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Improving the validity of shod human footstrike modelling with dynamic loading conditions determined from biomechanical motion capture trials

Iain Hannah

A doctoral thesis submitted in partial fulfilment of the requirements for the award of Doctor of Philosophy of Loughborough University.

2014
Abstract

This thesis presents and evaluates a number of finite element footstrike models developed to allow the performance of prospective athletic footwear designs to be evaluated in a virtual environment. Successful implementation of such models would reduce the industry’s traditional reliance on physical prototyping and therefore reduce the time and associated costs required to develop a product.

All boundary conditions defined in each of the footstrike models reported were directly determined from biomechanical motion capture trials to ensure that the loading applied was representative of shod human running. Similarly, the results obtained with each model were compared to digitised high speed video footage of experimental trials and validated against biomechanical measures such as foot segment kinematics, ground reaction force and centre of pressure location.

A simple model loaded with triaxial force profiles determined from the analysis of plantar pressure data was found to be capable of applying highly representative load magnitudes but the distribution of applied loading was found to be less accurate. Greater success at emulating the deformation that occurs in the footwear during an entire running footstrike was achieved with models employing kinematic foot segment boundary conditions although this approach was found to be highly sensitive to the initial orientation of the foot and footwear components, thus limiting the predictive capacity of such a methodology. A subsequent model was therefore developed to utilise exclusively kinetic load conditions determined from an inverse dynamic analysis of an experimental trial and demonstrated the greatest predictive capacity of all reported models. This was because the kinematics of the foot were allowed to adapt to the footwear conditions defined in the analysis with this approach.

Finally, the reported finite element footstrike models were integrated with automated product optimisation techniques. A topology optimisation approach was first utilised to generate lightweight midsole components optimised for subject-specific loading conditions whilst a similar shape optimisation methodology was subsequently used to refine the geometry of a novel footwear design in order to minimise the peak material strains predicted.
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Finally I would like to thank my family around the country and the technicians, administrative staff, academics, final year project students and researchers at Loughborough University’s Sports Technology Institute. Your support and friendship has made my time in the department very memorable.
Publications Arising from this Work


“Buddy, the wind is blowing.”

Truman Capote, *A Christmas Memory*
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1. Introduction

The global sporting goods industry is a hugely lucrative market forecast to reach an estimated $266 billion by 2017 (Lucintel 2012). The market is typically broken down into three primary segments - athletic apparel, athletic footwear and equipment. The focus of this research is on athletic footwear which was estimated to represent $13 billion of the $74 billion American sporting goods market in 2010 (SGMA 2011).

The competitiveness of this market has seen athletic footwear brands invest significantly in the design of novel footwear technologies, each trying to better the products of their rivals and satisfy increasing consumer demand for enhanced performance. The footwear development process is typically iterative with prototype designs evaluated via a combination of laboratory based mechanical tests and biomechanical user trials, although both test types have associated limitations.

Mechanical tests allow for standardised evaluations of prospective footwear designs to be made in a controlled environment, thus avoiding the variation inherent to human testing. The loads that occur during a human footstrike are however complex, transient and multiaxial whilst mechanical footwear tests commonly involve somewhat arbitrary loads being applied in a quasi-static, uniaxial manner. As such, the tests can often be considered to provide a poor representation of the loading that occurs during a human footstrike, thus limiting the value of the results obtained.

In contrast, employing human subjects in biomechanical user trials establishes a greater degree of realism in the test procedure but also introduces a number of related limitations. The results of any biomechanical test are unique to both the individual gait of the subject performing the trial and the footwear worn. Results obtained for a particular subject may therefore not necessarily be predictive for another, whilst it is also difficult to determine if variations observed between trials are as a result of the different footwear conditions tested or merely due to the poor repeatability associated with human testing. Extended user wear trials also introduce further uncertainty to the methodology as no control of variables such as the weather or running surface is possible.
In addition to the limitations of these traditional mechanical and biomechanical test methods, the iterative footwear development process is time-consuming, expensive and, most significantly to this thesis, entirely reliant on the manufacture of physical prototypes. As a consequence, the footwear industry has begun to adopt advanced computer-aided engineering (CAE) techniques to shorten development timelines, minimise costs and reduce this traditional dependence on physical prototyping.

### 1.1. Research Objectives

The principal goal of this research project was to develop a range of biomechanically representative CAE footstrike models employing finite element analysis (FEA), a numerical technique used to characterise an engineering problem. A robust and reliable dynamic finite element footstrike model would be a powerful tool in the footwear development process, allowing prospective footwear designs to be evaluated in a virtual environment such that informed design decisions could be made. This would reduce the expensive and time consuming dependence on physical prototyping and manufacture which currently exists as part of the typical footwear development process.

In order to provide an accurate prediction of an item of footwear’s response to loading a footstrike model would have to contain accurate footwear geometries, an appropriate mesh and sophisticated material models characteristic of those used in modern athletic footwear. Furthermore, the boundary conditions used in any model would have to be representative of the complex, multiaxial and dynamic loading applied to footwear structures during a human footstrike and it was this that formed the primary focus during the development of the finite element (FE) footstrike models reported in this thesis. To achieve this it was decided that all boundary conditions would be determined from biomechanical motion capture trials, irrespective of the modelling approach employed.

At the beginning of the project a number of fundamental research objectives were established with each footstrike model subsequently appraised against the following evaluation criteria:

- **Biomechanically representative**: A number of FE models developed for use in the footwear development process have already been reported (e.g. Gibbs 2006; Mara 2007) but these tend to merely emulate the mechanical laboratory tests discussed previously in that the loading is overly simplistic and thus a poor representation of that
applied during a human footstrike. When analysing the results of any model, an engineer must be aware of the sources of error inherent to the finite element method and ensure that all new problem types are thoroughly validated (Becker 2004). Output obtained with each of the FE footstrike models presented in this thesis was therefore compared against biomechanical test data recorded from experimental motion capture trials. The extent to which each methodology applied loading representative of a shod human footstrike could thus be appraised with improvement over current industry practice reported where appropriate.

- **Demonstrated predictive capacity**: The described validation procedure ultimately results in previously performed experimental trials being emulated with an FEA model. Whilst necessary, this negates the primary advantage of FEA as physical prototypes must still be manufactured and experimental testing still performed. The capacity of each FE footstrike model to be used to predict footwear performance and allow a comparison of two prospective footwear designs to be made was therefore evaluated.

- **Universal applicability**: All of the footstrike modelling methodologies reported in this thesis were initially developed to emulate the loads applied during a rearfoot running footstrike. Many types of athletic footwear are however required to withstand the loads associated with other types of athletic actions such as cuts, jumps and landings. The potential of each footstrike modelling methodology to be applied to alternative footstrike types was therefore assessed with more generic methodologies obviously considered to be of greater value.

- **Appropriate for industry implementation**: The athletic footwear industry is highly competitive with product development cycles becoming increasingly smaller. To justify the implementation of an FE footstrike model in the footwear development process it must demonstrate that its use enables quick but informed design decisions to be made. Strategies employed during the development of each FE footstrike model to achieve manageable solve times and minimise the pre- and post-processing required for each analysis are therefore reported throughout this thesis.
1.2. Thesis Overview

This thesis documents the development of a number of different FE footstrike modelling methodologies to be used for the virtual testing of athletic footwear. Beginning with an overview of the anatomy of the human lower extremity, Chapter 2 presents and reviews previously published research handling the topics of greatest relevance to the project. An introduction to the biomechanics of shod human running is provided followed by a discussion on the types of injuries reported to afflict runners most frequently. An overview of the anatomy of a modern running shoe is then presented with the primary role of each component discussed. Finally, previously reported FE models of the human foot and footwear considered most relevant to this thesis are discussed before an opportunity for novel research is identified.

Chapter 3 details both the procedures used during biomechanical test sessions and methodologies adopted to determine the kinematic and kinetic boundary conditions employed in the FE footstrike models reported in later chapters. An overview of the test methodology, subjects, footwear and hardware used during the performed motion capture trials is first presented with the biomechanical model used to determine foot segment kinematics reported thereafter. The methodology used to determine the coefficient of friction between each footwear type and the laboratory force platform is also discussed.

Chapter 4 introduces FEA, providing an overview of the fundamental mathematical principles of the method and modelling software selected. A preliminary footstrike model is then used to illustrate the procedures associated with the development of an FE model including the definition of appropriate material models, meshing strategies and output requests. The assumptions and limitations of the finite element method with regards to this thesis are discussed throughout.

Chapter 5 details a method to emulate the loading applied during a rearfoot running footstrike using in-shoe plantar pressure measurements to determine kinetic load conditions. Model output is compared to experimental high speed video footage before the predictive capacity of the modelling approach is evaluated by altering the defined material models.

Chapter 6 reports the development of a kinematically driven footstrike model employing six degree of freedom boundary conditions determined for a three segment foot model.
potential application of the method to predict rearfoot and forefoot footwear stiffnesses is presented followed by an analysis of the sensitivity of model output to the defined kinematic boundary conditions. A more advanced modelling approach also including simple kinetic load conditions is detailed in Chapter 7 with a comparison of the predictive capacities of the two respective methodologies subsequently provided.

Chapter 8 presents a footstrike modelling methodology driven with exclusively kinetic load conditions determined from the inverse dynamic analysis of a rearfoot running footstrike. After reporting the biomechanical techniques used to determine the applied load conditions, the model is used to predict foot and footwear kinematics after orthotic intervention with the analysis results compared to reported clinical studies.

Based on the pedobarographic footstrike modelling approach first presented in Chapter 5, Chapter 9 details a methodology developed to automatically output optimised footwear geometries. A topology optimisation approach is first employed to develop lightweight midsole designs optimised for the specific load conditions defined before a shape optimisation approach is employed to minimise predicted stress concentrations for a novel footwear design currently under development.

Finally, Chapter 10 provides a conclusion to the thesis and reviews to what extent the research objectives presented in Section 1.1 have been met. The practicality of using each of the previously reported FE footstrike models in the footwear development process is discussed before a number of areas for future research are suggested.
2. Literature Review

2.1. Anatomy of the Human Lower Extremity

An understanding of the anatomy of the human lower extremity is essential in the development of athletic footwear. Whilst 93 % of running injuries occur inferior to the waist, only 31 % occur in the foot or ankle (Epperly & Fields 2001) and different athletic footwear is also known to affect the loading and kinematics of the entire body (Clarke et al. 1983; Nigg & Morlock 1988). Although the focus of this thesis is undoubtedly the foot-shoe interface, this information suggests that it would be spurious to disregard the remainder of the lower extremity. Consequently, the anatomy of the hip and all inferior structures are discussed in this section with the anatomy of the foot and ankle being examined in greater detail.

2.1.1. Hip Joint and Pelvic Girdle

The acetabular femoral joint, or hip joint is one of the most stable in the body due to its bony architecture, strong ligaments and large, supportive muscles (Floyd 2006). The femoral head is received by the acetabulum of the pelvic girdle to form a ball and socket joint. The joint’s multiaxial arrangement and relative stability that it is well adapted for weight bearing and locomotion (Van de Graaff 1998).

The skeletal arrangement of the pelvic girdle and hip joint can be seen in Figure 2.1. The pelvic girdle consists of the right and left pelvic bones (ossa coxae) which meet at the symphysis pubis joint (Van de Graaff & Fox 1995). The pelvic girdle is attached to the axial skeleton posteriorly at the sacrum, an extension of the spinal column made up of five fused vertebrae. Each pelvic bone consists of three main bones - superiorly the ilium, inferiorly and anteriorly the pubis and inferiorly and posteriorly the ischium. These are all fused together in a mature adult and provide an extensive surface for the attachment of muscles, tendons and ligaments (Martini 2009).
The majority of the muscles that act at the hip are large and strong, providing great stability and a large range of motion. The muscles of the hip can be seen in Figure 2.2. It is possible to divide the muscles that control the hip joint into two groups - those primarily at the pelvis and those primarily at the thigh. It should however be noted that this area contains several biarticular muscles which act at both hip and knee joints (Floyd 2006). Only their role in controlling the hip is discussed here.

The muscles of the pelvis can be further divided into two main groups - the anterior iliac region and posterior gluteal region. The iliac region is responsible for hip flexion and contains the iliacus\(^1\), and the psoas major\(^2\) and psoas minor\(^3\) which together form the iliopsoas\(^4\). The 10 muscles of the gluteal region function primarily to extend and rotate the hip. They include the gluteus maximus\(^5\), gluteus medius, gluteus minimus and tensor fasciae muscles (Floyd 2006).

The muscles of the thigh can also be divided into distinct compartments. The anterior compartment controls hip flexion and rotation and consists of the sartorius\(^6\) and quadriceps muscle group (vastus medialis\(^7\), vastus intermedius\(^8\), vastus lateralis\(^9\) and rectus femoris\(^10\)). The hamstring group of muscles consisting of the biceps femoris\(^11\), semitendinosus\(^12\) and semimembranosus\(^13\) is contained within the posterior and lateral compartment and is responsible for hip extension and abduction. Lastly, the medial compartment contains the
adductor brevis\textsuperscript{14}, adductor longis\textsuperscript{15}, adductor magnus\textsuperscript{16}, pectineus\textsuperscript{17} and gracilis\textsuperscript{18} which are primarily responsible for hip adduction (Floyd 2006).

There is no definitive consensus about the exact possible range of motion at the hip due to individual differences, but the values are generally agreed to be $130^\circ$ of flexion, $30^\circ$ of extension, $35^\circ$ of abduction, $30^\circ$ of adduction, $45^\circ$ internal rotation and $50^\circ$ external rotation (Floyd 2006). The prevalence of biarticulate muscles in this region however means that the positioning of the knee can lead to a degree of active or passive insufficiency at the hip. Active insufficiency occurs when a biarticulate muscle cannot exert enough tension to complete the full range of motion in both joints simultaneously whilst passive insufficiency is the result of the muscle not being able to stretch enough to allow both joints their full range of motion simultaneously (Backhouse & Hutchings 1986).
2.1.2. The Knee

The joint of the knee is the largest in the human body. This complex joint has combined functions of locomotion and weight bearing which place it under considerable stress. Powerful knee joint flexor and extensor muscles combined with a strong ligamentous structure provide a strong functioning joint when healthy (Floyd 2006).

The knee joint is a ginglymus (hinge) joint and is primarily responsible for knee flexion and extension. According to Floyd (2006), the knee joint can be extended to 180° but it is not uncommon for an individual to be able to hyperextend the joint by 10° or more. The knee can typically flex to about 140°. Whilst no abduction or adduction is possible at the knee, a small degree of rotation is attainable provided that the knee is flexed by at least 30°. When this is the case, 30° of internal rotation and 45° of external rotation are achievable. As discussed previously, the position of the hip can also affect the range of motion possible at the knee due to active and passive insufficiency (Backhouse & Hutchings 1986).

The structures of knee can be seen in Figure 2.3. The femur\textsuperscript{1} articulates with the tibia\textsuperscript{2} to form the knee joint. Both bones have large condyles and the top of the tibial condyles, the tibial plateaus, act as receptacles for the femoral condyles. The fibula\textsuperscript{3} articulates with the tibia inferior to the lateral tibial condyle. It bears little weight and does not articulate with the femur but rather acts as an important attachment point for a variety of muscles, tendons and ligaments (Martini 2009). The patella\textsuperscript{4} is a sesamoid bone and sits anteriorly to the knee. It offers protection to the knee joint but also provides a mechanical advantage during knee extension by increasing the leverage of the quadriceps femoris muscle (Van de Graaff & Fox 1995).
The complexity of the knee and great amount of stress exerted upon it necessitate a number of stabilising structures. Static stability is afforded by ligamentous structures whilst the musculature of the quadriceps and hamstrings provide dynamic stability. In addition to the articular cartilage which exists in all diarthroidal joints, the knee contains menisci, specialised cartilage structures which cushion the bones and deepen the tibial plateaus, thereby enhancing stability (Floyd 2006). Rotary stability is provided by the anterior and posterior cruciate ligaments, so called as they cross within the joint between the tibia and femur. Medial and lateral stability are provided by the medial and lateral collateral ligaments respectively (Gray 2004).

