .081 - .341; not all reported in Paper 7). The weakest regional comparison was evident between the "cushioning in the heel" and the peak pressure in the heel (r = .081, p = .635). The strongest correlation was evident also with the same subjective variable and maximum regional heel pressure, which were significantly correlated (r = .341, p = .039, Paper 7, Table 3.22, pg. 161). Other correlations within the study ranged between r = .109-.266 for the regional pressure variables and the related subjective values (p = .137-.423). Contrasting the weaker relationships identified in previous research (pg. 53-55) and discussed. Zhang et al. (2014) identified stronger and significant relationships between "overall comfort" when correlated with plantar pressure variables. "Overall comfort" may be the most important measure in-terms of representing buying and wearing behaviour and therefore may be the seemingly most valuable measure to a footwear company. Plantar pressure under the midfoot resulted in the strongest correlation (r = 0.495, p < .01). However, this study did not explore the relationship between the regional perceptions of comfort and plantar pressure data as in Paper 7. Contrasting plantar pressure values, and consistent with the work previously explored (pg. 54), the relationship between mechanical testing and perceptions of cushioning continues to be reportedly strong (Sterzing et al., 2013) even when using Likert scale data (Worobets et al., 2009). This is not true, however, for footwear with different constructions such as cushioning technologies, although the study identified limited differences in biomechanical variables and perceptions between the shoes (Dinato et al., n.d.). In addition to minimal quantifiable differences in-terms of biomechanical response of the footwear conditions, the concepts quantified by questionnaires may not directly correspond to the objective measure. It may be that the measures quantified in the objective protocols in studies (e.g. peak pressure, pressure-time integral, contact area and peak acceleration) conceptualise discomfort as opposed to comfort (Goonetilleke, 2001), contrasting the comfort concepts quantified by the anchor words on (Mündermann et al., 2002) scale. This contradiction could explain the discrepancies from these correlation studies.

4.3.4.4 Limitations

Despite the demonstration of a combined protocol to assess footwear comfort, the nature of the protocol and research question reduce the information that can be gained from Paper 7. More specifically, the lack of anonymity of the footwear, the lack of isolation of single variables being modified between conditions and the lack of full definition of participant's foot characteristics reduce the information that can be inferred from the current comparison (Paper 7, pg. 166). The anonymity of companies is essential for future comparisons and further exploration of the concept of comfort should look to blind participants to the footwear conditions they are wearing, such as the method by (Dinato et al., n.d.), covering the footwear with black tape. As Au and Goonetileeke (2007) highlighted, 18.4% of the variance in comfort scoring can be attributable to footwear aesthetics. This methodology would be effective providing the inherent structure, thermal properties or flexibility of the footwear is not altered by this addition. The manipulation of single-footwear aspects is essential to be able to make directed conclusions to inform the footwear research and development or prescription process. For example, Lane et al. (2014) altered only the Shore A hardness of the footwear sole in their comparison of subjective comfort and plantar pressure in people with forefoot pain. Despite the research identifying no significant differences in terms of subjective comfort between the three hardness conditions included (25, 40 and 56 Shore A) the study design isolated a single variable and then considered the change in perception of patients. As discussed previously, the inclusion of a control shoe between conditions to provide a stable baseline may have assisted subjects to differentiate their perception of the footwear conditions. Otherwise, it may be that the older adults included in the research (65 years and over) demonstrated reduced foot sensitivity (Kenshalo, 1986), and despite the significant changes in plantar pressures with the sole hardness sensations were consistent. In addition to the footwear characteristics, participant foot dimensions and characteristics can influence both plantar pressure and perceptions of comfort (Cheng and Hong, 2010; Chuckpaiwong et al., 2008; Morag and Cavanagh, 1999) and thus should be more extensively defined in further work.

4.3.4.5 Summary

The footwear conditions tested in Paper 7 act as product testing to satisfy the primary objective of providing data for FitFlop™, as opposed to research studies, which limits their generalizability and conclusions. However, the assessment of comfort is essential for 'health and well-being footwear' and the protocol presented could be integrated in this footwear category to assess this important characteristic. Full definition of participants and strict study designs would enable valuable information to be gained from footwear comfort research. Integration of applied ergonomics findings into comfort assessment could advance the validity of scales to increase the sensitivity of comfort/discomfort measurement, which is particularly important for the quantification of variables on symptomatic populations. The integration of comfort variables into the footwear development process, for example to define regional requirements for pressure reduction on the plantar surface (Wenyan and Goonetilleke, 2009) or sole hardness (Dinato et al., n.d.), is essential to further the commercially available footwear concurrently with footwear biomechanics research.

4.3.5 Footwear biomechanics concepts relating to 'Health and Well-being' Footwear

From the study results and discussions above it is evident that the 'health and well-being' footwear tested within this range of studies influences the biomechanics of the wearer. There are quantifiable differences in the shock absorption properties of the footwear (Paper 1) and centre of pressure and muscle activation markers for stability of the wearer in standing and walking compared with a stable control shoe (Papers 3&4). Moreover biomechanical modifications of the gait of wearers is evident in varying degrees in 'health and well-being' footwear (Papers 5&6) and comfort variables highlight a potential to increase perceptrions of comfort in the wearer (Paper 7). Specific study findings are addressed with respect to the original objectives of the thesis in Figures 4.15 4.16 and 4.17.

OBJECTIVE ONE. To provide data on the influence of FitFlop footwear on walking and standing.

"Shock Absorption"

The heel velocity toward the ground in FitFlop (0.36±0.05 m·s^{-1}) is twice as fast as in barefoot walking (0.19±0.05 m·s^{-1}, $p < .05$) and consistent with a flip-flop (0.36±0.05 m·s^{-1}, $p > .05$) (Papers 1&6).

The occurrence of the heel strike transient in the FitFlop was less often in FitFlop (22.1%) compared to barefoot (98.5%) and flip-flop (24.6%) (Unpublished data Paper 1).