As the knee is a hinge joint, it is possible to divide the musculature of the knee into anterior and posterior compartments. The muscles which act at the knee are best observed in
Figure 2.3. The anterior compartment contains the sartorius\textsuperscript{6} and quadriceps femoris group\textsuperscript{7,8,9,10} which extend the knee by attachment to the tibial tuberosity via the patellar tendon (Van de Graaff 1998). The hamstring muscle group of the posterior compartment is responsible for knee flexion and rotation. The main knee flexors are the biceps femoris\textsuperscript{11}, semitendinosus\textsuperscript{12} and semimembranosus\textsuperscript{13}. The biceps femoris is also responsible for external rotation of the knee whilst the semitendinosus and semimembranosus are aided by the popliteus in rotating the knee internally.

2.1.3. The Ankle and Foot

The complexity of the foot is evidenced by the number of bones, muscles and ligaments which make up its structure. The primary roles of the foot are support and propulsion (Floyd 2006).

![Figure 2.4 - The bones of the right foot. a) Superior view, and b) inferior view. From Van de Graaff (1998).](image)

The bones of the foot are displayed in Figure 2.4. The talus articulates with the distal tibia and distal fibula to form the ankle joint, technically known as the talocrural joint. The ankle is a hinge joint and allows approximately 50° of plantarflexion and 15-20° of dorsiflexion (Luttgens & Hamilton 2007). A greater range of dorsiflexion is achievable when the knee is flexed as this reduces tension in the gastrocnemius (Floyd 2006). The rearfoot segment contains the talus\textsuperscript{1}}
and calcaneus² which accept weight from the tibia and distribute it to the ground through the other bones of the foot (Martini 2009).

The midfoot segment contains five bones which, together with the talus and calcaneus are known collectively as the tarsals³. The cuboid⁴ and navicular⁵ are the most proximal of these. The navicular separates the medial⁶, intermediate⁷ and lateral⁸ cuneiforms from the talus whilst the cuboid lies between the calcaneus proximally and the fourth and fifth metatarsals distally. It is the most lateral in the distal tarsal row (Gray 2004).

Distal to the tarsal bones are the five metatarsals⁹, each corresponding to one of the toes. With the exception of the great toe (hallux) which only contains two, each of the toes contain three phalangeal bones - proximal, middle and distal. The five metatarsals and 14 phalanges¹⁰ collectively form the forefoot segment (Floyd 2006).

The arches of the foot are held in place by the foot’s ligamentous and tendonous structures. They are illustrated in Figure 2.5. They help to support the weight of the body and provide leverage for propulsion. These arches are not rigid, but rather yield under loading and spring back as the load is removed (Van de Graaff & Fox 1995). The longitudinal arch can be divided into medial and lateral segments but is usually considered as one. The medial longitudinal arch extends from the calcaneus to the talus, the navicular, the three cuneiforms, and the distal ends of the three medial metatarsals. The longitudinal lateral arch originates at the calcaneus and extends to the cuboid and the distal ends of the fourth and fifth metatarsals. The transverse arch extends across the foot from one metatarsal bone to the other (Floyd 2006; Van de Graaff & Fox 1995).
Figure 2.5 - The arches of the foot. From Van de Graaff & Fox (1995).

At the ankle complex, dorsiflexion-plantarflexion occurs in the sagittal plane, abduction-adduction occurs in the transverse plane and inversion-eversion occurs in the frontal plane. This is illustrated in Figure 2.6:

Figure 2.6 - Axes of motion in the foot and ankle. From Sammarco & Hockenbury (2001).

Inversion and eversion are commonly believed to be articulations of the ankle joint but technically occur in the subtalar and transverse tarsal gliding joints. Combined, these joints enable 5-15° of eversion and 20-30° of inversion. The subtalar joint is formed by the talus and
calcaneus whilst the transverse tarsal joint is an articulation of the talus with the navicular, and of the calcaneus with the cuboid (Lundberg & Svensson, 1993). The calcaneocuboidal joint has a range of movement of 10-25° whilst the talonavicular joint allows for 30-50° of motion. There is limited movement in the remainder of the intertarsal joints (van Langelaan, 1983).

Motion of the forefoot relative to the rearfoot is facilitated by the transverse tarsal and tarsometatarsal joints. The tarsometatarsal joint is the articulation of the metatarsals with the four most distal tarsal bones - the three cuneiforms and the cuboid (Kapandji 1974). They are approximately plane synovial (gliding) joints but are only capable of relatively small displacements. More motion is achievable on the lateral side than on the medial side (Gray 2004).

The metatarsals meet the phalanges at the ellipsoidal metatarsophalangeal joints. At the hallux, flexion of approximately 45° and extension of 70° is available but a smaller range of motion is available at each of the lesser toes, approximately 40° of flexion and 40° of extension. A small degree of abduction and adduction is however available at the lesser toes (Kapandji 1974).

Ginglymus interphalangeal joints exist in all of the toes, one in the hallux and two, proximal and distal, in each of the lesser toes. The range of movement achievable varies greatly between individuals but can be approximated to 0° of extension and 90° flexion in the hallux, 0° of extension and 35° of flexion in the proximal interphalangeal joints, and 30° of extension and 60° of flexion in each of the distal interphalangeal joints (Floyd 2006).

The muscles which act extrinsically upon the foot can be broadly divided by location and function. They are shown in Figure 2.7. The primary dorsal flexors are contained within the anterior compartment of the foot and ankle, consisting of the tibalis anterior\(^1\), peroneus tertius, extensor digitorum longus\(^2\) and extensor hallucis longus. Dorsiflexion is facilitated by the anterior tibial tendon which attaches on the anteromedial dorsal aspect of the foot. The two most powerful evertors are the peroneus longus\(^3\) and peroneus brevis, located in the lateral compartment and connected to the medial cuneiform and first metatarsal foot via the peroneal tendons (Floyd 2006).

The seven posterior crural muscles can be divided into a superficial group and a deep group. The superficial group primarily contains plantarflexors. It comprises of the gastrocnemius\(^4\),
soleus⁵ and plantaris⁶ muscles which facilitate plantarflexion via the Achilles tendon which attaches to the posterior calcaneus. The deep posterior group consists of the popliteus, flexor hallucis longus, flexor digitorum longus and the tibalis posterior muscles. With the exception of the popliteus, they are plantarflexors but also act as invertors via the posterior tibial tendon. There is no musculature attachment on the anteromedial aspect of the tibia (Van de Graaff 1998).

Figure 2.7 - Superficial muscles of the right lower leg. a) Anterior view, and b) posterior view. From Floyd (2006).

The intrinsic muscles of the foot follow the primitive limb pattern of dorsal extensors and plantarflexors (Gray 2004). They can be grouped into four layers and are responsible for either moving the toes or for supporting the arches of the foot (Van de Graaff 1998). On the dorsal aspect of the foot are the extensor digitorum brevis and extensor hallucis brevis which act to dorsiflex the phalanges. Plantarflexion of the phalanges is facilitated by the flexor hallucis brevis¹, flexor digitorum brevis² and, to a lesser extent the abductor hallucis³ which also acts to abduct the hallux. The dorsal interossei is responsible for abduction of the lesser toes whilst adduction is facilitated by the plantar interossei⁴ (Platzer 1978). The intrinsic muscles of the foot are shown in Figure 2.8:
Due to its unique role in weight bearing and locomotion, the plantar aspect of the foot contains a number of specialised areas of soft tissue such as the heel pad, a region of fatty adipose tissue that is located inferior to the calcaneus and has an average thickness of between 17.8mm and 20.1mm (Steinbeck & Russell, 1964). This tissue is anatomically adapted to withstand the repeated high compressive and shearing loads which occur as a result of bipedal locomotion (Pain & Challis 2001). Fatty adipose tissue is also particularly thick around the metatarsophalangeal joint (MPJ) and acts to attenuate loading during push-off (Gray, 2004). The plantar fascia is a band of fibrous connective tissue which runs along plantar aspect
of the foot from the tuberosity of the calcaneus to the heads of metatarsals. It undergoes tension during weight bearing and thus acts to support the longitudinal arch of the foot (Kim & Voloshin 1995).

2.2. Biomechanics of Running

2.2.1. Fundamentals of Human Gait

The objective of bipedal locomotion, or gait, is to efficiently translate the body’s centre of mass in the overall direction of locomotion (Novacheck 1998). Human gait is cyclic and consists of phases of stance and swing. Each cycle begins when one foot first contacts the floor and ends when the same foot contacts the floor again. During walking, two phases of double limb support can be observed during a single gait cycle. Running is distinguished from walking by increased translational velocity and the presence of two periods of unsupported locomotion, or float (Dugan & Bhat 2005). A comparison of the two differing gait cycles is shown in Figure 2.9.

Figure 2.9 - Comparison of gait cycles. a) Illustration of walking, b) schematic of walking, and c) schematic of running. Adapted from Novacheck (1998), Dugan & Bhat (2005).
Novacheck (1998) stated that the transition from running to sprinting is generally recognised to occur at the point when initial ground contact transfers from the rearfoot to the midfoot. A study by Payne (1978) concluded that sprinters and middle distance runners generally initiate contact with the ball of the foot whilst most long distance runners make first contact at the heel. Kerr et al. (1983) also identified that initial contact moves anteriorly as foot velocity increases. The centre of pressure at initial contact is known to vary greatly amongst distance runners with some runners landing on the midfoot or forefoot irrespective of speed (Cavanagh & Lafortune 1980) but this distinction between running and sprinting is generally agreed upon. Whilst no differences in metabolic expense have been observed between rearfoot and forefoot strikers (Ardigo et al. 1995), a rearfoot striking technique is predominantly adopted by long distance runners (Payne 1983; Larson et al. 2011; Kerr et al. 1983) despite a reported increase in exposure to repetitive stress injuries (Daoud et al. 2012). As such, the biomechanics of rearfoot running is considered in more depth in this section.

2.2.2. Running Kinematics

Running velocity is the product of stride length and stride rate (Cavanagh & Kram 1990). A curvilinear relationship between stride length and running speed has been identified (Dillman 1975) but the relationship can be considered linear between 2.5m/s and 6m/s, a typical range for distance running (Cavanagh & Kram 1990; Luhtanen et al. 1978). As such, stride frequency can also be considered to increase proportionally to velocity within this range (Williams 1985).
Figure 2.10 - Ankle, knee and hip joint angles throughout a running cycle at two different running speeds. All curves are for the left side. Recreated from Williams (2000).

As illustrated in Figure 2.10, the kinematics of the lower extremity vary according to running speed. The following discussion of lower limb kinematics is based on this data set and further literature published by Milliron & Cavanagh (1990), Nigg et al. (1987) and Nilsson et al. (1985).

Prior to footstrike, extension of the hip has begun but the impact forces cause a brief period of flexion before extension resumes. The hip angle generally follows a sinusoidal pattern with maximum extension occurring just before toe-off and maximal flexion being observed in the mid to terminal stance phase. At higher speeds, additional hip flexion and extension is accompanied by rotation of the pelvic girdle to enable a larger stride length (Floyd 2006).

The knee experiences two periods of flexion during each gait cycle, once during support and once during swing. The knee is flexed during the swing phase to reduce its moment of inertia. This reduces the energy required to accelerate the leg through to the next footstrike and is
more pronounced at higher speeds. Maximum knee flexion is increased at higher running speeds but less extension is observed at the footstrike.

The ankle of a rearfoot striker experiences a period of rapid dorsiflexion after initial contact followed by plantarflexion during push-off. Increased speed increases the degree of plantarflexion observed during push-off but there is little discernible effect on the maximum angle of dorsiflexion.

The rearfoot kinematics of a footstrike are of particular interest to biomechanists due to the proposed relationship between excessive pronation and overuse injuries of the lower extremity. This is discussed further in Section 2.2.4. Three dimensional studies have been performed to obtain a more complex understanding (Engsberg 1996; Nawoczenski et al. 1998) but it is generally accepted that pronation occurs when the foot is everted, abducted and dorsiflexed, whilst supination occurs when the foot is inverted, adducted and plantarflexed (James et al. 1978). Three footstrikes displaying varying levels of pronation are illustrated in Figure 2.11:

---

**Figure 2.11** - Left rearfoot kinematics of three different landing styles. a) Overpronation, b) normal pronation, and c) supination. Adapted from Nigg (1986).
A rapid period of pronation was observed in rearfoot strikers for the first 70% of the footstrike by James et al. (1978). He states that the ability of the joints in the forefoot to be unlocked during pronation allows the foot to be flexible and adapt to the surface during the support phase of gait. During push-off, the foot must become a rigid lever to minimise energy losses. Supination occurs during the final 30% of the footstrike to enable this (Wright et al. 1964) although it should be noted that, as with any structure in the human body, strict rigidity is not achieved and some degree of flexion still occurs.

Several factors have been found to affect pronation. Williams and Ziff (1991) associated greater maximal pronation with a reduced vertical angle of the lower leg and a less supinated rearfoot position at heelstrike. It was also found that increased pronation speed was linked to a more supinated rearfoot position and a greater heel angle with the vertical at heelstrike. In addition to this, increased stride length was found to reduce pronation. Nawoczenski (1998) found a relationship between arch shape and pronation with a flat foot being associated with more pronation and a high arch being associated with less pronation.

The type of running footwear worn can also have an effect on pronation. Hamill et al. (1988) found that greater pronation occurred when wearing lightweight racing shoes than when wearing heavier training shoes. Nigg et al. (1986) found harder midsole materials increased initial pronation velocity as a result of a stiffer lever arm system whilst overall pronation was increased by a more compliant midsole in a study by Clement et al. (1981). Consequently, Nigg et al. (1986) concluded that a softer midsole material at the lateral heel and a harder material at the medial heel would reduce pronation.

Evidence suggests that the proportions of the human forefoot have evolved to minimise the metabolic demands of endurance running (Rolian et al. 2009). Bojsen-Møller & Lamoreux (1979) studied the kinematics of the propulsive phase of gait and recognised that it could be broken down into two distinct phases. These are illustrated in Figure 2.12. The digitigrade phase involves a 60° rotation of the calcaneus about the metatarsophalangeal joint axis. The unguligrade phase follows and involves simultaneous plantarflexion of the toes and a 90° rotation of the MPJ about the tip of the hallux.
2.2.3. Running Kinetics

Despite having no clearly established definition, “cushioning” is a commonly used term in footwear literature and is referenced frequently in this section. When used in this thesis, “cushioning” refers to a material or piece of footwear’s ability to dissipate energy, avoid plantar pressure peaks and minimise the acceleration of foot tissues.

Ground reaction forces

A ground reaction force (GRF) is equal in magnitude and opposite in direction to that applied to the ground during the stance phase of gait. Interpretation of GRF data is frequently used to evaluate the cushioning performance of running shoes but this approach has some key limitations (Hamill 1999). A force platform can only describe the acceleration of the whole body’s centre of mass. The GRF is actually the vector sum of the accelerations acting on several segmental masses (Bobbert et al. 1991). It is therefore not possible to evaluate specific actions at the foot-ground interface and this technique should be used with caution when assessing cushioning performance.

As an alternative, Shorten & Winslow (1993) stated that tibial acceleration could be evaluated in the frequency domain rather than the time domain. A fast Fourier transform can be used to convert GRF data into the frequency domain for analysis. This information could then be used to compare the shock attenuating properties of different footwear constructions (Hamill 1999). Despite these and further documented limitations of GRF analysis (e.g. Miller 1990; Nigg 2001), it remains a useful and commonly used tool in footwear design.
The resultant GRF from a force platform can be divided into three components - the vertical force component, anteroposterior force component and mediolateral force component. A typical GRF profile of a shod rearfoot striker is shown in Figure 2.13:

![Figure 2.13 - Typical ground reaction force components for shod rearfoot running.](image)

For a running speed of 4.5 m/s, vertical GRF magnitudes of 2.2 bodyweights (BW) and 2.8 BW were reported for the passive peak and active peak respectively (Cavanagh & Lafortune 1980). The passive peak was found to occur in the first 50 ms after contact whilst the active peak was identified to occur at between 35 % and 50 % of total stance time. Similar GRF values were found in a re-examination of the subject area by Munro et al. (1987).

Published literature has demonstrated a positive correlation between the vertical GRF component and running speed. Cavanagh & Lafortune (1980) reported a passive peak of 2 BW at 3 m/s and 3 BW at 6 m/s. Similarly, Nigg (1986) found that the magnitude increased from between 1300 N and 1400 N to between 2090 N and 2240 N at the same speeds. Munro (1987) examined the value of the active peak and found that it increased from 2.5 BW to 2.8 BW at 3 m/s and 6 m/s respectively.
The effect of footwear cushioning on GRF was explored by Bates (1989) with a lack of difference in impact force magnitudes reported. Similar results were found by Stiles & Nixon (2007) in their study of tennis specific movements whilst one study actually found more compliant midsoles increased impact forces (Nigg et al. 1981). The results of these studies are not consistent with accepted theory, mathematical models or in vivo and in vitro test results.

The limitations of GRF analysis have been suggested as an explanation for this but it also seems apparent that runners make kinematic adjustments to their style in order to control the amount of loading that occurs (Snel et al. 1985; Stiles & Dixon 2007). Proposed adjustments include increased initial knee flexion, reduced vertical heel velocity at impact and the adoption of a reduced sole angle when impacting stiffer surfaces (Bobbert et al. 1992; De Wit & De Clercq 1997; Denoth 1986; Dixon et al. 2005; Frederick 1986; Gerritsen et al. 1995). Lafortune & Lake (1995) also highlighted that the properties of the plantar soft tissue play a significant role in load attenuation and remain almost constant in all shod conditions.

Initial loading rate was however found to be affected by footwear’s cushioning properties. Clarke et al. (1983) found that the time to the initial passive force peak was greater in soft soled shoes, therefore reducing the rate of loading on the body. Similar findings were reported by Snel et al. (1985).

This was also supported by a study by De Wit et al. (2000) which compared the vertical GRF profiles for shod and barefoot runners. As shown in Figure 2.14, barefoot running is characterised by a significantly higher loading rate after initial impact. It has been suggested that peak loading rate is a better indicator of cushioning performance than peak loading force (Stiles & Dixon 2007) whilst Hennig et al. (1996) found that subjects could accurately determine different shoes’ cushioning performance based on the peak loading rates and maximum rearfoot plantar pressures experienced.
The anteroposterior GRF component consists of two clear phases. The initial negative phase represents the period of deceleration in which a braking force is being applied to the foot, primarily by the quadriceps muscle group (Hamner et al. 2010). The second, positive phase corresponds to the push-off in which a propulsive force is applied to the ground in order to accelerate the centre of mass of the runner. The quadriceps and plantarflexors are the main contributors to this acceleration (Hamner et al. 2010). The runner’s speed is maintained if the breaking and propulsive forces are equal. A typical anteroposterior GRF component for a rearfoot striker running constantly at 4.5 m/s was found to be 0.43 BW by Cavanagh & Lafortune (1980).

A limitation of conventional force platforms is that they are only able to resolve the reaction forces applied upon the contact area as a whole. The distribution of vertical loading on the foot can be determined with pressure measurement systems and is discussed later but the distribution of horizontal force components cannot be recorded with conventional pressure measurement systems. To overcome this, a shoe sole that can be loaded with five miniature triaxial sensors (USL06-H5, size: 20 × 20 × 5 mm, TecGihan Co. Ltd., Japan) was developed by Moriyasu & Nishiwaki (2009). These can then be repositioned so that values for the three force components can be recorded at 19 different locations on the shoe sole. Good agreement has been found with existing literature, suggesting that it would be advantageous to develop this measurement system further. Loading at the heel was not however measured as magnitudes beyond the rate capacity of the force sensor were anticipated.
**Plantar pressure distribution**

In-shoe plantar pressure measurement has been used to gain a better understanding of the kinetics of a running footstrike. The subjective perception of “comfort” is an essential element of athletic footwear design and has been well correlated to a reduction in peak plantar pressures (Hennig et al. 1996; Shorten 2009). The peak in-shoe pressures observed in eight regions of the foot when walking at 1.8 m/s and running at 3.3 m/s as reported by Chuckpaiwong et al. (2008) are presented in Table 2.1:

<table>
<thead>
<tr>
<th></th>
<th>Walking (kPa)</th>
<th>Running (kPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rearfoot</td>
<td>207.3 ± 37.8</td>
<td>259.8 ± 93.9</td>
</tr>
<tr>
<td>Medial midfoot</td>
<td>109.7 ± 29.6</td>
<td>159.8 ± 39.1</td>
</tr>
<tr>
<td>Lateral midfoot</td>
<td>123.4 ± 26.0</td>
<td>181.9 ± 29.2</td>
</tr>
<tr>
<td>Medial forefoot</td>
<td>202.6 ± 72.4</td>
<td>304.6 ± 118.5</td>
</tr>
<tr>
<td>Middle forefoot</td>
<td>201.2 ± 53.9</td>
<td>295.4 ± 89.3</td>
</tr>
<tr>
<td>Lateral forefoot</td>
<td>175.4 ± 48.4</td>
<td>246.2 ± 71.6</td>
</tr>
<tr>
<td>Hallux</td>
<td>242.9 ± 76.4</td>
<td>303.4 ± 91.8</td>
</tr>
<tr>
<td>Lesser toes</td>
<td>187.9 ± 54.6</td>
<td>260.0 ± 86.7</td>
</tr>
</tbody>
</table>

Table 2.1 - Peak pressure loads in the foot during walking and running. Adapted from Chuckpaiwong et al. (2008).

Hennig & Milani (1995) studied the effect of differing shoe constructions on plantar pressure distributions during rearfoot running. Eight discrete pressure sensors were placed under selected anatomical structures so that the pressures experienced by the plantar surface of the foot could be recorded over the duration of the entire footstrike. The sensors were positioned at the medial and lateral heel, medial and lateral midfoot, under the first, third and fifth metatarsal heads and beneath the hallux.

The authors concede that the use of discrete sensors presents a highly simplified representation of the pressures exerted but allows data to be obtained at specific anatomical landmarks and other areas of interest. Figure 2.15 shows the mean pressure distribution for 22 subjects wearing an athletic shoe with a standard, homogenous midsole and running at 3.3 m/s. Peak pressures are significantly greater than those reported by Chuckpaiwong et al. (2008) but this can be attributed to the more primitive sensor technology used in this earlier study. The vertical GRF signal was used to determine a total contact time of 0.269 s.
Initial contact was made at the lateral heel as most rearfoot runners prepare for impact by adducting the hip and supinating the foot (Cavanagh & Lafontune 1980). Heel pressure remains significantly higher on the lateral side for the first 50 ms as the foot pronates into a neutral position. Midfoot and forefoot loading begins between 20 ms and 30 ms. Cavanagh & Lafontune (1980) reported that the centre of pressure (COP) reached 50% of the shoe length in the anterior direction after 42 ms. This is slightly earlier than Hennig & Milani (1995) observed any significant pressure in the midfoot but can be attributed to the higher running speed used in the first study.

Lateral loading of the midfoot and forefoot was notably higher than on the medial side until about 70 ms, particularly in comparison to a previous study by Hennig & Milani into pressure distributions during bipedal walking (1993). This suggests that in comparison to walking, the increased loads experienced during running act to collapse the longitudinal arch. Between 90 ms and toe-off, the highest loading is seen under the first metatarsal head, suggesting that the first metatarsal is the main load bearing structure during the propulsive phase of gait.
Kinematic studies have reported that the rearfoot is supinated at toe-off but little lateral pressure was seen at the forefoot during this phase. This suggests that a rotation has occurred with the forefoot segment becoming everted relative to the rearfoot at toe-off.

A limitation of centre of pressure calculations is that no pressure may actually be being exerted at the COP. Despite this, COP analysis can still provide valuable data for the analysis of footwear performance. Cavanagh & Lafortune (1980) state that the passive and active force peaks occur when the centre of pressure is at 25 % and 70 % of the shoe’s length respectively. The cushioning properties of these areas are therefore of particular importance. Similarly, Hennig & Milani (1995) report that heel strikers spend the majority of the support phase with the centre of pressure lying between 60 % and 80 % of the length of the shoe. This area is clearly of paramount importance with respect to traction and must provide adequate resistance to slipping. The motion of the centre of pressure is illustrated in Figure 2.16.

![Third-party copyright material removed from electronic version.](image)

**Figure 2.16 - Net force vectors beneath the right shoe during the support phase of gait. Adapted from Cavanagh & Lafortune (1980).**

A further study by Hennig et al. (1996) reported that harder midsole materials initiate higher peak stresses at the heel whilst softer midsoles reduce peak stresses by increasing the contact area. De Wit et al. (1997) also reported that subjects can attenuate loading and reduce peak pressures by reducing sole angle at impact. This improves cushioning performance by increasing the plantar loading surface.

The studies reported here have been selected for discussion because their results are of the greatest relevance to this thesis. Pressure sensor technology has however progressed significantly since many of these studies were performed and is presented in more depth in Section 3.3.3.
2.2.4. Biomechanics of Running Injuries

A wide variety of methods have been employed to gather epidemiology data on running injuries. This has made it difficult to determine the true incidence of injury but studies of over 500 runners have reported annual injury incidence rates of between 37% and 56% (van Mechelen 1992; Walter et al. 1989). Between 50% and 75% of running injuries appear to be overuse injuries with incidence rates varying in the literature from 2.5 to 5.8 injuries per 1000 hours of running (Lysholm & Wiklnader 1987; van Mechelen 1992). Discussion of traumatic injuries is less prevalent in the literature, perhaps because laboratory testing must be performed in a clean, level environment, as opposed to on the irregular, variable surfaces where most running is actually performed. A comparison of different epidemiology studies (James et al. 1978; Clement et al. 1981; Ballas et al. 1997; Bennell & Crossley 1996) is shown in Table 2.2:

<table>
<thead>
<tr>
<th></th>
<th></th>
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<th></th>
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<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>No. of injuries</td>
<td>Runners</td>
<td>Runners</td>
<td>Runners</td>
<td>Runners</td>
<td>Sprinters</td>
</tr>
<tr>
<td>180</td>
<td>1650</td>
<td>860</td>
<td>39</td>
<td>19</td>
<td></td>
</tr>
<tr>
<td>Type of data</td>
<td>Clinic</td>
<td>Clinic</td>
<td>Clinic</td>
<td>Interview</td>
<td>Interview</td>
</tr>
<tr>
<td>Site of injury (%)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee pain</td>
<td>29.0</td>
<td>25.8</td>
<td>13.8</td>
<td>15.0</td>
<td>14.0</td>
</tr>
<tr>
<td>Shin splints - tibial stress syndrome</td>
<td>13.0</td>
<td>13.2</td>
<td>7.8</td>
<td>13.6</td>
<td>5.0</td>
</tr>
<tr>
<td>Achilles tendinitis</td>
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<td>6.0</td>
<td>2.2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plantar fasciitis</td>
<td>7.0</td>
<td>4.7</td>
<td>4.0</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle/foot tendinitis</td>
<td></td>
<td></td>
<td>13.9</td>
<td>6.0</td>
<td></td>
</tr>
<tr>
<td>Stress fracture</td>
<td>6.0</td>
<td>5.8</td>
<td>9.3</td>
<td>25.1</td>
<td>18.0</td>
</tr>
<tr>
<td>Tibial stress fracture</td>
<td></td>
<td>2.6</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Metatarsal stress fracture</td>
<td></td>
<td>3.2</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Iliotibial tract tendinitis</td>
<td>5.0</td>
<td>4.3</td>
<td>3.8</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Patellar tendinitis</td>
<td></td>
<td>4.5</td>
<td>2.2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hamstring strain</td>
<td></td>
<td></td>
<td>5.2</td>
<td>4.3</td>
<td>38.0</td>
</tr>
<tr>
<td>Adductor strain</td>
<td></td>
<td></td>
<td>6.0</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ankle lateral ligament sprain</td>
<td></td>
<td>4.9</td>
<td>8.9</td>
<td>3.0</td>
<td></td>
</tr>
<tr>
<td>Others</td>
<td></td>
<td></td>
<td>9.4</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 2.2 - Common injury sites in running. Adapted from Williams (2000).

A review of the literature on running injuries identifies pronation as a prevalent but controversial theme. “Overpronation” is difficult to explicitly define but has been described by
Hreljac et al. (2000) as a level of pronation that falls outside “normal” physiological limits. This type of excessive pronation has been associated with several common running injuries, including tibial tendinitis, shin splints, Achilles tendinitis and plantar fasciitis (Clement et al. 1984; Gehlsen & Seger 1980; James et al. 1978; Messier & Pittala 1988; Nigg et al. 1984; Vtasalo & Kvist 1983). As a result, pronation control has become a major focus in modern running shoe design (Hilgers et al. 2009).

James et al. (1978) first hypothesised that a link between excessive pronation and injury exists. They believed that this was due to the increased rotational loading that occurs at the knee. Nigg et al. (1984) also speculated that severe pronation could be linked to increased loading of the musculoskeletal system but concluded that conclusive proof was not available at the time.

Hreljac et al. (2000) examined the kinematic differences between injury-free runners and runners who had sustained overuse injuries. A trend, albeit insignificant, was reported with injury-free runners displaying higher angles of supination at initial contact and greater pronation velocity during the footstrike. It was suggested that moderate pronation allows loads to be attenuated over a longer period of time and that a pronated foot is more stable prior to toe-off. They did however note that pronation must end before midstance so that the foot can become a rigid lever during the propulsive phase of gait (Subotnick 1985).

In addition, Hreljac et al. (2000) warned that severe overpronation would increase the torques generated at the ankle and could increase injury occurrence. This is supported by other studies (Clement et al. 1984; James et al. 1978; Nigg et al. 1984). The results of this kinematic study did however contradict some previously reported findings (Gehlsen & Seger 1980; Messier & Pittala 1988; Vtasalo & Kvist 1983) but it was noted that methodological differences existed, primarily that all of these studies had analysed the kinematics of runners who were still injured as opposed to those no longer experiencing pain.

Many associations have also been made between the impact loading and injury but there is little understanding of the specific mechanisms that may cause this (Frederick 1986). Design can have an obvious effect on the shock absorbing properties of a shoe but the apparent adaptations of a runner’s style mean that little correlation is seen between midsole compliance and force platform measurements (Clarke et al. 1983; Nigg 1986). Despite this, Hreljac et al. (2000) concluded that runners who adopt a style characterised by relatively low levels of
impact loading and moderately rapid pronation throughout the footstrike are at a reduced risk of overuse injuries. This is reflected in modern athletic shoe design (Hilgers et al. 2009) despite a lack of conclusive epidemiological or clinical evidence for efficacy of this approach (Nigg & Morlock 1988; Stacoff et al. 2001; Richards 2009; Shorten 2002).

2.3. Running Shoe Anatomy

The main roles of a running shoe are to prevent injury, provide traction with the running surface and optimise performance through the minimisation of energy losses whilst causing minimal discomfort to the wearer (Nigg & Segesser 1992; Lafortune 2008; McPoil 2000). Modern running shoes are designed to reduce the occurrence of injury by attenuating impact forces and restricting potentially dangerous movements, despite the lack of conclusive efficacy data to support this approach (Shorten 2000).

The three primary components of a running shoe are the upper, midsole and outsole. They are shown in Figure 2.17. Adhesives are employed to connect the midsole to the lasted upper superiorly and the outsole inferiorly (Morton-Jones 1986). The function and construction of each of these components is discussed in this section.

![Figure 2.17 - The primary components of a modern running shoe](image)

### 2.3.1. The Upper

The main role of the upper is to hold the foot to the midsole and to provide protection from the external environment (Chase 2009; Promjun & Sahachaisaeree 2012). The upper is traditionally constructed of the vamp which covers the forefoot and toes, and the quarter panels which contain the lacing eyelets and cover the midfoot and heel (McPoil 1988). In modern running shoes however, the vamp and quarter panels are usually made from one
continuous piece of material, commonly nylon or a similarly breathable fabric to ensure that thermal comfort is maintained (Reinschmidt & Nigg 2000). Additional support can be provided by attaching reinforcing fabrics or polymers. For example, most running shoes include a reinforced toe cap which is positioned around the toe box (McPoil 2000).

Rearfoot stability is provided by a heel counter which is made of a stiff material and helps to locate the heel within the shoe and prevent excessive motion (Cavanagh 1980). The heel counter typically has an elongated medial side to assist pronation control (Cheskin 1987). Within the upper and separating the foot from the lasting material and midsole is the sockliner. The sockliner is responsible for absorbing perspiration and providing additional cushioning but is usually removable to create space for orthotic devices (Chase 2009).