The magnitude of the heel strike transient was lower in FitFop (299.7±126.1 N) compared to barefoot (379.5±119.7 N) and flip-flop (370.6±122.2 N) and heel strike transient occurred later in FitFlop (27.3±0.0 ms) compared to barefoot (12.5±3.1 ms) and flip-flop (25.6±2.0 ms) (Paper 1).

Peak axial tibial acceleration in FitFlop (21.8±8.5 m·s^{-2}) is significantly lower than in barefoot walking (40.8±16.1 m·s^{-2} $p < .05$) and consistent with in a flip-flop (22.0±9.8, $p > .05$) (Paper 1).

Reducing the depth of the FitFlop footbed from 41 to 35 to 28 mm does not significantly or consistently alter variables related to shock absorption in walking or in a mechanical drop-test protocol (Paper 2).

Reducing the hardness of the heel section of the FitFlop footbed decreases the magnitude of variables quantifying shock (e.g. maximum instantaneous loading rate of the vertical ground reaction force from 55 Shore A: 36.7±8.9 kN·s^{-1} to 30 Shore A: 21.1±7.2 kN·s^{-1}) and increases the duration of the impact phase (e.g. time to peak positive axial tibial acceleration 55 Shore A: 19.4±5.3 ms to 30 Shore A: 26.7±9.2 ms) (Paper 2).

FitFlop (21.7±5.4 BW·s^{-1}) significantly reduced the maximum instantaneous loading rate of the vertical ground reaction force compared to barefoot (41.4±22.9 BW·s^{-1}, $p < .001$) at a self-selected walking speed (Paper 6).

Using a modified drop-test to replicate heel contact in walking, the peak tibial acceleration value was lower in FitFlop SuperT (2.2 *g*) than a market leading item of footwear (3.3 *g*) (Paper 7).

"Instability"

In single-leg standing lower limb kinematics and centre of pressure deviations did not differ in FitFlop from the control sandal (Paper 3).

In single-leg standing the RMS muscle activation was significantly higher in FitFlop (+27 (CI: 0-56) %, $p = .046$) than the control sandal in medial gastrocnemius, but no other significant differences existed (Paper 3)

When walking in FitFlop (19.7±2.6%) double-support time was a significantly smaller proportion of the gait cycle than in the control sandal (20.7±3.1%, $p = .046$). No further variables relating to temporal-spatial, kinematic or centre of pressure characteristics differed (Paper 4).

An increase in stance time was recorded in the FitFlop (.540±.050 s) compared to flip-flop (.535±.050 s, $p = .004$) (Paper 5).

During mid-stance in FitFlop muscle activation of biceps femoris (-24 (CI: -58- -4) %, $p = .025$), medial gastrocnemius (-33 (CI: -44.1- -18.3) %, $p = .005$) and soleus (-24 (CI: -36- -15) %, $p = .005$) was lower than in the control sandal. In the peroneus longus at pre-swing muscle activation recorded in FitFlop exceeded that in the control sandal (46 (CI: 4-79) %, $p = .025$) (Paper 4).

Figure 4.15 Specific findings related to objective one

“Gait Modifications”

Gripping with the hallux in swing in FitFlop is of a lower magnitude and for a shorter duration (.05±.50 kPa for .002±.092 s) than in a flip-flop (.36±.62 kPa, p = .001, .094±.210 s, p = .020) (Paper 5).

Total contact area of the plantar foot is higher in FitFlop (87.9±10.0%) than a flip-flop (80.6±8.3%, p > .001), largely attributable to a median 19.9% (p = .001) increase in the midfoot in this condition compared to the flip-flop (Paper 5).

Peak pressures under the hallux and first metatarsal-phalangeal joint did not differ between FitFlop and flip-flop. Pressure under the heel was significantly lower in FitFlop (170.3±120.9) than flip-flop (176.6±135.0, p = .010) (Paper 5).

Centre of pressure mean position was more lateral in FitFlop (p > .001) and the maximum (22.0%) and mean (6.6%) velocity toward the toes was slower in FitFlop than flip-flop (p = .002) (Paper 5).

Walking in FitFlop footwear increased maximum dorsiflexion angle in swing (+0.9°, p = .05) and tibialis anterior activation in terminal swing (32.6%, p < .001) compared to barefoot (Paper 6).

Maximum eversion was significantly lower when walking in FitFlop (-3.5±2.2°) compared to barefoot (-4.3±2.1°, p = .032). Peroneus longus activation did not differ significantly in stance (Paper 6).

FitFlop (21.7±5.4 BW·s^{-1}) significantly reduced the maximum instantaneous loading rate of the vertical ground reaction force compared to barefoot (41.4±22.9 BW·s^{-1}, p < .001) at a self-selected walking speed (Paper 6).

“Comfort”

The FitFlop SuperT trainer (78 comfort points) was rated as equally as comfortable as a market leading item of footwear (76 comfort points, p =.814). The cushioning in the heel (+19 comfort points, p = .001) and comfort under the ball of the foot (+12 comfort points, p = .016) were rated as significantly more comfortable (Paper 7).

Regional peak pressures in the medial (-131.6 kPa, p = .008), central (-479.9 kPa, p < .001) and lateral (-356.5 kPa, p < .001) metatarsal heads were significantly lower in the FitFlop SuperT condition than a market leading item of footwear (Paper 7).

Peak pressures in the medial (-197.3 kPa, p < .001), central (-205.8 kPa, p < .001) and lateral (-107.2 kPa, p < .001) metatarsal heads were significantly lower in the FitFlop SuperT condition than a market leading item of footwear (Paper 7).

Peak regional pressures in the medial (+258.7 kPa, p < .001) and lateral (+99.4 kPa, p < .001) midfoot were significantly higher in the FitFlop SuperT condition than a market leading item of footwear (Paper 7).

Peak pressures in the heel were significantly reduced in FitFlop SuperT (398.0 (CI: 350.9-472.7) kPa) condition than a market leading item of footwear (509.1 (CI: 438.5-623.4) kPa, p < .001) with a medium effect size (Paper 7).