2.3.2. The Midsole

The wedge shaped midsole provides cushioning to the foot, attenuating the shock of impacts and reducing the magnitude of localised pressure peaks by distributing the load over a larger area of the plantar surface. It is also designed to prevent movements that could potentially cause injury such as excessive subtalar motion and traumatic inversion/eversion (Shorten 1993). Viscoelastic foams such as ethylene-vinyl acetate (EVA) and polyurethane (PU) are most frequently used for the midsoles of running shoes due to their ability to damp motion and dissipate heat (Khemani 1997). EVA is used more frequently than PU because of its lower density but PU is more durable and has better resistance to compaction (McPoil 2000). Most modern midsoles are actually made up of a variety of materials. Different durometers of EVA and PU are combined to alter a midsole’s flexibility, durability, cushioning and ability to correct non-neutral gaits (Shorten 1993).

One drawback of using viscoelastic foam materials is that the composition of each midsole is unique and that this can lead to problems with consistency in service (Morton-Jones 1986). Designers are constantly developing alternative midsole technologies to help control the motion of the foot and attenuate impact forces and these can come in a variety of mechanical and chemical forms. Two examples that have been well received by consumers are adidas ForMotion™, a decoupled heel unit which reduces eversion velocity and impact loading rates by extending heel contact time (Hilgers et al. 2009) and ASICS GEL, a silicone based midsole.
cushioning system that claims to offer similar cushioning performance to an EVA midsole foam (Asics 2010).

Several manufacturers have also developed midsoles which include or exclusively employ structural thermoplastic polyurethane (TPU) structures. The advantages of TPU structures over traditional foam midsoles include reduced thermal sensitivity, improved material consistency and more consistent performance across the product’s lifespan. Certain TPU midsoles have also displayed cushioning properties comparable to those of a top model EVA running shoe (Aguinaldo & Mahar 2003). Two popular products employing this type of technology, Nike Shox and adidas Bounce™ are shown in Figure 2.18.

![Third-party copyright material removed from electronic version.](image)

*Figure 2.18 - Shoes incorporating structural TPU elements. a) Nike Shox, and b) adidas Bounce™, From Nike (2010), adidas (2010) respectively.*

Some manufacturers wish to increase the torsional rigidity of their footwear with the aim of preventing traumatic injury during extreme loading scenarios. Consequently, the rearfoot and forefoot segments of the midsole are often connected with a rigid midfoot bridge, generally made of plastic or carbon fibre (Chase 2009). However, the concept of torsion is still not fully understood and overly stiff shoes have however been found to impede the natural rotation of the foot and encourage excessive pronation (Stacoff et al. 1991). As a consequence, modern running shoes are commonly designed to protect the foot from excessive loading without restricting its natural freedom of movement (Michel et al. 2009).

Commonly found in trail running shoes, further protection is offered to the plantar surface of the foot by a plastic plate or mesh sandwiched between the midsole and outsole. This component is often called the topplate and acts to shield the foot from any elements protruding from the running surface (Chase 2009).
2.3.3. The Outsole

The main role of the outsole is to protect the midsole from abrasion and to provide traction with the running surface, but it can also influence the cushioning performance of a shoe (Cavanagh 1980). Most modern running shoes utilise outsoles made of rubber for its strength in tension and resistance to abrasion and flex fatigue (Woods 1982) with more aggressive tread patterns employed on shoes designed for off-road running (Chase 2009).

2.3.4. Running Shoe Energetics

Many sports activities involve significant amounts of energy storage and return (Nigg 2009). For example, during tumbling a gymnast uses part of the returned deformation energy of the track to aid performance (McMahon & Greene 1979). Biomechanists have long studied the potential of running shoes to similarly store and return energy so that performance can be enhanced (Frederick et al. 1982).

To dissipate energy at the heelstrike, the rearfoot segment of a high performance running shoe is commonly fitted with a material exhibiting high hysteretic energy losses. In contrast, an elastic material is inserted at the forefoot to minimise energy losses. Running shoe manufacturers frequently claim that this approach enhances performance but Thomson et al. (1999) reported this to be a misconception and that it led to no significant difference in oxygen uptake, heart rate or perceived exertion for the shoes tested. One proposed explanation for this is that there is no current footwear design capable of taking the energy stored from the heel and returning it during the propulsive phase of gait (Nigg 2009).

For an optimal energy return system, the energy must be of a significant magnitude and be returned at the right time, frequency and location (Nigg & Segesser 1992). Current running shoe designs are net energy dissipaters and energy storage and return levels are modest when compared to the strain energy stored and recovered in the lower limbs. Energy storage levels within the shoe are however considered to be sufficient to affect the energetics of the foot and lower limb and future running shoe designs should seek to better exploit this energy in order to improve performance (Nigg 2009; Shorten 1993).
2.4. Finite Element Modelling

The application of computer simulation to dynamic problems has long been recognised as a powerful tool in the study of biomechanics and the industrial design of sports equipment (Hubbard 1993). FE modelling is a powerful and versatile simulation tool that has become an area of significant research interest in recent years. A well developed and validated FE model of the foot-shoe interface would allow researchers to virtually evaluate footwear performance and avoid the variation inherent in human testing (Erdemir et al. 2005). This is particularly useful during the early stages of footwear development when the performance of two prospective footwear designs can be compared under identical loading conditions. Later in the design process it is favourable to perform sensitivity analyses such that the potential effects of real world variation can be investigated.

The published FE models of the human foot and footwear most relevant to this thesis are reviewed in this section such that an area of project novelty can be identified. A more thorough introduction to the fundamental concepts of FE modelling is presented in Section 4.1.2.

2.4.1. Human Foot Models

The footstrike models reported in this thesis were not developed to evaluate the loading that occurs within the human foot itself, but rather to ensure that biomechanically representative loading was applied to a virtual footwear assembly. Conversely, the majority of the human foot models discussed in this section are relatively complex and designed from a medical perspective to investigate the loading that occurs in human foot tissue during gait. Despite this contrast in objectives, the human foot obviously plays a significant role in loading an item of footwear so it was considered necessary to review the available literature on existing FE models of the human foot.

The first 3-D model of the foot to be used for stress analysis was developed by Jacob & Patil (1999). The bones, cartilage and ligaments of the foot were modelled as linearly elastic, homogenous materials with geometries taken from subject X-rays. All material properties were determined from published literature with plantar soft tissue being modelled as a linearly elastic material with a Poisson’s ratio of 0.49. Key muscle forces were applied to simulate the heelstrike, midstance and propulsive phases of walking gait with values taken from published
The highest von Mises stress (21.5 MPa) was found to occur in the lateral metatarsals during push-off. Simulated plantar stress values were in the same order of magnitude as during experimental testing (0.2 - 0.5 MPa).

Giddings et al. (2000) created a 2-D whole foot model to investigate calcaneal loading during walking and running with geometries formulated from a mid-sagittal CT foot scan. As shown in Figure 2.19, the bones of the foot were combined into three bodies - the talus, calcaneus and forefoot, and modelled as linearly elastic, isotropic and homogenous materials. To constrain foot motion, soft tissues including the joint capsules, short and long plantar ligaments and plantar fascia were constructed with tension-only linear truss elements. The talocalcaneal and calcaneocuboidal joints were modelled with a frictionless contact-coupled formulation and experimental ground reaction forces were applied to the model. Externally measured GRFs and high speed cineradiography were used to position the skeletal structures and determine the loads to be applied on the model throughout the footstrike. Maximal calcaneal loading was found to occur at 60 % of the stance phase for running and 70 % for walking. Peak talocalcaneal and calcaneocuboidal loads of 11.1 and 7.9 BW respectively were predicted for running, with these values corresponding to those obtained with a previously reported lower extremity model (Scott & Winter 1990).

Figure 2.19 - A simplified 2-D finite element of the human foot. Adapted from Giddings (2000).

The first 3-D foot model incorporating material non-linearity was developed by Gefen et al. (2000). This approach increased the computational expense of the model but provided a more accurate simulation than previous analyses employing purely linear material models. Six phases of gait were simulated with foot anatomy positioned using Digital Radiographic Fluoroscopy (DRF) imagery. The simplified model included 17 bony elements interconnected
by cartilaginous joints. Plantar soft tissue was included under the calcaneus, 4th and 5th metatarsal bones and all five metatarsal heads.

Ligamentous structures determined to transfer a significant force were also modelled. Bones were modelled as linear, elastic, homogenous and isotropic but ligaments and soft tissues were considered non-linear with the load-deflection relationship defined from uniaxial tensile testing. Loading was applied quasi-statically by adding inertial forces to the body load. The model was validated by comparing model predictions to respective experimental contact stress and contact pressure distributions. This model was the first to utilise digital radiographic fluoroscopy to determine skeletal motion during gait. This novel approach allowed in-shoe bone kinematics to be observed but no information on its reliability was published in this study.

The distribution of stress in the foot that occurs between midstance and toe-off in barefoot walking gait was modelled by Chen et al. (2001). Geometries of the bones, cartilage and soft tissue were determined from computed tomography (CT) data of a male subject with all structures assumed to be homogenous. Linear elastic material models were assigned with parameters adopted from previously reported literature. Sagittal plane kinematics of the foot relative to the floor were determined from the displacement of two markers attached to the lateral malleolus and second metatarsal head and applied to the model. A peak plantar pressure of 1003 kPa was reported under the lateral metatarsal heads at terminal stance, a magnitude which falls within the range found in previously reported studies (e.g. Soames 1985).

The first FE model to include all 28 bony structures of the foot was developed by Cheung & Zhang (2006). Barefoot standing and the midstance phase of gait were simulated in order to develop optimised foot orthoses with the geometry of all foot structures determined from magnetic resonance (MR) images. With the exception of the encapsulated soft tissue which was defined as hyperelastic, all structures were modelled as linearly elastic, isotropic and homogenous. 72 ligaments were modelled with tension-only truss elements and all articulating surfaces were simulated with a frictionless, automated surface to surface contact algorithm. Material properties for biological tissues were selected from literature whilst material properties for the orthotic material were determined after mechanical testing in a Hounsfield material testing machine (Hounsfield, Redhill, UK). The geometry of the model is
shown in Figure 2.20. Experimentally measured ground reaction forces were applied to the plantar surface of the foot and extrinsic muscle forces were applied at the relevant insertion points to simulate balanced standing and the midstance phase of gait. Orthosis shape was found to be more significant than material compliance in reducing plantar pressures. In vivo and cadaveric experimental measurements were used to validate the model.

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Figure 2.20 - a) Surface model, b) FE meshes of the encapsulated soft tissue, and c) bony structures. From Cheung & Zhang (2006).

Ledoux et al. (2008) developed a model of the human foot that included distal and proximal cartilage and was the first to differentiate between the plantar fat pad and other soft tissue. Geometries were determined by applying density segmentation techniques to MR and CT scans. Bony structures were modelled as rigid bodies and all cartilage was defined as linearly elastic. Based on previous research (Ledoux & Blevins 2007), the soft tissue and plantar fat pad was modeled with an Ogden hyperelastic rubber formulation. Neutral balanced standing was simulated by applying compressive loading to the tibia and calcaneus. The model was validated by comparing plantar pressures and joint angles to literature sources and cadaveric test data. The authors reported that all peak plantar pressure values fell within one standard deviation of those reported in the literature but only eleven of eighteen joint angles fell within two standard deviations of those recorded during cadaveric testing.

Balanced standing was simulated by Antunes et al. (2008) with anatomical geometries determined from a series of 482 CT scans. All structures were assigned linearly elastic, isotropic material models except for soft tissue which was modelled with a non-linear, hyperelastic material model. Truss elements were employed to represent the plantar fascia whilst axial connectors were used to model the Achilles tendon. The model was compressed by a vertical force equal to half of one body weight applied to the ground reference node. The
stabilising load exerted by the triceps surae during balanced standing was also applied via the achilles tendon. All material properties were selected from the literature. The loading of the model is demonstrated in Figure 2.21. A maximum plantar pressure of 0.108 MPa was predicted for balanced standing. The authors state that validation is to be performed by comparing plantar pressure predictions with experimental plantar pressure measurements but no further evidence of this procedure has since been reported.

![Third-party copyright material removed from electronic version.](image)

**Figure 2.21 - Modelling the Achilles tendon in an FE model of the human foot.** From Antunes et al. (2008).

The first FE model of the human foot to be coupled to a musculoskeletal model and represent the effects of neuromuscular control was published by Halloran et al. (2010). 2-D foot bone and tissue geometries were generated from a sagittal plane cross section at the second metatarsal. Bones were modelled as rigid and the soft tissue as an Ogden hyperelastic material model based on heelpad indentation tests performed by Erdemir et al. (2006). Bones other than the phalanges were grouped as one rigid segment and joined to the ankle with a hinge joint. The phalangeal bones were modelled as a second rigid segment and were unconstrained throughout the simulation. Ankle joint kinematics were directly coupled between the FE and musculoskeletal models. The direct collocation method (Ackermann & van den Bogert 2010) was used to determine optimal gait kinematics which would minimise peak plantar tissue strains and reduce fatigue whilst tracking normative gait. Convergence to realistic gait patterns was accomplished with a 44 % reduction in peak tissue strain density achieved in the optimised solution. The authors do however state that experimental validation would have to be performed before this methodology could be applied clinically.
A multidomain approach was also reported by Rao et al. (2011) with a 3-D finite element model of the human foot driven with muscle forces determined from electromyographic (EMG) analysis of a walking footstrike. Only the forces of the four major muscles to cross the ankle joint (tibalis anterior, gastrocnemius medialis, gastrocnemius lateralis and soleus) were included but the FE model was able to output a vertical GRF displaying similar magnitudes and timings to that recorded during the experimental trial. Whilst not presented, promising results for the kinematics of the foot were also reported.

2.4.2. Footwear Models

The first known FE analysis of a shod human foot was performed by Nakamura et al. (1981). The foot was modelled with a linearly elastic unified bone structure and non-linearly elastic plantar soft tissue. Ankle joint and Achilles tendon forces were applied to simulate midstance. The Young’s modulus of the shoe sole was varied so that a value which minimises peak compressive and shear stresses in the soft tissue of the foot could be identified. The shoe sole was modelled with both linear and non-linear material models and a range of 0.1-10 MPa for the Young’s modulus was found to provide optimum performance in both cases.

Shiang (1997) developed a 3-D model of a midsole and human heel to investigate the effect of different cushioning systems on plantar pressure. The model was loaded with vertical heel pressure and shear forces derived from experimental testing in order to simulate a heelstrike during walking and running. Non-linear elasticity was defined for the soft tissue of the foot and it was determined that a second-order hyperfoam material model provided the most accurate approximation of midsole behaviour due to the large deformations involved. Introducing a contoured insole on top of the midsole was found to reduce peak plantar stresses which were predictably greater during running than for walking.

The effect the shape and material of the soleplate has on shoe performance was investigated with a dynamic FE model by Oda et al. (2003). A 3-D representation of the foot incorporating bony structures and soft tissue was loaded to simulate a running heelstrike. A modified shoe sole was found to offer 14 % cushioning but stability was reduced by 4 % and increased levels of calcaneal pronation were predicted. The parameters “cushioning” and “stability” are not defined and no information on the methodology utilised to determine material properties or
component geometries is presented. Furthermore, model validation is not discussed by the authors.

Verdejo & Mills (2004) developed a 2-D hyperelastic model to compare the stress distribution in the heel pad during barefoot running and with an EVA midsole. As seen in Figure 2.22, the heel bone was modelled as linearly elastic whilst the heel pad and midsole were modelled as non-linearly elastic. The upper boundary of the calcaneus was kinematically ramped down in order to simulate a heelstrike. Plantar pressure distributions were used to validate the model and it was found that barefoot heel pressure would be double that of the pressure when shod. Material fatigue was also investigated and peak plantar pressures were found to increase 100% after 500 km of running. The validity of the material model was confirmed by comparison to published studies on heel pad deformation although it was conceded that uncertainty in the moduli existed as material viscoelasticity was not included in the model.

![Third-party copyright material removed from electronic version.](image)

*Figure 2.22 - Undeformed mesh used in a study to predict heel pad stress distribution during running. Adapted from Verdejo & Mills (2004).*

Shown in Figure 2.23, Goske et al. (2006) developed a hyperelastic model of a shod human heel which incorporates the heel counter, insole and midsole. The effect of insole conformity, insole thickness and insole material was investigated by comparing predicted plantar pressure distributions. The heel bone was modelled as rigid, the heel counter as linearly elastic and the healpad, insole and midsole were modelled with hyperelastic and hyperfoam material models. Midsole material properties were determined from mechanical compression tests. The model was validated with experimental plantar pressure trials with predicted pressures falling within 3% of those recorded during experimental testing. Conformity of the insole was found to be the most important design variable with fully conforming insoles offering a reduction of 43.3%
in peak plantar pressure in comparison to the barefoot state. Increasing the insole thickness provided a further reduction but peak pressures proved relatively insensitive to midsole material selection.

Figure 2.23 - Finite element model of footwear interacting with the heel. From Goske et al. (2006).

A 3-D non-linear viscoelastic FE model was developed by Even-Tzur (2006) to explore the effects of EVA wear on heelpad stress and strain. The calcaneus was designated as linearly elastic, the heelpad as non-linearly viscoelastic and the midsole EVA foam was modelled with a uniform, linearly viscoelastic material model. A sinusoidal vertical load was applied on the heel bone to simulate a dynamic heelstrike. Three approaches were used to simulate wear - by reducing midsole thickness, increasing the elastic modulus and by decreasing the relaxation time constants. Heelpad stress was found to be most affected by midsole thickness with a 50 % reduction in midsole thickness resulting in a 19 % increase in predicted peak heelpad stress. The model was validated by comparison to experimental plantar pressure tests.

Cheung et al. (2007) developed a quasi-static FE model of the foot-shoe interface with which to study the biomechanical effect of custom-moulded orthotics during gait. Assembled footwear consisting of insole, midsole, outsole and upper was added to the FE human foot model developed previously by Cheung & Zhang (2006). The shoe soles were assigned hyperelastic material models whilst the upper was assumed to be linearly elastic. No information on how material properties were determined was presented. Nine extrinsic musculotendinous forces
were assigned from normalised EMG data to simulate the heelstrike, midstance and push-off phases of gait. Within the bones, the highest von Mises stress was found to occur at the posterior subtalar joint during push-off (30.5MPa). Peak plantar pressure was predicted to occur beneath the metatarsal heads, also during push-off (0.47MPa). No information on model validation was presented. The results are displayed in Figure 2.24:

![Figure 2.24 - Deformed mesh plot and predicted plantar pressure distribution of the foot-shoe interface during simulated heelstrike, midstance, push-off. From Cheung et al. (2007).](image)

**2.5. Opportunity for Original Research**

As evidenced in Section 2.4, a significant amount of research has been performed in order develop a finite element model of the human foot, allowing the internal loading of foot structures to be investigated without the need for costly and invasive in vivo experimentation (Cheung & Nigg 2008). However, the primary aim of this research is to develop a virtual engineering environment with which to evaluate the performance of prospective of footwear designs and thus reduce the industry’s traditional dependence on physical prototyping. The models discussed in Section 2.4 generally have a clear clinical focus and would not be appropriate for the testing of athletic footwear for a variety of reasons. These include being limited to two-dimensional analyses (e.g. Halloran et al. 2010; Verdejo & Mills 2004), the use of quasi-static loading conditions (e.g. Cheung et al. 2007) and the largely simplified boundary conditions that are applied (e.g. Chen et al. 2001; Even-Tzur et al. 2006; Goske et al. 2006).
Of the previously reported FE footstrike analyses reviewed in Section 2.4, the model that could best achieve the goals of this project is that reported by Rao et al. (2011) in that it is a dynamic, 3-D model capable of applying biomechanically representative loading throughout an entire walking footstrike. Despite this, no evidence has been presented for a running footstrike model and the analysis has clearly been developed as a clinical tool for the investigation of foot soft tissue loading. Sissler et al. (2013) reported initial successes when using the model developed by Rao et al. to evaluate footwear performance but agreement between the experimental and predicted GRF is weaker than in the barefoot model and, whilst it is stated that development is ongoing, evidence of a running footstrike model is yet to be presented.

In order to provide an accurate prediction of an item of footwear’s response to loading, a footstrike model would have to contain accurate footwear geometries, an appropriate meshing strategy and material models characteristic of those used in modern athletic footwear. Most importantly, the boundary conditions used in any model must be representative of the complex, multiaxial and dynamic loading applied to footwear structures during a complete human footstrike. The development and evaluation of a number of such models are presented in the subsequent chapters of this thesis.
3. Determining Boundary Conditions

A fundamental component of this thesis was to ensure that the finite element models developed were biomechanically representative of the complex loading that occurs during a shod running footstrike. It was thus decided that irrespective of the modelling methodology adopted, all boundary conditions used throughout the thesis would be determined directly from biomechanical motion capture trials. The data obtained for each trial would also serve a secondary function of being used to validate each of the models developed.

This chapter details the methodology adopted for all biomechanical trials and explains the processes developed for determining the six degree of freedom (DOF) foot segment kinematics and footstrike kinetics for each trial. The procedure used to determine the coefficient of friction (COF) between the laboratory force platform and the shod foot is also presented. Subjects were required to provide informed ethical consent and all testing was performed in accordance with protocol approved by the Loughborough University Ethical Advisory Committee.

3.1. Test Methodology

Biomechanical trials were performed by multiple subjects for a range of different footwear conditions. In order to allow for the development of a comprehensive “boundary condition library” to be used throughout the development of different FE footstrike modelling approaches, multiple sporting actions such as running, sprinting, cutting, jumping and landing were performed in each footwear condition. After a modelling methodology were established and validated, this would allow the performance of a prospective footwear design to be evaluated under a wider range of loading conditions, thus increasing the predictive capacity of the modelling approach.

3.1.1. Subjects

Motion capture trials were performed by two healthy, male subjects with a natural, rearfoot striking running style and no history of lower-limb injury. Rearfoot strikers were selected for testing as only 20% of distance runners were identified as making initial ground contact with their midfoot or forefoot by Kerr et al. (1983). Anthropological information for the two subjects is presented in Table 3.1:
### Table 3.1 - Anthropological data for biomechanical test subjects.

<table>
<thead>
<tr>
<th></th>
<th>Subject A</th>
<th>Subject B</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
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<td>24</td>
</tr>
<tr>
<td>Height (m)</td>
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<td>1.76</td>
</tr>
<tr>
<td>Mass (kg)</td>
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<td>69</td>
</tr>
<tr>
<td>Shoe size (UK)</td>
<td>8.5</td>
<td>8.5</td>
</tr>
</tbody>
</table>

#### 3.1.2. Test Footwear

Biomechanical trials were performed in four different footwear types, each provided by the industrial sponsor of the thesis, adidas. These are shown in Figure 3.1:

![Footwear](image)

**Figure 3.1 - Footwear used in biomechanical testing.** a) Reference Shoe, b) Feather Zero, c) Titan Bounce, and d) Springblade.

The *Reference Shoe* is constructed of simple, wedge-shaped EVA midsole and a traditional laced upper with embedded heel counter. The *Feather Zero* is a light-weight prototype of the running shoe that was eventually released to the market as the *Adizero Feather*. It primarily consists of a split EVA midsole and nylon 12 topplate whilst the upper is an elastane based knitted fabric designed to hold the foot tight to the midsole. The rearfoot section of the upper is reinforced with additional material to increase structural rigidity in this region and act as a heel counter.
Replacing the traditional EVA midsole, the *Titan Bounce* features 12 thermoplastic polyurethane (TPU) structures sandwiched between a polyamide topplate. The upper is constructed of the same elastane based fabric employed by the *Feather Zero* and there is a small region of EVA connected to the underside of the topplate at the forefoot. Finally, the *Springblade* is a novel footwear construction only released to the market in August 2013. Load attenuation is provided by a total of 16 nylon leaf springs which form two topplate segments joined at the midfoot. The topplate is extended proximally at the rearfoot to act as a heel counter and is embedded in a traditional, laced upper. Additionally, all of the shoes tested employ a blown rubber outsole attached to the underside of the midsole or topplate and were fitted with standard sockliners of a suede and potassium hydrogen phthalate (KHP) foam construction.

### 3.1.3. Hardware

Speed of locomotion has been shown to affect whole body kinematics (e.g. Williams 2000) and kinetics (e.g. Keller et al. 1996) so had to be controlled during all appropriate trials to ensure consistent, repeatable results. This was achieved with the use of reflective laser timing gates (Leuze electronic GmbH, Owen, Germany) positioned to calculate the subject’s speed through the capture volume. Running trials were conducted at 4 m/s (± 0.1 m/s) whilst sprint trials were performed at 6 m/s (± 0.1 m/s).

The laser timing gates were connected to a multi-output transistor-transistor logic (TTL) trigger to ensure that the activation of all data acquisition systems occurred simultaneously. This enabled synchronised high speed video (HSV) footage of each trial to be obtained with each of four cameras (Photron, Tokyo, Japan) orientated medially, laterally, anteriorly and posteriorly to the footstrike area. These were set to capture at a rate of 200 frames per second.

### 3.2. Foot Segment Kinematics

#### 3.2.1. Motion Capture

*Selection of hardware*

An objective comparison of the clinical performance characteristics of eight commercially available motion capture systems was performed by Richards (1999). Of these, two systems - CODA (Charnwood Dynamics Ltd., Rothley, UK) and Vicon (Vicon Motion Capture Systems,
Oxford, UK), were available for the experimental trials reported in this thesis. A comparison of their suitability and capability to accurately capture foot segment kinematics was therefore required.

Most motion capture systems can be divided into two broad categories - active marker systems and passive marker systems. Vicon uses passive markers whilst CODA is an active marker system. A limitation of active marker systems such as CODA is that each marker requires a power source which have been shown to influence body kinematics (Figueroa et al. 2003; Macellari 1983; Maynard et al. 2003). Furthermore, the wiring required to attach the drive boxes to the active markers can impede subject motion.

Richards (1999) evaluated both systems by calculating the root mean square (RMS) and maximum errors of each when tracking two markers of a known displacement rotating on a rigid aluminium bar. In nearly all cases, the magnitude of the error reported for CODA was greater than that reported for Vicon although it is conceded that the technology used in both systems will have changed since this study was carried out. As a result of being an active marker system, CODA’s sole strength in comparison to Vicon was that the marker labels did not become confused when passing within close proximity of a third, static marker although this error could be manually corrected in post-processing.

In order to provide realistic boundary conditions for an FE footstrike model, the kinematic data obtained during motion capture trials must be highly accurate and unaffected by the test apparatus. In order to identify relative motion between the foot’s functional segments, a large number of markers would have to be attached to the foot. The volume, mass and potential encumbrance of the CODA drive boxes and attached wiring makes this impractical. Consequently, and as a result of its reported superior accuracy, Vicon was selected as the most suitable system with which to capture foot segment kinematics.

**Motion capture with Vicon**

The Vicon motion capture system utilises a suite of networked cameras and dedicated analysis software which enable the kinematics of a set of passive, retroreflective markers to be captured at up to 1000 fps (Vicon 2007). Each MX camera units consist of a video camera, strobe head unit, lens and optical filter and, dependant on the model selected, has a maximum resolution of either two or four megapixels. A typical camera unit is shown in Figure 3.2a.
The strobe unit illuminates the capture volume by emitting near infrared light at a rate coinciding with the opening and closing of the camera’s shutter. The lens filter removes background light by attenuating wavelengths other than those emitted by the strobe LEDs, allowing the camera lens to form an image on the sensor pane consisting almost exclusively of reflected light. This image is evaluated in greyscale by the MX camera and centroid-fitting algorithms are employed to determine if a patch of reflected light is likely to be a marker. If so, data representing the marker’s spatial orientation is sent to the host computer for further processing.

The performance of the Vicon system is dependent on the accuracy with which the system is calibrated. Calibration is performed in two stages - the dynamic camera calibration stage and the static capture volume calibration stage. During the dynamic calibration stage, the physical position of each camera is calculated based on the motion of a calibration wand. The calibration wand consists of five markers of known displacement attached to two perpendicular axes and can be seen in Figure 3.2b. The optical distortion of each lens and any other non-linearity in the system is calculated and a correction matrix is applied to all subsequent frames of data. The position and orientation of each camera is then calculated using the direct linear transformation photogrammetric technique as described by Chen et al. (1994).

Figure 3.2 - a) Vicon MX camera, and b) Vicon MX calibration wand with retroreflective markers (circled). Adapted from Vicon (2007).

The purpose of the static calibration phase is to define the origin and orientation of the global coordinate system (GCS). This is achieved by placing the calibration wand at the location of the desired origin and allowing the application software to record its position. False markers can
be created by any reflective objects in the capture volume or sources of infrared light other than the Vicon strobe unit. Ideally, these should be covered or removed from the area but it is also possible during calibration to create a “camera mask” which obscures unwanted reflections from the analysis. Once the system has been sufficiently calibrated, motion capture trials can be performed.

The output of all networked cameras is transmitted to the host PC and processed by Nexus 1.x, Vicon’s core motion capture software. The motion of any particular marker may become lost if it cannot be seen by less than a specified number of cameras. Appropriate camera placement can minimise the risk of this, but any gaps in marker trajectory can be filled with an interpolated best fit spline or by matching the motion of an associated marker. This entire procedure allows the three-dimensional linear kinematics of each marker to be determined and output by the system, typically in the form of CSV and C3D files. Motion capture of each trial was performed at 200 fps.

3.2.2. Biomechanical Modelling

Prior to each trial, an experienced biomechanist attached ten markers to the shod left foot of the subject according to the Heidelberg foot measurement method (HFMM) as described by Simon et al. (2006). Anatomical landmarks were identified in accordance with the descriptions detailed in Table 3.2, using guidelines described by van Sint Jan (2007). The HFMM marker set was selected as it allows the foot to be divided into rearfoot, midfoot and forefoot segments and has been successfully employed in a variety of clinical studies (e.g. Dubbeldam et al. 2010; Twomey et al. 2010). Furthermore, it reports levels of accuracy and repeatability (inter-tester, inter-day and inter-trial) in line with alternative multi-segment foot models.
Marker label | Description
---|---
MCL and LCL | Medial and lateral point on the calcaneus defined by heel alignment device as described in Simon et al. (2006)
CCL | Placed dorsal on the calcaneus at the landmark of the insertion of the Achilles tendon
NAV | Placed on the navicular such that in the frontal view the marker axis is seen at 45° to the floor
PMT1 | Joint gap between the first cuneiform and first metatarsal placed such that in the frontal view the marker axis is seen at 45° to the floor
DMT1 | Head of first metatarsal at 45° angle between marker axis and floor
HLX | Midpoint of the distal phalanx of hallux
DMT2 | Head of second metatarsal
DMT5 | Head of fifth metatarsal at 45° angle between marker axis and floor
DMT5 | Tuberositas ossis fifth metatarsal

Table 3.2 - Description of marker placement. From Simon et al. (2006).

A static, standing trial was performed at the beginning of each test session so that each marker could be manually labelled. This allowed the Vicon analysis software to automatically label the entire marker set in all subsequent dynamic trials. The placement of each marker is shown in Table 3.2 and Figure 3.4:

![Subject B performing a static, standing trial prior to dynamic testing.](image)

Figure 3.3 - Subject B performing a static, standing trial prior to dynamic testing.
Figure 3.4 - Marker placement. a) Illustration adapted from Simon et al. (2006), and b) Feather Zero with attached markers.

The trial methodology described allowed the linear kinematics of each marker to be recorded but a biomechanical model separating the foot into a number of distinct segments was required in order to obtain six DOF kinematic boundary conditions with which to perform an FE footstrike analysis.

In order to design an appropriate biomechanical model, it is necessary to have a clear idea of the purpose of the study and required complexity whilst the realism of a model depends upon the validity of the approximations and simplifications used (Oxford Metrics 2002). One approach is to divide the body into segments, each assumed to behave as a perfectly rigid element. This method is computationally cheap but does not consider any deformation of the body tissue that may occur, as can be accounted for with an alternative “wobbling mass” approach (Gruber et al. 1998). The deformation of body segments can be significant when modelling dynamic actions so all results must thus be interpreted with an understanding of this limitation. When developing a model using this “rigid body assumption” the characteristics of the joints linking each rigid segment must also be understood and appropriately defined (Yeadon & King 2008). Consequently, the design of any biomechanical model should begin with a study of human anatomy.