Peak tibial acceleration did not differ in FitFlop compared to the market leading item of footwear (p = .946). However, peak acceleration in a modified drop-test was lower in FitFlop (2.2 v 3.3 g) (Paper 7).

Figure 4.16 Specific findings related to objective 1

OBJECTIVE TWO. To demonstrate the importance of biomechanical testing in walking footwear.

OBJECTIVE THREE. To modify testing from running footwear protocols for walking footwear.

“ShockAbsorption”

The currently used ASTM protocol over estimates peak acceleration (80.7-302 $m{\cdot}s^{-2}$, $p < .001$) and peak force (416.6-2411.9 $m{\cdot}s^{-2}$, $p < .001$) compared to walking values in a range of footwear. Values utilising the adapted protocol reduced the error in the estimation of these values in both acceleration (0.6-10.5 $m{\cdot}s^{-2}$, $p < .001$) and force (57.6-151.2 $m{\cdot}s^{-2}$, $p < .786$) data (Paper 1).

In modified footbed hardness conditions, the mechanical protocol overestimated peak axial tibial acceleration by 9.4 $m{\cdot}s^{-2}$ ($p < .001$) and underestimated the maximum instantaneous loading rate of the ground reaction force by 9.1 $kN{\cdot}s^{-2}$ ($p < .001$). Correlation between the mechanical and human maximum instantaneous loading rate neared significance ($r = .916$, $p = .084$) (Paper 2).

“Comfort”

The modified comfort questionnaire was repeatable for participants with intra-class correlation coefficients of 0.778-0.967 between-session for the different scales making up the questionnaire. A minimal important difference to denote change in comfort was estimated at 10.1±3.3 (range 5.2-14.1). Both of these aspects are consistent with the reported repeatability of the regularly used running comfort scale (Paper 7).

OBJECTIVE FOUR. To measure footwear biomechanics concepts relating to ‘health and well- being’ footwear (in addition to those listed above)

“Instability”

Different styles of ‘health and well-being’ footwear induce instability in the wearer at different phases of the gait cycle.

The influence of Masai Barefoot Technology^{TM} footwear is pronounced in the sagittal plane, for example decreasing median tibialis anterior RMS (-46%, $p = .005$) activation and increasing gastrocnemius RMS (69%, $p = .010$) activation at loading response compared to the control sandal.

The influence of the other outsole styles was more subtle and not in sagittal plane motion. However FitFlop^{TM} (46%, $p = .025$), Reebok EasyTone^{TM} (33%, $p = .025$) and Skechers Shape-Ups^{TM} (51%, $p = .010$) increased peroneus longus activation at pre-swing compared to the control sandal.

“Gait Modifications”

Walking in FitFlop footwear increased dorsiflexion throughout swing compared to flip-flop (e.g. maximum dorsiflexion in swing FitFlop 8.5±3.4°, flip-flop 7.6±2.6°) (Paper 6).

Maximum eversion was significantly lower in FitFlop (-3.5±2.2°) compared to walking in the flip-flop (-4.4±1.9°, $p = .008$). Peroneus longus activation did not differ significantly in stance (Paper 6).

FitFlop condition reduced the maximum instantaneous loading rate of the vertical ground reaction force at heel contact by 19% compared to flip-flop ($p < .001$) (Paper 6).

Figure 4.47 Specific findings related to objectives two, three and four

4.4 Dissemination and Wider Impact

The dissemination of the work presented within this thesis has been broad; spanning from scientific marketing on the funding company website to podium presentations at academic conferences. Examples include scientific reports, technical demonstrations at media events and presentations within the company.

4.4.1 Conference Presentations and Posters

In addition to this research being published (or in review) as seven papers in the Gait & Posture, Journal of Foot and Ankle Research and Footwear Science journals, other scientific dissemination activities have been undertaken. The work was utilised to prepare two conference posters and a podium presentation at the I-FAB conferences from 2010-2014 (Appendix C):

1. Presentation: The impact of a health Flip Flop on asymptomatic gait
   I-FAB Congress, University of Washington, Seattle, United States, September 2010.
2. Poster: Single-leg balance in "instability" footwear
   I-FAB Congress, University of Sydney, Sydney, Australia, April 2012.
3. Presentation: Testing a mechanical protocol to replicate impact in walking footwear
   I-FAB Congress, Busan, Korea, April 2014.

This enabled the work to be disseminated internationally directly to approximately 600 delegates from clinics, footwear companies and academic researchers from biomechanics-, footwear- and podiatry-focused research groups.

4.4.2 Reports, Presentations, Marketing and Internal Documents

The work encompassed in this thesis was utilised for over 40 internal reports within both the Knowledge Transfer Partnership (April 2009-April 2011) and the following research projects

(May 2011-May 2012, June 2012-June 2013) (Appendix D). These reports were integrated into the footwear technology and marketing departments of the funding company. They also enhanced company knowledge and served as educational sources for staff inductions. Other work relating to the biomechanical concepts addressed in this research was utilised for promotional events and materials for the sponsoring company such as research sections on their website, staff training ad press releases. The wider dissemination of this research therefore is relatively diverse and international during a time when the company's sales turnover incremented substantially and employees increased by 56. Furthermore, within-house the findings from this product-testing informed and assisted in the development and sale of new footwear styles and products by the company, enabling them to diversify their wearer, most notably from female only footwear to having children's and men's footwear.

4.4.3 Article Views and Citations

The wider-scientific dissemination of the work undertaken demonstrates how the work has expanded knowledge and added value to the existing body of walking footwear biomechanics literature. A search was conducted using web searches and hosting journal websites to quantify citations and article views (Table 4.11).

It is evident from the information in Table 4.11 that citation numbers for the papers included in the thesis are low. This can in-part be attributed to the short duration of time for which the articles have been in the public domain prior to submission of this thesis. The longest duration between paper publication and thesis submission is 18 months and some of the papers are yet to be published in the public domain. It is expected that citations will increase over the next few years as research is undertaken and published that finds the current work relevant and builds upon the inherent themes. The nature of Paper 1 proposes a methodology to alter footwear drop-testing methodologies. It is thought that this papers dissemination will be wider, influencing footwear companies and research institutes as well as academics. This form of dissemination and added value cannot realistically be easily quantified as obviously as peer-reviewed citations. Similarly, clinicians using extrapolations from the work, particularly using inferences from the unstable footwear or flip-flop papers, are likely and, unfortunately, largely unquantifiable.

Table 4.11 Publication details and dissemination.

Title	Journal	Year	Views	Citations
The impact of a health Flip Flop on asymptomatic gait	Foot and Ankle International	2011	U	1
Single-leg balance in “instability” footwear (Abstract)	Journal of Foot and Ankle Research	2012	847	2
Testing a mechanical protocol to replicate impact in walking footwear (Abstract)	Journal of Foot and Ankle Research	2014	162	0
1. A mechanical protocol to replicate impact in walking footwear	Gait and Posture	2014	280	0
3. The effect of unstable sandals on single-leg standing	Footwear Science	2013	50	1
4. The effect of unstable sandals on instability in gait in healthy female subjects	Gait and Posture	2013	U	1
5. A comparison of plantar pressures in a standard flip-flop and a FitFlop using bespoke pressure insoles	Footwear Science	2013	56	1

Numbers relate to published article views from journal websites on 27th August 2014. Where U denotes unavailable information.

4.5 Conclusions

These research papers combine to produce a collection of kinematic, kinetic, electromyography and plantar pressure data in ‘health and well-being’ footwear and specifically in FitFlop™ footwear. The work provides data on 128 individual participants wearing ‘health and well-being footwear and captures in-shoe plantar pressure, lower limb kinematics, ground reaction force, tibial acceleration, electromyography and subjective questionnaire data. The specific contributions and findings with respect to the overall thesis aims are depicted in Figures 4.14, 4.15 and 4.16. The dissemination has spanned demonstrations, conference presentations and peer-reviewed published papers in addition to informing the footwear technology department of FitFlop ltd for four years (2009-2013). This data was collected on relevant participants to replicate what may be a representative population of wearers of ‘health and well-being’ footwear, as opposed to simply utilising a convenient sample. The work demonstrates that commonly used methodologies from clinical and running footwear biomechanics could inform the walking footwear product development processes if effectively implemented. The appropriate adaptation of protocols (e.g. the mechanical drop-device and the comfort questionnaire) has demonstrated that a thorough consideration of aspects relating to walking footwear can produce useful data for manufacturers or companies producing or selling daily walking footwear as well as 'health and well-being' footwear for both research and development and marketing. Contrasting the results for running protocols to the bespoke walking protocols (as in Paper 1) demonstrate that simply applying these to walking footwear is not sufficient. This promotes the specific adaptation of protocols to replicate the demand placed on footwear in walking, particularly for different styles of walking footwear, to better estimate the influence of footwear style and design choices on wearer perceptions and outcomes.

Future research and development of the ideas within this thesis would further explore the mechanisms which result in gait modifications in footwear and altering this footwear style to try and reduce these mechanisms following a multi-segment analysis of walking in this footwear. The wearer-centred meaning of specific values of loading rate of the vertical ground reaction force and peak axial tibial acceleration in terms of injury or discomfort should be further clarified. Additional consideration of footwear testing protocols and their relevance to gait, footwear style and wearer may also result in specific recommendations for altered or additional testing of walking or 'health and well-being' footwear prior to sale.

The limitations of the work are generally due to the commercial nature of the research project that the studies were designed under. The subject numbers limit the effect size of some of the studies and this should be considered when interpreting results.

Chapter 5 Appendix

5.1 Appendix A: Co-author statement of work

Manchester Metropolitan University

I declare I am aware that the following manuscripts, for which I am a co-author, will form part of the PhD by Carina Price to be submitted at the University of Salford.

A mechanical protocol to replicate impact in walking footwear.

Price, C, Cooper, G & Graham-Smith, P & Jones, R (2014).

Gait & Posture, Feb 2014. doi.org/10.1016/j.gaitpost.2014.01.014

The manipulation of midsole properties to alter impact characteristics in walking footwear

Price, C & Cooper, G & Jones, R

Footwear Science: in review.

Testing a mechanical protocol to replicate impact in walking footwear

Journal of Foot and Ankle Research, 7 (S1), A68. April 2014. doi:10.1186/1757-1146-7-S1-A68

I hereby confirm that Carina Price contributed to a major portion of both the research and writing phases, was Corresponding Author for the manuscripts and presented the abstract. I confirm that I was involved in the development of the premise for this research and contributed to a proportion of the research, final writing and revision phases.

Signature:

Name: Dr Glen Cooper

Date: 14/05/14

University of Salford MANCHESTER

I declare I am aware that the following manuscripts, which I am a co-author, will form part of the PhD by Carina Price:

The effect of unstable sandals on instability in gait in healthy female subjects

Price, C & Smith, L & Graham-Smith, P & Jones, RK

Gait & Posture, 38(3),410-415.

The effect of unstable sandals on single-leg standing

Price, C & Smith, L & Graham-Smith, P & Jones, RK

Footwear Science, 5(3),147-154.

I hereby confirm that Carina Price contributed to a major portion of both the research and writing phases. Carina Price was also Corresponding Author for the manuscripts. I confirm that I contributed to a proportion of the research and final writing phases.

Signature: L C Smith

Name: Laura C Smith

Date: 29 April 2014

Carina Price was appointed as Project Assistant on a Knowledge Transfer Partnership between Brandhandling (Fitflop) and the University of Salford under the supervision of Professor Richard Jones and myself. As part of the KTP Carina performed a number of projects for the company addressing specific questions around the health benefits and design of their footwear. Carina's Phd research followed the same theme and consequently the 5 papers presented here for her Phd submission were designed and undertaken by herself with minor contributions from the co-authors. She was the main contributor to all aspects of the research process from the study design, data collection, analysis and write up. The five papers were approved for publication by Fitflop.