As described previously in Section 2.1.3, the foot is complex structure consisting of numerous compound joints and small bones (Oxford Metrics 2002) and is commonly segmented into
three functional zones - the rearfoot, midfoot and forefoot (Floyd 2006; Gray 2004; Martini 2009). A three segment foot model was thus adopted for use in this thesis. The rigid rearfoot segment encompasses the talus and calcaneus and is separated from the midfoot segment by the navicular, cuboid and cuneiform bones. The midfoot segment is formed of the five metatarsal rays. Finally, the forefoot segment consists of the fourteen phalangeal bones. An illustration of the three functional segments of the foot is shown in Figure 3.5:

![Foot Diagram](image)

**Figure 3.5 - An illustration of the foot showing the rearfoot, midfoot and forefoot segments. Adapted from Van de Graaff (1998).**

In order to define both the position and orientation of each rigid segment, the linear kinematics of three markers (a total of nine coordinates) are required. One approach is to first nominate a chosen marker as the segment origin, whilst the direction to a second marker defines the segment’s primary axis. A second axis is positioned at a right angle to the first, orientated so that all three markers are located on this plane. Finally, a third axis is created perpendicular to this plane, thereby defining a complete origin-and-axes system for the rigid body and, on the condition that no markers are co-linear, describing the attitude and orientation of the segment (Oxford Metrics 2002). Shown in Figure 3.6 are the three markers used to define the motion of each segment in the biomechanical foot model detailed previously in this section. The navicular marker (NAV) is redundant in this model but was retained for all motion capture trials to facilitate the potential introduction of a fourth, tarsal segment in a future modelling approach.
The development of the aforementioned biomechanical foot model has required human anatomy to be approximated as a simplified system of linked rigid bodies and this approach is subject to a number of inherent errors and assumptions (Oxford Metrics 2002). The segments defined do not behave as rigid bodies in reality. The bones themselves are not rigid and each segment contains a number of articulating joints, each allowing relative motion between the bones within the segment. For example, the rearfoot segment contains the subtalar joint - an interaction of the talus with the calcaneus, and the forefoot segment contains ellipsoidal interphalangeal joints, both facilitating some degree of motion within each assumedly rigid segment (Kapandji 1974; Lundberg & Svensson 1993). The foot model was however designed so that the rigid body approximation would be reasonably valid for a range of dynamic sporting actions, particularly linear actions such as running and walking, thereby reducing the amount of error to be expected in the results.

Two other primary sources of error are present when applying markers directly to the skin but these are particularly significant when mounting markers on a shod foot. The first is the degree of uncertainty involved in accurately placing markers on the underlying skeletal structure. This can be minimised by selecting attachment points at which bony landmarks can be more easily identified but can never be removed entirely (Oxford Metrics 2002). The second source of error is relative motion between the marker and the skeletal components. With skin mounted markers, this is as a result of skin movement artefact and a number of modelling approaches have been suggested to correct for this (e.g. Maurel et al. 1998). It does
however remain a concern when attaching markers to the shoe upper due to the compliant nature of the materials typically employed here. Both of these sources of error are likely to be more significant for the Reference Shoe and Springblade than for the Feather Zero and Titan Bounce shoe due to the former shoes’ looser fit and the thicker upper materials employed. The residual associated with fitting a rigid model to a non-rigid system is discussed in Section 3.2.3.

The described foot model was constructed in Visual3D Standard (C-Motion, Inc., Rockville, USA), a biomechanical modelling and analysis software. To enable the model to be constructed for each footwear type, the position of all markers and the orientation of the global coordinate system were imported from the C3D file of the relevant static trial. When defining a segment, it is normally recommended to place the origin at the distal centre of rotation (COR) and define the first axis to coincide with the proximal COR (Oxford Metrics 2002). The reported foot model does not form a continuous kinematic chain and each segment does not have a clear COR, so this advice is not appropriate. It was therefore decided to place the origin of each segment at its posterior centre and orientate the primary axis towards the anterior marker, in the direction of locomotion. The secondary axis is defined at a right angle to the primary axis in the lateral direction and the tertiary axis is orientated normal to this plane in the positive vertical direction. The axes-and-origin systems defined for the three segment foot model can be seen in Figure 3.7.
The rearfoot, midfoot and forefoot segments are each modelled as simplified rigid bodies and must be joined to form a linked segment model. Relative motion of the midfoot relative to rearfoot is facilitated by calcaneocuboidal, talonavicular and tarsometatarsal joints whilst the forefoot interacts with the midfoot at the metatarsophalangeal joint (Gray 2004; Kapandji 1974; Lundberg & Svensson 1993). The rigid segments in a biomechanical model are typically linked with frictionless joints, with adjacent segments sharing a common point or line to form a kinematic chain (Yeadon & King 2008). Due to the number and intricacy of the compound joints in the foot, linking segments in this way would result in an overly complex and impractical model beyond the scope of this thesis (Oxford Metrics 2002). The interactions between each segment were therefore modelled as unconstrained “free” joints, allowing for six DOF relative motion between each segment.

After importing the necessary C3D files, the static model must be assigned to all dynamic trials for that footwear type. Visual3D Standard allows two different mathematical methods to be employed by the user to fit a static model to a dynamic trial - a segment optimisation approach (SOM) (Spoor & Veldpaus 1980), and a global optimisation approach (GOM) (Lu & O’Connor 1999). With the SOM, segment pose is estimated by minimising the marker array deformation from its reference shape in a least squares sense such that the following optimisation problem is solved:
\[
\min f = \sum_{i=1}^{3} (Rx_i + v - y_i)^T (Rx_i + v - y_i)
\]
\[
s.\ t. \ R^T R = I
\]

where \(x_i\) and \(y_i\) are position vectors of marker \(i\) in the marker array in the reference and dynamic frames respectively, \(R\) is the rotation matrix and \(v\) is the translation vector. The \(R^T R = I\) constraint corresponds to rigid body motion and ensures that the transformation is orthogonal (Spoor \\& Veldpaus 1980). The minimal value of \(f\) is the segment residual value and is discussed in Section 3.2.3. The SOM allows for skin movement artefact and other sources of non-rigidity in the model but does not impose any joint constraints between segments as with the GOM (Lu \\& O’Connor, 1999). As the biomechanical model developed for this thesis does not contain any intersegmental constraints, the SOM was considered most appropriate for fitting the static model to each dynamic trial. The linear kinematics of each foot segment could thus be calculated for all trials.

3.2.3. Data Processing and Analysis

Calculating segment rotations

Based on the assumption that each foot segment behaves as a rigid body, the six DOF motion of each segment can be described by the translation vector of the segment origin and a three-dimensional rotation about this point (Spoor \\& Veldpaus 1980). The method for obtaining the translation vector of each segment origin has been described previously but determining the rotation matrix is more complex as it must be calculated from the three dimensional translation of each of the segment’s three markers.

The linear kinematics of each marker can be used to build a 3 x 3 orthogonal transformation matrix. To describe the segment’s position in a way that is easily interpretable, this matrix must then be decomposed into a series of primitive sequential rotations (Herter \\& Lott 1993; Phadke et al. 2011). The matrix can be decomposed with a number of different algorithms but Cardan angle parameterisation is most appropriate for defining models in the chosen FE modelling software (Abaqus 2012d) and is recommended by the International Society of Biomechanics (ISB) (Wu et al. 2002). Angles output in this way correspond to the roll, pitch and yaw about the local coordinate axes system (Stuelpnagel 1964). These calculations can be
performed by Visual3D Standard but the rotation sequence used to parameterise the transformation matrix can significantly affect the Cardan angles calculated. Selection of an appropriate rotation sequence is therefore essential (Senk & Chèze 2006).

The ISB has published recommendations on appropriate rotation sequences to adopt when analysing various joints of the human body (Wu et al. 2002), but, as of November 2013, no specific advice exists for the intrinsic joints of the foot (ISB 2013). Consequently, a study on the Cardan angle amplitudes output when using different rotation sequences was performed to determine which was most appropriate for this thesis. To allow further discussion, the results of this study for the rearfoot segment are shown in Figure 3.8 with X, Y and Z corresponding to rotations in the sagittal (M-L axis), frontal (A-P axis) and transverse (D-V axis) planes respectively. All angles are presented relative to the origin of the GCS.
Figure 3.8 - Comparison of Cardan angles output using different rotation sequences, rearfoot segment. a) Sagittal plane, b) frontal plane, and c) transverse plane.
When assessing a rotation sequence’s appropriateness for calculating accurate Cardan angle amplitudes, two key properties should be considered - amplitude coherence and the occurrence of gimbal lock (Bonnefoy-Mazure et al. 2010). True gimbal lock occurs when the rotation of one axis approaches 90°, bringing two axes into alignment and causing a mathematical indetermination of angle values (Vicon 2007). This occurs very rarely but a second scenario which resembles gimbal lock does occur more frequently. At certain instances, it may be possible for the pose of a body to be described by two different combinations of Cardan angles. Switching from one solution to the other can occur as a result of this ambiguity, resulting in sudden discontinuities in the angle amplitudes calculated. An example of this can be seen for the for the XYX rotation sequence in Figure 3.8b and Figure 3.8c.

Amplitude coherence is understood as the relationship between the calculated angle amplitude and the maximum amount of rotation known to be achievable (Senk and Chèze, 2006). A sequence is considered coherent if the calculated angle amplitude remains within the maximum angular range available in literature. To determine which rotation sequence was most appropriate for this study, rotation sequences which displayed gimbal lock were first excluded. Angle amplitudes were compared to available literature (Floyd 2006; Gray 2004; Kapandji 1974; Lundberg & Svensson 1993) and sequences that did not satisfy the coherence criteria were also excluded. With this method, XYZ and YZX were found to be appropriate rotation sequences for calculating segment rotations in the form of Cardan angle amplitudes. The XYZ rotation sequence was selected for outputting the rotations of each functional foot segment as this approach is recommended for lower extremity analyses by the ISB (Wu et al. 2002) and was supported by further analysis by Sinclair et al. (2012).

**Data filtering**

The segment residual value describes the “goodness of fit” of a static model to a dynamic trial and can be used to assess the validity of the assumption that each segment behaves as a rigid body (C-Motion 2011). A segment residual value of zero indicates that the segment behaved as a genuinely rigid body, with no markers diverging from the defined reference shape. Larger values are indicative of greater discrepancies in the fitting process and are the product of errors introduced by the measurement system and deformation of the assumedly rigid segment. Across all trials, maximum segment residual values were typically calculated to be
2 - 4 mm, with smaller residual values generally observed in the rearfoot segment due to the stiff heel counter employed in this region for all footwear types.

Compared to the size of the segments modelled, the calculated segment residuals are reasonably high and indicate that there are some issues with the rigid body assumption. However, this is to be expected when using markers attached to the shod foot due to relative motion of the foot and footwear and the compliance of the upper material. Further to this, inspection of the linear kinematics of each marker confirmed that skin movement artefact was a significant issue. When defining boundary conditions for an explicit FE simulation, it is important that there are no sudden changes in acceleration between increments as the stress waves induced can lead to noisy or inaccurate solutions (Abaqus 2012e). As such, filtering of the raw kinematic data output for each trial was deemed necessary.

A comparison of commonly used automatic data filtering techniques was performed by Giakas & Baltzopoulos (1997) and power spectrum assessment was found to provide the most acceptable results when analysing gait. Previous studies of foot segment kinematics during running have used visual inspection of marker displacement frequency spectrums to determine appropriate data filters. Dixon et al. (2011), Mündermann et al. (2003) and Queen et al. (2006) all utilised 4th order low-pass Butterworth filters with cut off frequencies of between 8 Hz and 12 Hz. Morio et al. (2009) also used a 4th order low-pass Butterworth filter but chose a cut off frequency of 30 Hz. It was thus decided that the raw kinematic data obtained for each trial would be filtered with a zero lag, bidirectional, 4th order low-pass Butterworth filter. A cut off frequency of 8 Hz was determined from visual inspection of marker displacement frequency spectrums and from analysis with automated data filtering software (GGPSA, Giakas & Ariel 2001). The importance of applying a digital filter to raw displacement data in order to prevent sudden accelerations can be seen in the comparison of the filtered and unfiltered data in Figure 3.9:
Figure 3.9 - A comparison of filtered and unfiltered kinematic data, translation of the rearfoot segment in anteroposterior direction. a) Displacement, b) velocity, and c) acceleration.
3.3. Footstrike Kinetics

3.3.1. Ground Reaction Forces

Triaxial GRFs for each footstrike were obtained with a piezoelectric analogue force platform (Kistler, Winterthur, Switzerland), networked to the Vicon host PC and set to capture at 1000 Hz. The vertical, anteroposterior and mediolateral force components of a typical footstrike for a rearfoot striking runner wearing the *Reference Shoe* are shown in Figure 3.10:

![Figure 3.10 - Typical GRF components for a rearfoot striking runner wearing the Reference Shoe.](image)

3.3.2. Centre of Pressure

The centre of force application can be determined by analysis of the three force components measured at each of the four load cells that a standard force platform contains. Provided that the location of each load cell is known, the reaction moment at each sensor can be calculated and the distance of the COP from the force platform true origin determined (Grabiner et al. 1993).

The location of the COP during a footstrike is an important variable in footwear development (Cavanagh & Lafortune 1980; Hennig & Milani 1995) but a study by Bobbert & Schamhardt (1990) found typical errors in the COP values output of ± 20 mm. Due to the relative age of this study, it was decided to recreate their methodology in the biomechanics lab used for all motion capture trials. This was achieved by positioning calibrated masses on a metal platform which then applied a force vector to a measured location via a stylus. The difference between the COP output and the known location could then be calculated. As seen in Figure 3.11,
similar results to those reported by Bobbert & Schamhardt were observed with larger errors observed at the extremities of the force platform and a maximum error of 18.5 mm. Whilst these errors are systematic as opposed to random, there is no simple and convenient way to for them to be corrected.

A further study was performed by Middleton et al. (1999) and reported similar findings when applying a load at a single point but also investigated the accuracy of COP accuracy when applying distributed loads with two symmetrically placed metal blocks. Designed to better represent human stabilometry testing, the maximum error found with this method was 1.1 mm. Although this indicates improved COP accuracy can be achieved with distributed loading, this figure could be expected to increase for dynamic load cases such as those applied during a running footstrike (Bobbert & Schamhardt 1990).

### 3.3.3. Plantar Pressure

A comparative assessment of the most commonly used pressure measurement devices was performed by Grigioni & Giacomozzi (2010) and showed that the highest accuracy for both dynamic and static pressure tests was obtained with capacitive elastomer based technologies (RMS error < 0.5 %). A basic elastomer based capacitive sensor is formed of two grids of parallel conductive strips orientated in parallel to one another so that a capacitive node is formed at each intersection. The capacitance of each node is determined by the cross-section of the intersection and the distance between each strip. As pressure is applied, the distance
between the strips decreases and a process of calibration can be used to correlate this change in capacitance to pressure. Multiplexing circuitry can then be used to determine the pressure distribution over the entire sensor matrix (Grigioni & Giacomozzi 2010).

An example of a market-leading platform using this technology is the Pedar-x system (Novel, Germany) which allows in-shoe plantar pressures to be recorded at 99 discrete locations for a UK size 8.5 sensor. Up to four insole pads can be used simultaneously to allow dorsal, medial and lateral pressure distributions to also be obtained and the system has been found to provide reliable in-shoe pressure measurements during gait analyses (Boyd et al. 1997; Putti et al. 2007) although it should be noted that both of these studies dealt with walking rather than running. The Pedar-x system was also selected for its relatively long battery life, high resolution of measurement (5 kPa) and low insole thickness of 1.9 mm (Novel 2010) with plantar pressure recorded at 100 Hz for each biomechanical trial.

3.4. Friction

The contact interface of an elastomer such as that used in the outsoles of the footwear tested for this thesis is extremely complex with adhesion, hysteresis and tearing all contributing to the frictional forces (Tabor 1974; Moore 1975). The Coulomb friction model has a number of well-documented limitations when applied to viscoelastic materials including the assumptions that the COF of a contact interface is independent of contact area (Hutchings 1992), slip velocity (Hofstetter et al. 2006) and contact pressure (Moore 1975). It is however a simple and convenient method for characterising the relationship between the normal and frictional forces at a contact interface and is also the least computationally expensive approach used in the selected FE modelling package to define the relationship between contact pressure and equivalent shear stress (Abaqus 2012c).