A mechanical protocol to replicate impact in walking footwear
Price, C & Cooper, G & Graham-Smith, P & Jones, R

The effect of unstable sandals on single-leg standing
Price, C & Smith, L & Graham-Smith, P & Jones, R

The effect of unstable sandals on instability in gait in healthy female subjects
Price, C & Smith, L & Graham-Smith, P & Jones, R

A comparison of plantar pressures in a standard flip-flop and a FitFlop using bespoke pressure insoles
Price, C & Graham-Smith, P & Jones, R

Does flip-flop style footwear modify ankle biomechanics and foot loading patterns?
Price, C & Andrejevas, V, Findlow, A & Graham-Smith, P & Jones, R

Signature: P Graham

Name: Dr Philip Graham-Smith

Date: 27th May, 2014

Dr Philip Graham-Smith - Head of Biomechanics - Aspire Academy - PO Box 22287 - Doha - Qatar
www.aspire.qa
philip.grahamsmith@aspire.qa
Tel: +974 44138114
Mob: +974 55236240

I declare I am aware that the following manuscript, of which I am a co-author, will form part of the PhD submission by Carina Price:

Does flip-flop style footwear modify ankle biomechanics and foot loading patterns?

Price, C & Andrejevas, V& Findlow, A & Graham-Smith, P & Jones, R

Journal of Foot and Ankle Research (In Review).

I hereby confirm that Carina Price contributed to a major portion of both the literature review, research design, data collection and writing phases. Carina Price was also Corresponding Author for the manuscript.

Signature:

Name: Dr Andrew H. Findlow

Date: 22 May 2014

NHS

Blatchford clinical services

Orthotic Department
Specialist Mobility Centre
17A Meridian East
Leicester
LE19 1WZ

Tel: 07871259470

I declare I am aware that the following manuscript, of which I am a co-author, will form part of the PhD submission by Carina Price:

Does flip-flop style footwear modify ankle biomechanics and foot loading patterns?
Price, C & Andrejevas, V& Findlow, A & Graham-Smith, P & Jones, R
Journal of Foot and Ankle Research (In Review).

I hereby confirm that Carina Price contributed to a major portion of the literature review, research design, data collection and writing phases. Carina Price was also Corresponding Author for the manuscripts. Some of the data within this paper comprised part of my dissertation submission of my Undergraduate degree.

Signature:
Name: Vaidas Tony Andrejevas
Date: 22 May 2014

Yours Sincerely

5.2 Appendix B: Journal Information

Gait & Posture

Impact Factor: 2.123

5 year impact factor: 2.693

Aims and Scope: Gait & Posture is a vehicle for the publication of up-to-date basic and clinical research on all aspects of locomotion and balance. The topics covered include: Techniques for the measurement of gait and posture, and the standardization of results presentation; Studies of normal and pathological gait; Treatment of gait and postural abnormalities; Biomechanical and theoretical approaches to gait and posture; Mathematical models of joint and muscle mechanics; Neurological and musculoskeletal function in gait and posture; The evolution of upright posture and bipedal locomotion; Adaptations of carrying loads, walking on uneven surfaces, climbing stairs etc.; spinal biomechanics only if they are directly related to gait and/or posture and are of general interest to our readers; The effect of aging and development on gait and posture; Psychological and cultural aspects of gait; Patient education.

Footwear Science

Impact factor: N/A

Editor: Edward Frederick

Associate Editors: Nachiappan Chockalingam, Joseph Hamill, Darren Stefanyshyn.

Editorial Board (example participants): Gert-Peter Brüggemann, Dirk De Clercq, Sharon J Dixon, Ewald Hennig, Mario Lafortune, Thomas Milani, Benno Nigg, Martyn Shorten.

Aims and Scope: Footwear Science publishes reports of original research in the disciplines of biomechanics, ergonomics, physiology, clinical science, kinanthropometry, physics, engineering and mathematics. The use of footwear or footwear components, or application of the results to footwear is a major component of the research published in this international, peer-reviewed Journal. Papers published in the journal may cover a wide range of topics within the broad scope of footwear science, including, but not limited to:

- Influence of footwear on kinematics and kinetics of human movement
- Influence of footwear and footwear design on human performance
- Applications of research to design of all types of functional and purpose-built footwear
- Research applied to casual, dress, fashion, duty, athletic, and specialty footwear

- Footwear in prevention and treatment of diseases of lower extremity
- Role of footwear in the prevention and treatment of athletic injury
- Shoe properties and human perceptions
- Human factors applied to fit and function of footwear
- Measurement of footwear biomechanical and physical properties

Journal of Foot and Ankle Research

Impact factor: 1.47

Editors-in-Chief: Hylton Menz and Mike Potter.

Aims and Scope: Journal of Foot and Ankle Research, the official journal of the Australasian Podiatry Council and The College of Podiatry (UK), is an open access, peer reviewed, online journal that encompasses all aspects of policy, organisation, delivery and clinical practice related to the assessment, diagnosis, prevention and management of foot and ankle disorders. Journal of Foot and Ankle Research covers a wide range of clinical subject areas, including diabetology, paediatrics, sports medicine, gerontology and geriatrics, foot surgery, physical therapy, dermatology, wound management, radiology, biomechanics and bioengineering, orthotics and prosthetics, as well the broad areas of epidemiology, policy, organisation and delivery of services related to foot and ankle care. The journal encourages submission from all health professionals who manage lower limb conditions, including podiatrists, nurses, physical therapists and physiotherapists, orthopaedists, manual therapists, medical specialists and general medical practitioners, as well as health service researchers concerned with foot and ankle care

5.3 Appendix C: Conference Abstracts

5.3.1 Abstract: The impact of a health Flip Flop on asymptomatic gait (I-FAB Congress, University of Washington, Seattle, United States, September 2010).