The dynamics of friction during shod locomotion have been studied extensively (e.g. Redfern et al. 2001; Myung et al. 1992; Perkins 1978) but there is significant variability in the coefficient of friction (COF) values reported (Chang et al. 2001). It was therefore necessary to determine the COF between the force platform in the laboratory used for all trials and the specific rubber outsole material employed in all test footwear. This would enable appropriate contact interactions to be defined for all subsequent FE footstrike models.
3.4.1. Test Methodology

A bespoke traction testing rig was employed to characterise the frictional properties of the test footwear outsole and laboratory floor surface by measuring the frictional reaction force applied when translating a shod last horizontally. The ratio between the recorded horizontal force and user-defined normal load could thus be used to calculate the COF for the contact interface between the two surfaces.

In accordance with Theismann (2012), the COF was determined with a horizontal displacement velocity of 10 mm/s and normal load of 70 kg, selected to match the mass of the subjects used during experimental motion capture trials (see Table 3.1). All testing was performed with the Reference Shoe as it has the simplest geometry of all footwear types used with five trials performed in both the positive (anterior) and negative (posterior) horizontal directions. Each trial was performed for five seconds meaning that a total displacement of 50 mm occurred.

The traction rig used for experimental testing is shown in Figure 3.12:

![Figure 3.12 - Traction test rig used to determine experimental COF.](image)

3.4.2. Results

Typical output from the traction test rig is shown in Figure 3.13. In each trial the calculated COF increases as the shoe begins to slip across the force platform before settling at an approximately constant value once a steady slip velocity has been achieved.
The COF for each trial was determined by averaging the COF calculations output between 4 and 4.5 s for each trial. This time period was selected as a steady slip velocity could be ensured for all trials at this stage of the trial. The average COF value calculated across all trials in both test directions was 0.742 (s.d. = 0.011) with no distinction between static and kinetic COF made to ensure computational efficiency. Despite being a simplistic characterisation of the complex mechanics that occur at the contact interface, this value was used to define shoe-floor contact interaction in all appropriate FE footstrike models.

3.5. Chapter Summary

The methods adopted to obtain biomechanically accurate boundary conditions for the modelling of a shod human footstrike have been reported in this chapter. In order to build a “boundary condition library,” the six DOF kinematics of a three segment foot model, triaxial GRF components, COP location and distribution of in-shoe plantar pressure were all recorded in four different shoe conditions for a variety of different sporting actions. The procedure employed for determining the COF between the ground and the shod foot has also been described.

The use of this data in subsequent FE footstrike analyses ensures that the boundary conditions applied are representative of the complex loading that occurs during a shod human footstrike. The data can also serve a secondary purpose of being used to validate such analyses, allowing
the accuracy of the simulated output to be verified. The FE footstrike models developed with this set of boundary conditions are presented in the following chapters of this thesis.
4. Introduction to Finite Element Analysis

The finite element method (FEM) is a numerical technique used to provide an approximate solution to a system of differential equations which characterise an engineering problem (Pepper & Heinrich 1992). It is generally not possible to obtain an analytical solution to a complex system involving complicated geometries, material properties and loadings. If physical tests are also not possible or impractical FEA can be used as an alternative approach to obtain an acceptable solution (Logan 2011). Driven by the ongoing development of inexpensive and powerful computers, it has become an effective tool used most extensively in solid and structural mechanics (Rao 2005).

This chapter provides an overview of the finite element method, detailing its basic principles and the history of it use. An overview of the core components of an FE model is provided with a preliminary footstrike model used to illustrate some key concepts applicable to all of the models presented in this thesis. FEA concepts which are only applicable to particular modelling approaches are discussed in later chapters as relevant.

4.1. Overview of the Finite Element Method

4.1.1. History

First developed in the 1950s, the finite element method was originally only applied to structural problems but its power and versatility has led to an marked increase in its usage since the 1970s (Pepper & Heinrich 1992). Zienkiewicz and Cheung (1965) were among the first to apply the method to field problems and a significant amount of work on non-linear problems was performed by Oden (1972). As a result of this emerging research area, a number of special purpose FE codes tailored to particular problem types were written. Furthermore, a variety of multidisciplinary commercial software packages were also developed for use across a range of engineering disciplines (Wriggers 2008). The selection of an appropriate FE code of is therefore of significant importance when modelling any engineering problem and is discussed in Section 4.1.2.

4.1.2. Basic Principles

As previously discussed, an FE analysis requires the continuum or solution region to be discretised into a mesh of finite structures commonly referred to as elements. Each element
represents a segment of the physical structure and contains a number of nodal points. Nodes are usually positioned at element boundaries such that adjacent elements are linked by shared nodes. Element properties and boundary conditions are then defined so that a system of equations can be derived in which the unknowns are the nodal values of a field variable such as displacement or temperature. This system of coupled equations must then be solved so that appropriate interpolation functions can be used to determine the variation of the field variable across the continuum (Huebner et al. 2001; Lewis & Ward 1991). For example, when conducting a stress analysis, spatial displacement is the fundamental value that is calculated at each node. Once this is known, the stress and strain in each finite element can be easily calculated from interpolation of the material properties (Abaqus 2012e).

**Numerical methods**

A number of numerical techniques can be employed to solve a system of equations and predict the response of a continuum under loading. If the system describes a static or quasi-static problem, it is best solved implicitly with a set of linear or non-linear algebraic equations. Alternatively, if the problem is dynamic, a set of ordinary differential equations can be used to explicitly determine the value of the unknowns in the most efficient manner (Huebner et al. 2001).

When conducting a stress analysis, an implicit solver works by building a stiffness matrix containing each of the system unknowns and forming a matrix equation of the form

\[ \mathbf{K}\{\mathbf{u}\} = \{\mathbf{f}\} \]

*Equation 4.1*

where \( \mathbf{K} \) is a known stiffness matrix, \( \{\mathbf{u}\} \) is the deformation vector and \( \{\mathbf{f}\} \) is the load vector (Turner et al. 1956; Martin & Carey 1973). The stiffness matrix contains one row and one column for each unknown in the model whilst both the deformation vector and load vector take the form of column matrices (Altair 2008). Linear iterative techniques are then employed to determine the values of the unknowns in the stiffness matrix.

Alternatively, explicit solvers allow transient problems to be modelled by integrating the equations of motion with time. The equilibrium state for any one time increment is determined by kinematically advancing the solution from the previous time increment using the central difference integration operator. In this way, iterative techniques are not required.
Explicit solvers build diagonal element mass matrices at the beginning of each time increment and then determine the triaxial acceleration of each node with Newton’s Second Law (Sun et al. 2000; Abaqus 2012c). The explicit integration rule can thus be written as

\[ \ddot{u}^{(i)} = M^{-1}(F^{(i)} - I^{(i)}) \]

Equation 4.2

where \( \ddot{u} \) is acceleration, \( M \) is the lumped mass matrix, \( F \) is the applied load vector and \( I \) is the internal force vector. The superscript \( (^{(i)}) \) refers to the increment number.

The central difference operator is conditionally stable. The maximum difference in time between two increments is therefore governed by the length of the system’s stable time increment. This can be approximated from the shortest transition time of a dilational wave across any element in the mesh (Abaqus 2012d). Increasing material stiffness and decreasing material density or element size all shorten the stable time increment, multiplying the number of increments required to complete a simulation and thereby increasing computational expense.

Many types of problem can be solved efficiently with either implicit or explicit techniques. However, linear and non-linear static problems, linear dynamic problems and low-speed linear dynamic problems normally favour implicit methods. By comparison, the explicit method better suits non-linear quasi-static simulations and high speed dynamic analyses in which inertial effects are significant (Abaqus 2012e).

The iterative methods employed by implicit solvers allow for relatively long time increments to be used. However, solving each of these increments is computationally expensive. In contrast, explicit methods require much shorter time increments to be used. A higher number of increments are therefore required to reach a solution but each of these is comparatively cheap. When solving complex, non-linear problems, the implicit method must use shorter time increments to prevent divergence and the computational cost increases significantly. In comparison, explicit methods are well suited to such problems as the large number of increments used discourages divergence and solution time is approximately proportional to the number of degrees of freedom in the model (Sun et al. 2000). This is illustrated in Figure 4.1. The smaller increment size adopted by explicit codes also means that they are better suited for modelling brief, transient problems (Abaqus 2012e).
Introduction to Finite Element Analysis

Figure 4.1 - Effect of mesh refinement and increased model complexity on computational cost. Recreated from Abaqus (2012e).

Selection of FE modelling software

Becker (2004) states that there are three main sources of non-linearity in solid mechanics:

- Material non-linearity occurs when materials are loaded beyond their elastic yield point and display elasto-plastic behaviour or when their response to loading is governed by time-dependant effects such as creep or viscoelasticity.

- Geometric non-linearity is introduced to an analysis if calculations are based on a model’s deformed shape as opposed to its original shape. This is important when large deformations occur as this change in geometry affects both kinematic and equilibrium relationships.

- Boundary non-linearity is present in most contact problems and occurs when the displacements of the contacting bodies are not linearly dependant on the applied loads so that the boundary conditions change during the analysis.

As is the case with in many engineering disciplines, all three sources of non-linearity are present in the problems addressed in this thesis. Materials incorporated in footwear designs commonly display elasto-plastic, viscoelastic and strain rate dependent behaviour. Loading experienced during a footstrike is complex, multiaxial and transient and large deformations of both the foot and footwear anatomies are common. To simulate a footstrike as a purely linear interaction between the foot and the ground would be a significant simplification of the problem and therefore result in questionable model output.

Abaqus (Dassault Systèmes, Vélizy-Villacoublay, France) is a of suite software applications for FE analysis and computer-aided engineering originally released in 1978. It was initially
developed to address non-linear physical problems and has an extensive range of elastomeric material capabilities (Abaqus 2012b). As a result, it was deemed an appropriate software package with which to develop the FE models presented in this thesis.

Abaqus includes three core software products that have been used during this thesis - Abaqus/CAE for model pre-processing and visualisation of results, Abaqus/Standard for FE analyses employing implicit integration methods, and Abaqus/Explicit for FE analyses employing an explicit solver. ATOM, an accompanying optimisation module was also employed for the models reported in Chapter 9.

### 4.1.3. Finite Element Modelling with Abaqus

The Abaqus software package provides the user with a graphical user interface (GUI) with which to create a model of a problem. Once pre-processing of the model is complete, an input file is written and submitted to the chosen solver. Modelling functionality not supported within the GUI can be defined in the model via the Keywords Editor or by directly modifying the input file. A numerical analysis of the problem is then performed with the requested outputs written to an output database. These results can then be visualised within the aforementioned GUI or exported for analysis with third-party software.

Accompanying Abaqus documentation (2012a) states that the following information is required to define an FE model in Abaqus:

- **Discretised geometry**: An approximation of the problem geometry comprising of a mesh of discrete, interconnected elements. The greater the mesh density, the more accurate the results, but the results of an analysis converge to a unique solution as mesh density increases. The concept of a mesh convergence study is discussed in Section 4.2.6.

- **Element section properties**: The geometry of some element types is not completely defined by their nodal coordinates. Additional geometric data can be defined as a physical property of an element in order to represent model geometry more completely. For example, 2-D shell elements can have an assigned thickness in order to more accurately model thin 3-D structures.
• **Material data:** The validity of model output is dependent on accurate material models. Assuming appropriate material data are available, Abaqus has the capability to model complex material behaviours relevant to footwear engineering. These include hyperelasticity, viscoelasticity, hysteresis and the Mullins effect.

• **Loads and boundary conditions:** Kinetic loads and kinematic boundary conditions such as concentrated forces, moments, displacements and velocities can be applied to the model. Constraints defining the degrees of freedom available to each node and the interaction between different parts of the model can also be specified.

• **Analysis type:** Abaqus is capable of performing many different types of simulation so the appropriate simulation method must be selected. An example of this is choosing between the implicit and explicit solvers for a dynamic stress analysis.

• **Output requests:** Abaqus can generate a large amount of output so it is necessary to specify which output parameters are required and select an appropriate output frequency such that excessive disk space is not consumed. Output can be separated into two broad categories - field output which is spatially distributed over the entire model, and history output which is data requested at specific points in the model and typically displayed using X-Y plots.

### 4.2. Preliminary Footstrike Modelling

All finite element models must be designed to ensure that the modelling approach allows for useful results to be output in a cost effective manner. For example, simple, computationally cheap models are best employed early in the design process to allow for a large number of design iterations to be investigated. Equally, more sophisticated models should be employed during later design stages when longer solve times can be tolerated in exchange for more accurate predictions of footwear performance.

A simple, preliminary footstrike model developed to predict the response of a midsole to the deformations encountered during a typical running footstrike is presented in this section. This model is also used to illustrate some of the key factors that must be considered when
developing an FE analysis and highlight the sensitivity of model output to each of these parameters.

### 4.2.1. Background

During a typical running footstrike an item of footwear undergoes some degree of bending and torsional loading. The stiffness of the footwear to resist these deformations is of significant interest to a footwear engineer (Chase 2009) and is traditionally evaluated with mechanical tests.

The standard method for evaluating bending stiffness used by the project’s industrial collaborator is to perform a displacement driven three-point bend test at various points along the footwear’s midsole with the reaction force exerted used to compare differing shoe constructions. The test apparatus is shown in Figure 4.2:

![Three point bend test](image)

**Figure 4.2 - Three point bend test used to evaluate footwear bending stiffness.**

Similarly, torsional stiffness is assessed with a bespoke test rig. The shoe is fitted with a deformable split last and is constrained at the rearfoot whilst a defined rotational displacement is applied to the forefoot segment of the footwear. The relationship between angle of displacement and measured reaction moment can thus be used to characterise the torsional stiffness of the footwear through the midfoot segment. The bespoke torsion rig and split last used for testing is shown in Figure 4.3:
Both of these test methods have associated limitations. Firstly, the stiffness data output are sensitive to the positioning of the footwear within each test setup and are therefore not very repeatable. This can make it difficult to make direct comparisons between two pieces of footwear based on the results of such tests. Furthermore, applied load magnitudes and somewhat arbitrary and applied uniaxially and therefore do not necessarily represent the loading that is typically applied to an item of footwear during a footstrike.

Finite element models of the described mechanical tests overcome problems with the repeatability of the results obtained (e.g. Gibbs 2006; Mara 2007) but still apply overly simplistic loading to the footwear assembly. It was therefore decided that an FE model would be developed with which to virtually evaluate the stiffness of an item of footwear to bending and torsion, employing boundary conditions biomechanically representative of the loads applied during a typical footstrike, thus overcoming the aforementioned issues with imprecise and unrepeateable data. This would also allow stiffness to bending and torsion to be evaluated from a single analysis such that the performance of two prospective footwear designs could be compared, without the need for physical prototypes to have been manufactured. Such a tool would be extremely useful during the initial stages of footwear development although sensitivity analyses would have to be performed subsequently in order to fully understand the performance of the modelled footwear.
4.2.2. Model Overview

The model consists of the EVA midsole of the Reference Shoe being deformed by kinematic boundary conditions derived from a motion capture running trial performed by Subject B (see Table 3.1) as reported in Section 3.2 and is solved with the Abaqus/Explicit solver. Rigid plates representing the rearfoot and forefoot segments of the foot are tied to the top surface of the midsole with a kinematic constraint placed on the forefoot plate preventing any change in its pose. No constraint was applied at the midfoot segment of the footwear. The rearfoot plate is driven with angular displacements determined from the biomechanical analysis of a running trial. Rotational stiffness about the dorsoventral (D/V) axis is not typically investigated during the design of athletic footwear but displacement about this axis was included to fully emulate the 3-D kinematics of a footstrike.

Examination of the reaction forces and moments output at the forefoot can then be used to evaluate the behaviour of the midsole under biomechanically representative loading. Figure 4.4 shows the angular displacements applied to the rearfoot plate whilst Figure 4.5 shows the model in both its undeformed state and at the times of maximum torsion and bending. These occur during midstance and just prior to toe-off respectively. Figure 4.6 shows typical model output for the forefoot reaction moment about the A-P and M-L axes which are representative of the midsole’s stiffness in torsion and bending respectively and can be seen to be approximately proportional to the angular displacements applied.