Foot & Ankle International/Vol. 32, No. 3/March 2011 **i-FAB 2010 Podium Abstract** **329**

The impact of a health Flip Flop on asymptomatic gait

C. Price, R. K. Jones, P. Graham-Smith.
Centre for Rehabilitation and Human Performance Research, University of Salford, U.K
Web: www.healthcare.salford.ac.uk/research, email for correspondence: c.l.price@salford.ac.uk

INTRODUCTION

The area of health and wellbeing footwear is highly commercially valuable, however it lacks significant scientific validation. A health flip flop has been produced consisting of a multi-density midsole to produce 'micro-instability' during mid-stance. Consumers report alleviated conditions such as lower back pain, heel pain and knee pain, however the reason for these improvements is not evident. The impact of the health flip flop on gait was assessed in a group of healthy females in order to quantify gait alterations and identify potential mechanisms.

METHODS

Seventeen asymptomatic female subjects (age: 43±11.4 years, height: 162.8±5.13 cm, mass: 64.8±11.46kg) participated in the study. Subjects walked in 4 conditions: barefoot (BF), health flip flop (HF), standard flip flop (SF) and rocker soled sandal (RS). Subjects performed five walking trials in each condition at a self-selected walking pace for each condition. Data was collected from walking trials using a 12 camera Qualysis system and 2 AMTI Force Plates operating at 100 and 3000 Hz respectively. Marker placement defined the foot, shank, thigh, pelvis and thorax segments. Data was exported to Visual 3D for processing and analysis. Variables included temporal parameters, lower limb kinematics, posture, ground reaction force and lower limb joint moments. Kinetics were normalised for body and footwear mass. Statistical analysis, undertaken in SPSS, included ANOVA and Bonferonni adjustment (significance set at .05).

RESULTS

Temporal characteristics identified a non-significant faster walking speed in the HF and longer stance times than other footwear conditions and significantly longer than SF. Kinematics of the lower limb and trunk showed few statistically significant differences between conditions with gait patterns remaining relatively consistent. GRF variables indicated a lower magnitude and longer durations in loading, significant when compared to SF and RS respectively (Table1).

DISCUSSION

The functionality of the HF is evident by the increased walking speed. The increased stance time can potentially be attributed to the construction of the footbed prolonging mid-stance. The HF demonstrated a reduction in foot motion in the frontal plane compared to the SF as well as reductions in loading (Table 1), which may relate to some reported symptoms. Differences with the RS condition were consistent with those previously identified at the ankle and knee [1,2]. Future studies will involve symptomatic populations.

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ACKNOWLEDGEMENTS

Research was supported by FitFlop ltd., and part-funded by the Technology Strategy Board.

Table 1: Selected variables demonstrating differences for HF on the right leg/foot.

Variable	Condition				Significance		
	BF	HF	SF	RS	HF-BF	HF-SF	HF-RS
Walking Speed ($m.s^{-1}$)	1.26±0.097	1.30±0.096	1.28±0.082	1.28±0.090	0.172	1.000	0.875
Stance Time (s)	0.616±0.0377	0.647±0.0362	0.632±0.0372	0.642±0.0377	0.001	0.117	1.000
Maximum Ankle Eversion (°)	-4.4±1.92	-3.7±2.69	-4.1±2.21	-5.7±2.21	1.000	1.000	0.011
Time of Maximum Eversion (% Stance)	46.1±10.6	42.8±11.1	41.2±10.6	26.7±9.7	1.000	1.000	0.000
Total Ankle Coronal Excursion (°)	14.8±3.46	14.2±3.58	15.9±4.54	14.2±3.64	1.000	0.042	0.147
Peak Loading GRF (BW)	1.08±0.055	1.09±0.047	1.12±0.058	1.11±0.056	1.000	0.042	1.000
Time of Peak Loading GRF (% Stance)	22.8±2.34	24.2±2.57	22.7±2.48	21.8±3.72	0.099	0.007	0.048

5.3.2 *Abstract: Single-leg balance in "instability" footwear (I-FAB Congress, University of Sydney, Sydney, Australia, April 2012).*

Price *et al. Journal of Foot and Ankle Research* 2012, **5**(Suppl 1):P10
http://www.jfootankleres.com/content/5/S1/P10

JOURNAL OF FOOT AND ANKLE RESEARCH

POSTER PRESENTATION — Open Access

Single-leg balance in "instability" footwear

Carina Price, Laura Smith, Philip Graham-Smith, Richard Jones*

From 3rd Congress of the International Foot and Ankle Biomechanics Community
Sydney, Australia. 11-13 April 2012

Background

The concept of instability footwear is to reduce stability, increase muscle activation and "tone". Recently numerous brands have developed instability footwear for significant sales. Despite extensive marketing claims there are few empirical studies quantifying effects of instability footwear on muscle activity or motion in healthy individuals aside from Masai Barefoot Technology (MBT^{TM}) [1,2]. The aim of the study was to quantify instability in single-leg standing in a variety of commercially available instability sandals.

Methods

Fifteen female subjects participated (age: 29±6.7 years, mass: 62.6±6.9 kg, height: 167.1±4.2 cm). The protocol quantified Centre of Pressure (CoP) excursion (Kistler) and lower extremity integrated muscle activity (IEMG) (Noraxon) for three thirty second single-leg standing trials in four experimental conditions and one control (Earth Footwear^{TM}). The instability footwear conditions were FitFlop^{TM}, MBT^{TM}, Reebok Easy-Tone^{TM} and Skechers Tone-Ups^{TM}. IEMG is presented normalised to control.

Results

Repeated measures ANOVA revealed significant differences in CoP with MBT having significantly greater anterior-posterior range than Control (p=0.012), FitFlop (p=0.033) and Skechers (p=0.014) (Table 1). Medial-lateral ranges were consistent between conditions. Testing identified increased CoP velocity in anterior-posterior and medial-lateral directions in MBT compared to other conditions, but neither reached significance. IEMG was higher in instability shoes with average increases for gastrocnemius (44%) and peroneals (18%). The only statistical IEMG difference was gastrocnemius in Skechers with a 45% increase compared to control (p=0.042).

Conclusions

Increased anterior-posterior CoP range in MBT is expected due to the rocker profile [2]. Other conditions have footbeds with intrinsic instability not an external feature, which may increase effectiveness in gait. IEMG increased in experimental conditions showing instability shoes increased total activation, however high variability masks statistical differences. Inter-subject differences forms part of on-going analysis. Limitations of single-leg

Table 1 CoP and IEMG results for the footwear conditions.

	Control	Fitflop	MBT	Reebok	Skechers
CoP medial-lateral range (mm)	36.5 (±7.8)	35.5 (±4.1)	34.9 (±3.8)	34.6 (±4.7)	34.0 (±4.3)
CoP anterior-posterior range (mm)	49.6 (±11.1)	53.0 (±8.4)$^{\#}$	64.0 (±10.9)$^{*\#}$	50.3 (±15.0)	49.3 (±12.3)$^{\#}$
CoP medial-lateral velocity (mm.s^{-1})	29.8 (±4.8)	28.7 (±4.9)	30.0 (±6.1)	29.3 (±5.6)	28.5 (±6.2)
CoP anterior-posterior velocity (mm.s^{-1})	26.4 (±3.6)	27.7 (±4.7)	28.4 (±5.0)	27.9 (±4.9)	26.8 (±5.1)
Medial gastrocnemius IEMG (%)	-	1.37 (±0.52)	1.53 (±0.75)	1.39 (±0.64)	1.45 (±0.51)*
Peroneals IEMG (%)	-	1.19 (±0.33)	1.21 (±0.31)	1.15 (±0.22)	1.16 (±0.21)

* Denotes significant difference between control and instability condition (p<0.05)
\# Denotes significant difference between instability conditions (p<0.05)

Centre for Health Sciences Research, University of Salford, Salford, Greater Manchester, M6 6PU, UK

5.3.3 *Poster: Single-leg balance in "instability" footwear (I-FAB Congress, University of Sydney, Sydney, Australia, April 2012).*

University of Salford MANCHESTER

Single-leg balance in "instability" footwear

Carina Price, Laura Smith, Philip Graham-Smith and Richard Jones
Centre for Health Sciences Research, University of Salford
Carina Price: c.l.price@salford.ac.uk

i-FAB.org

Background

The concept of instability footwear is to reduce stability, increase muscle activation and "tone". Recently numerous brands have developed instability footwear for significant profit. Despite extensive marketing claims there are few empirical studies quantifying effects of instability footwear on muscle activity or motion in healthy individuals aside from Masai Barefoot Technology (MBT™) [1, 2]. The aim of the study was to quantify instability in single-leg standing in a variety of commercially available instability sandals.

Methods

Fifteen female subjects participated (age: 29.0±6.7 years, mass: 62.6±6.9 kg, height: 167.1±4.2 cm). The protocol quantified Centre of Pressure (CoP) excursion (Kistler) and lower extremity integrated muscle activity (IEMG) (Noraxon) for three thirty second single-leg standing trials in four experimental conditions and one control (Earth Kalso™). The instability footwear conditions were FitFlop™, MBT™, Reebok Easy-Tone™ and Skechers Fun-Flops™. IEMG is presented normalised to control.

Sandal	Abbrev	Mass (g)	Description
Earth Kalso	CO	193	3.7° incline in footbed from heel to toe with firm footbed and flip flop upper.
FitFlop Walkstar	FF	187	Multi-density ethylene vinyl acetate (EVA) midsole incorporating high-density heel, low-density midfoot and a mid-density forefoot.
MBT Kisumu	MB	534	Thermoplastic polyurethane (TPU) heel and midfoot, compressible heel and pivot under the metatarsals with fibre glass forefoot. Rocker-sole in the anterior-posterior direction.
Reebok Easy-Tone	RE	250	Air-filled compressible elliptical pods positioned under the heel and forefoot, which allow air to travel between the two.
Skechers Tone-Ups	SK	195	Multi-density polymer midsole with firm forefoot.

Results

Repeated measures ANOVA revealed significant differences in CoP with MB having significantly greater anterior-posterior range than CO (p=0.012), FF (p=0.033) and SK (p=0.014) (Table 1). Medial-lateral ranges were consistent between conditions. Testing identified increased CoP velocity in anterior-posterior and medial-lateral directions in MB compared to other conditions, but neither reached significance. IEMG was higher in instability shoes with average increases for gastrocnemius (44%) and peroneals (18%). The only statistical IEMG difference was gastrocnemius in SK with a 45% increase compared to CO (p=0.042).

	CO	FF	MB	RE	SK
CoP medial-lateral range (mm)	36.5 (±7.8)	35.5 (±4.1)	34.9 (±3.8)	34.6 (±4.7)	34.0 (±4.3)
CoP anterior-posterior range (mm)	49.6 (±11.1)	53.0 (±8.4)#	64.0 (±10.9)*,#	50.3 (±15.0)	49.3 (±12.3)#
CoP medial-lateral velocity ($mm.s^{-1}$)	29.8 (±4.8)	28.7 (±4.9)	30.0 (±6.1)	29.3 (±5.6)	28.5 (±6.2)
CoP anterior-posterior velocity ($mm.s^{-1}$)	26.4 (±3.6)	27.7 (±4.7)	28.4 (±5.0)	27.9 (±4.9)	26.8 (±5.1)
Medial gastrocnemius IEMG	-	1.37 (±0.52)	1.53 (±0.75)	1.39 (±0.64)	1.45 (±0.51)*
Peroneals IEMG	-	1.19 (±0.33)	1.21 (±0.31)	1.15 (±0.22)	1.16 (±0.21)

* Denotes significant difference between control and instability condition (p<0.05)
# Denotes significant difference between instability conditions (p<0.05)

Conclusion

Increased anterior-posterior CoP range in MB is expected due to the rocker profile [2]. FF and SK have footbeds with intrinsic instability, with softer EVA enclosed by a flat outsole surface, as opposed to outsole design features, so may not be sensitive within balance trials. RE balance pods may be compressed when weight is evenly distributed across both forefoot and heel pods, negating their effect. Again, dynamic movements may however elicit greater differences. IEMG increased in experimental conditions showing instability shoes increased total activation, however high variability masks statistical differences, consistent with previous literature [1]. Inter-subject differences forms part of on-going analysis. Limitations of single-leg balance mimicking gait are recognised; increased duration of muscle activation is claimed by brands and fixed-duration testing negates this.

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Conflict of interest: The primary author of the paper works on a project funded by FitFlop ltd, supervised by Dr Jones and Dr Graham-Smith. L. Smith has no conflict of interest. Work was undertaken with scientific diligence, data was collected, analysed and the abstract written with no influence from the funding company.

5.3.4 Abstract: Testing a mechanical protocol to replicate impact in walking footwear (I-FAB Congress, Busan, Korea, April 2014).

Price *et al. Journal of Foot and Ankle Research* 2014, **7**(Suppl 1):A68
http://www.jfootankleres.com/content/7/S1/A68

JOURNAL OF FOOT AND ANKLE RESEARCH

MEETING ABSTRACT **Open Access**

Testing a mechanical protocol to replicate impact in walking footwear

Carina Price[1*], Glen Cooper[2], Philip Graham-Smith[1], Richard Jones[1]

From 4th Congress of the International Foot and Ankle Biomechanics (i-FAB) Community
Busan, Korea. 8-11 April 2014

Impact testing is commonly undertaken to quantify the shock absorption characteristics of footwear. The current widely reported mechanical testing method mimics the vertical heel velocity at touchdown and effective mass of the lower limb recorded in running. This therefore results in a greater impact energy than would be expected at touchdown in walking. Despite this mismatch, the methodology is utilised to quantify the shock absorption properties of running and walking footwear alike. The current work modifies the mechanical testing methodology to replicate the kinematics, specifically the vertical heel velocity, identified in walking footwear. Kinematic and kinetic data was collected for 13 subjects walking in four different styles of footwear used for walking (trainer, oxford shoe, flip-flop and triple-density sandal). The kinematic data was utilised to quantify heel velocity at touchdown and accelerometer and force plate data was utilised to estimate the effective mass of the lower limb. When walking in the toe-post style footwear significantly faster vertical heel velocity toward the floor was recorded compared to barefoot and the other footwear styles (Figure 1 for example flip-flop: $0.36\pm0.05 m.s^{-1}$ compared to trainer: $0.18\pm0.06 m.s^{-1}$). The mechanical protocol was adapted by altering the mass and drop height from 10.6-17.3 kg and 2-7 mm, compared to the original protocol of 8.45 kg dropped from 50 mm. As expected, the adapted mechanical protocol produced significantly lower peak force and accelerometer values than the ASTM protocol ($p < .001$). These values more closely resembled those recorded in walking. The mean difference between the human and modified protocol was $12.7\pm17.5\%$ ($p < .001$) for peak acceleration and $25.2\pm17.7\%$ ($p = .786$) for peak force values. The timing of peak force and acceleration variables was less representative of the real-life data with larger mean differences. This pilot test has demonstrated that the altered mechanical test protocol can more closely replicate loading on the lower limb in walking. Further research should consider more variables related to the shock absorption properties of footwear. The results also demonstrate that testing of material properties of footbeds not only needs to be gait style specific (e.g. running versus walking), but also footwear style specific due to the differences in heel touch-down velocity in footwear styles.

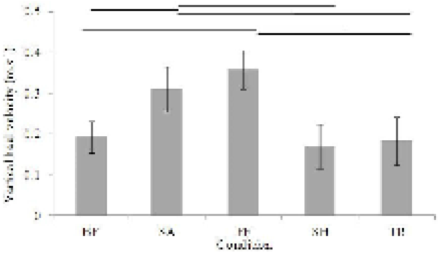

Figure 1 Vertical heel velocity towards the floor in the human testing for the four footwear conditions. Triple-density sandal = SA, flip-flop = FF, shoe = SH and trainer = TR and Barefoot (BF). Error bars denote standard deviation across the 13 subjects tested. Horizontal lines denote statistically significant results (determined by ANOVA where $p < .05$).

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Published: 8 April 2014

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Full list of author information is available at the end of the article

5.4 Appendix D: Reports, Presentations, Marketing and Internal Documents

Research Reports *Studies undertaken as part of the KTP:*	
Symptomatic Gait Studies	Clinical Studies Summary
	Over-pronation Report
	Lower Back Pain Report
	Knee Osteoarthritis Report
Asymptomatic Gait Studies	Summary Healthy Gait Report
	Healthy Gait Report
	Website Healthy Gait Report
Pressure Studies	Female Pressure Summary
	Female Pressure Report
	Press Release from Female Pressure
	Male Pressure Summary
	Male Pressure Report
	Influence of toe-bar on plantar pressure
	Wasted footbed testing
	Varying depth footbeds
Shock Studies	Shock Studies Summary
	Shock Footwear Comparison
	Shock Hardness and Thickness Report
	Pressure Thickness and Hardness Report
Instability study	Instability/Competitor Comparison Report
Comfort	Quantification of subjective and objective comfort in walking footwear, part I
	Quantification of subjective and objective comfort in walking footwear, part II
Other	Stiffness effects of footwear thickness changes
	Hardness review
	Footbed compression

New Product Development *Testing on new styles and comments for new product and development.*	
New Styles	FitFlop Trainer Assessment
	Chic Testing Report
	Leather Boot Report
	Kids Hyker Test
	Pietra Testing
Processes and Ideas	Foot Scanner Repeatability and Validity
	Arch Height Measures
	Timeframe Considerations
	Fit Considerations
	Impact Characteristics SATRA Comparison
	Product Consistency Report
	Product Consistency Update
	Wear Trial Report
Presentations *Presentations undertaken for informative, handover or academic purposes.*	
	KTP Introduction Presentation
	UK AW10 Launch
	UK AW10 Handout
	US AW10 launch
	AW13 Sales Presentation
Other documents	
	'Health and Well-being' Footwear Review
	FitFlop Consumer Testimonials Feedback
	Scientific Marketing Review
	Strategic Product Research and Development Plan

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