![Angular Displacements Graph](image)

*Figure 4.4 - Rearfoot angular displacements applied in preliminary footstrike model.*
Figure 4.5 - *Reference Shoe* midsole analysis showing rearfoot (blue) and forefoot (red) plates. a) Undeformed, b) maximum torsion (0.125 s), and c) maximum bending (0.22 s).

Figure 4.6 - Typical forefoot reaction moment about A-P (torsion) and M-L (bending) axes.

4.2.3. Material Modelling

All material models described in this chapter were provided by the project’s industrial collaborator. No original work was completed in this area as the focus of the thesis was to improve the validity of the loading conditions applied in FE footstrike models. Despite this, the validity of an FE model is completely dependent on material models which accurately
represent a material’s behaviour under loading and typical procedures undertaken to develop appropriate material models are presented hereafter. Due to the commercial sensitivity of the materials modelled, no material parameters are reported in this thesis.

**EVA**

The results of several studies which determined the material parameters of FE elastomeric foams models from single modes of testing were called into question by Petre et al. (2006). The response of material samples to uniaxial tension, simple compression, planar shear and torsional loading was thus evaluated as these were the modes of deformation deemed most critical to footwear testing. All tests were performed using a 1 kN load cell attached to an Instron 5565 universal testing machine (Instron, Norwood, USA) in accordance with guidelines published by MSC.Software (2010). When forming an EVA sample, heat transferred from the mould causes a solid skin to form at the edge of the material. To allow a homogenous material model to be developed, this skin was removed from all test samples with an abrasive buffer prior to testing. Material density was determined by recording the mass of samples of known dimensions with an electronic balance.

In Abaqus, the hyperfoam material model is best used to characterise elastomeric foams whose porosity permits large volumetric changes (Abaqus 2012a). For simplicity, a first order strain energy density function was selected:

\[
U = \frac{2\mu}{\alpha^2} \left[ \lambda_1^a + \lambda_2^a + \lambda_3^a - 3 + \frac{1}{\beta} (J_el^{-\alpha\beta} - 1) \right]
\]

*Equation 4.3*

where \( U \) is the specific strain energy, \( \lambda_{1-3} \) are the principle stretches, \( J_el \) is the elastic volume ratio and \( \alpha \) and \( \mu \) are material parameters calculated from fitting the model to experimental test data (Abaqus 2012c). The coefficient \( \beta \) determines the degree of compressibility and is related to the Poisson’s ratio, \( \nu \) by the following expression:

\[
\beta = \frac{\nu}{1 - 2\nu}
\]

*Equation 4.4*

Material constants were evaluated using a least squares fit of experimental stress-strain behaviour to the results of single element tests. The validity of these values was then checked
by emulating the four mechanical tests performed and comparing the output to experimental results. As shown in Figure 4.7, the EVA material model developed was found to be 18% too compliant in torsion but provided good agreement in all other modes of deformation. No viscoelastic effects were included in the material model to reduce the computational cost of performing each analysis.

![Figure 4.7 - A comparison of material response to loading in simulated and experimental tests. a) Uniaxial tension, b) simple compression, c) planar shear, and d) torsion.](image)

**Rubber**

Similarly, material parameters for the blown rubber used in shoe outsoles were determined from uniaxial tension, biaxial extension and simple compression tests. A hyperelastic material model was selected as this allowed for non-linear material behaviour and was valid for approximately incompressible materials that exhibit instantaneous elastic response up to large strains (Abaqus 2012a). The material was found to be best represented with a third-order Yeoh strain energy function

\[
U = \sum_{i=1}^{3} C_{i0} (\bar{T}_i - 3)^i + \sum_{i=1}^{3} \frac{1}{D_i} (\bar{J}_{et} - 1)^{2i}
\]

Equation 4.5
where $\bar{I}_1$ is the first deviatoric strain invariant, $f_{el}$ is the elastic volume ratio and $C_{i\theta}$ and $D_i$ are coefficients calculated from input material test data (Abaqus 2012c). Near incompressible material behaviour was ensured by defining a Poisson’s ratio of 0.475.

4.2.4. Boundary Conditions

Kinematic boundary conditions
The rotations of the rearfoot relative to the forefoot were applied to the model via rigid plates connected to the midsole top surface. In Abaqus, the motion of a rigid body is governed by that of its associated rigid body reference node (Abaqus 2012e). These were positioned at the location of each segment origin as defined in the biomechanical model described in Section 3.2.2. The three-dimensional angular displacement amplitudes determined from a motion capture running trial were thus applied to the rearfoot rigid body reference node with the translational degrees of freedom left unconstrained. The rigid body reference node for the forefoot segment was constrained in all six spatial DOFs, restricting its motion completely with an arbitrary mass applied to both rigid body reference nodes to allow the analysis to be submitted to the Abaqus/Explicit solver.

Constraints
Rigid bodies are able to interact with the deformable elements in an analysis through contact (to be discussed in Section 6.1.4) or user-defined nodal connections called constraints. During this thesis, two forms of constraints were used - ties and couplings. A tie constraint acts to fuse together two surfaces with dissimilar meshes - master and slave, for the duration of a simulation whilst a coupling differs slightly in that the nodes on a surface can be constrained to match the motion of a single point for all or a user-defined selection of spatial DOFs. For the analyses presented in this chapter, tie constraints were used to link the rigid rearfoot and forefoot segments to the deformable midsole mesh.

4.2.5. Damping
Both the human body and the materials used in athletic footwear have mechanisms in place which allow for energy to be dissipated when deformed during a footstrike (Nigg 1986; Chase 2009). Quantifying these sources of dissipation is problematic and thus difficult to incorporate in an FE footstrike analysis but data output from a dynamic analysis can be very noisy if no material damping is specified. One method to overcome this is to indiscriminately apply
numerical damping to the modelled components. This approach can result in the footwear being assigned properties that it does not possess in reality but offers a cost effective solution to the aforementioned issues and was considered appropriate at this stage of model development. As such, Rayleigh damping was employed as a convenient abstraction of the energy dissipation mechanisms that exist in the foot and footwear and was specified for all deformable bodies discussed in this thesis (Abaqus 2012a).

Rayleigh damping can be prescribed in Abaqus by defining two damping factors - low frequency mass proportional damping ($\alpha_R$), and high frequency stiffness proportional damping ($\beta_R$). For a given mode $i$, the fraction of critical damping $\xi_i$ can be expressed as follows:

$$\xi_i = \frac{\alpha_R}{2\omega_i} + \frac{\beta_R \omega_i}{2}$$

where $\omega_i$ is the natural frequency at this mode (Abaqus 2012a). The natural frequency of the foot and footwear system is not known so the most appropriate damping parameters were determined from an iterative solution-oriented trial and error procedure performed for each footstrike modelling approach. The optimal combination of Rayleigh damping factors for each problem type was identified by a minimisation of output noise and manageable reduction in the stable time increment. Applying appropriate damping parameters to the model presented in this chapter resulted in a 212 % increase in solve time but the necessity of taking this step to achieve interpretable model output is highlighted in Figure 4.8:
4.2.6. Meshing

Preparation of geometries

The surface-based geometries of all footwear components presented in this thesis were supplied by the project’s industrial collaborator and originated as computer-aided design (CAD) tooling profiles prepared for the manufacture of the various components. It should be noted that small but significant discrepancies existed between the CAD geometries provided and those of the manufactured footwear used during biomechanical testing due to deformations that occurred during the component manufacture and footwear assembly process. This issue is discussed as relevant in subsequent chapters.

In order to prepare a mesh that enabled accurate and computationally inexpensive FE analyses to be performed, some simplification of the component geometries was required. This procedure and all subsequent meshing were performed with the FE pre-processing software package HyperMesh (Altair, Troy, USA). Free edges indicating the existence of surface discontinuities were stitched and duplicate surfaces highlighted by non-manifold edges were removed to create precise and continuous 3-D component volumes. Furthermore, extraneous geometrical features that would have a negligible effect on component performance but drastically increase the required mesh density were removed by suppressing the necessary
feature edges and regenerating any adjacent surfaces. A comparison of a raw, imported component profile and a geometry prepared for meshing can be seen in Figure 4.9:

![Figure 4.9 - A comparison of a) raw imported CAD geometry (solid view), and b) “cleaned” geometry ready for meshing (wireframe view). Reference Shoe forefoot segment.](image)

**Element selection**

Abaqus has a vast library of available elements which provide a powerful set of tools with which to model different problem types. These elements can be characterised with the following criteria - family, degrees of freedom, number of nodes, formulation and order of integration (Abaqus 2012e). One of the primary differences between element families is the geometry that they assume. The elements used most commonly in stress analyses and most relevant to this thesis are shown in Figure 4.10:
Figure 4.10 - Element families.

The continuum family of elements is the most comprehensive and can be used to model the widest variety of components. They are conceptually simple to visualise and are used to model small, three-dimensional blocks of material in a component. Structural elements such as the shell and beam element families allow 3-D parts to be represented with fewer elements than would be required in an equivalent continuum model, thereby reducing computational expense. For example, a surface of shell elements with a defined thickness can efficiently model the geometry of a 3-D component provided its thickness is significantly smaller than its other two dimensions.

The degrees of freedom of an element is the number of fundamental field variables calculated during an analysis to describe its state and configuration. In a stress analysis, the translations of each node are the only degrees of freedom available to continuum elements. The shell and beam element families also have rotational degrees of freedom so that their spatial orientation can be fully defined.

An element’s formulation is the mathematical theory used to define its behaviour. In a stress analysis, the formulation of all elements is based on a Lagrangian formulation. This means that the material associated with an element remains with an element throughout an analysis and cannot flow across element boundaries as with an Eulerian formulation.

Elements within the same family and of the same structure can contain different numbers of nodes and be of different orders of integration. First order reduced elements which only contain nodes at their corners use linear interpolation methods and must assume that the variation of a field variable is distributed linearly across its volume. Alternatively, second order elements contain midside nodes and thus allow a field variable to vary quadratically. Material response is evaluated at each integration point in an element. Abaqus also allows for reduced-
integration elements which contain fewer integration points than their fully-integrated equivalents. A full suite of elements is offered in Abaqus/Standard but the Abaqus/Explicit element library consists almost exclusively of linear reduced integration elements. The concept of element interpolation and reduced integration is illustrated in Table 4.1:

<table>
<thead>
<tr>
<th></th>
<th>Full integration</th>
<th>Reduced integration</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>First order interpolation</strong></td>
<td><img src="image1.png" alt="Diagram" /></td>
<td><img src="image2.png" alt="Diagram" /></td>
</tr>
<tr>
<td><strong>Second order interpolation</strong></td>
<td><img src="image3.png" alt="Diagram" /></td>
<td><img src="image4.png" alt="Diagram" /></td>
</tr>
</tbody>
</table>

![Node and Integration Point Diagram](image5.png)

**Table 4.1 - Element integration and interpolation strategies.**

Fully integrated quadratic elements are generally more accurate than their linear and reduced integration counterparts but are significantly more expensive to solve (Abaqus 2012e). When modelling a problem, an FE engineer must therefore base element selection on the optimal compromise between accuracy and computational expense that can be achieved. A comparison of the performance of linear and quadratic elements for a specific footstrike model is presented in Section 6.1.5.

**Mesh quality**

When using continuum elements to mesh a 3-D geometry it is necessary to choose between brick-shaped hexahedral and pyramid-shaped tetrahedral elements. Hexahedral elements are computationally more efficient (Abaqus 2012e) and were found to provide improved contact pressure and contact shear stress output in a variety of foot and footwear simulations by Tadepalli et al. (2011). However, this study only evaluated two of many relevant field outputs and only those elements available in Abaqus/Standard whilst the majority of analyses performed during this thesis used the Abaqus/Explicit solver due to their dynamic nature. Furthermore, meshing of complex geometries such as those commonly assumed by athletic footwear components can prove difficult with hexahedral elements (Abaqus 2012e). Linear tetrahedral (C3D4) and modified quadratic tetrahedral (C3D10M) elements were therefore
predominantly employed for continuum meshing in the FE models reported in this thesis with C3D4 elements utilised for all models presented in this chapter.

In order to achieve manageable solve times when using the Abaqus/Explicit solver it is necessary to maintain a suitably large stable time increment. Assuming a homogenous material structure within a component, the stable time increment is approximately proportional to the smallest dimension of any element in a mesh. Analysis solve time can thus be minimised by ensuring a consistent and high quality mesh. When meshing a component it is first necessary to create a 2-D mesh on the exterior surfaces of the volume. 2-D mesh quality was ensured by remeshing any triangular elements with interior angles greater than 120° or less than 20° or with any side smaller than half the target element size. 3-D mesh quality was ascertained by ensuring the minimum “tet collapse” ratio (max. height / max. edge length) of any undeformed element was at least 0.23. This prevents the creation of any elements of near-zero volume (Altair 2008).

Mesh density
When performing a finite element analysis, a finer mesh typically results in more accurate model output but is computationally more expensive to solve (Abaqus 2012e). As a mesh is refined it converges upon a unique solution with additional refinement resulting in near identical output. It is therefore good practice to perform a mesh convergence study for each new problem type to ensure that an appropriate mesh density is selected and a satisfactory compromise between solution accuracy and computational expense is achieved (Lewis et al. 2008).

In order to achieve this it is necessary to identify a critical analysis output and determine how this value varies with increasing mesh density. For the model presented in this chapter the critical parameter was identified as the maximum bending moment at the forefoot as this was the mode of deformation which experienced both the largest angular displacements and reaction moments. The mesh convergence criterion was set as change of less than 5 % in this value with the results of the study presented hereafter.
Figure 4.11 - Comparison of six meshes used in convergence study. a) 20 mm, b) 10 mm, c) 5 mm, d) 3.5 mm, e) 2 mm, and f) 1 mm.

Figure 4.12 - Illustration of maximum M-L reaction moment converging on single solution as mesh density increases.

Figure 4.13 - Difference in maximum M-L reaction moment compared to 1 mm target element size.
Plotting a logarithmic trendline through the data presented in Figure 4.12, showed that convergence was predicted to occur at a bending moment 0.633 Nm. As can be seen in Figure 4.13, the mesh with a target element size of 3.5 mm was thus the coarsest to satisfy the convergence criterion of a change in maximum M-L reaction moment of less than 5%. The 3.5 mm mesh had a solve time 221 times quicker than the 1 mm mesh and 17 times quicker than the 2 mm mesh. It was thus considered to provide a satisfactory compromise between solution accuracy and computational expense and was used in all subsequent analyses for this problem type.

4.2.7. Discussion

Some of the fundamental stages in the development of a basic FE footstrike model have been presented in this section but it is also necessary to discuss a major limitation of the model developed. An FE analysis can provide an approximate solution to a problem but model output must be validated against analytical calculations or experimental testing in order to ensure that the output is realistic. This is not easily achievable for the analysis presented in this chapter as the system is too complex to be solved analytically and no experimental equivalent of the simulation exists due to the intricacy of the modes of deformation modelled. All results must therefore be interpreted with an appreciation of the model’s limitations. Despite this, the model developed still provides a useful tool for the virtual comparison of different footwear designs.

To illustrate this, the model presented in this section was altered to include an outsole such that its contribution to the footwear’s bending and torsional stiffness could be evaluated. As can be seen in Figure 4.14, the outsole was attached to the midsole with a kinematic surface-based tie.

Figure 4.14 - Two footwear assemblies analysed. a) Midsole only, and b) midsole and outsole.
Figure 4.15 - Comparison of predicted forefoot reaction moments for analyses with and without the outsole included.

As shown in Figure 4.15, the outsole was predicted to have a significant effect on the footwear’s response to bending and torsional loading. Assuming stiffness to be proportional to the maximum predicted reaction moment, adding the outsole to the footwear assembly would increase bending and torsional stiffness by 115% and 64% respectively.

As discussed previously, there are a number of limitations to this modelling approach, primarily the lack of data against which to validate predicted model output and the lack of constraint applied at the footwear midfoot. Despite this, the model has been shown to apply biomechanically representative and repeatable dynamic loading to a footwear assembly, without requiring a physical prototype to have been manufactured. The approach adopted is computationally cheap and overcomes a number of the shortcomings of the mechanical test methods discussed in Section 4.2.1 so could serve as a useful tool in the early development of a new footwear product.

4.3. Chapter Summary

The fundamental principles of finite element analysis and the components that make up an FE model were discussed in this chapter with justification for the choice of FE modelling software provided. A simple footstrike model was used to illustrate a number of key procedures that are required in the development of any FE analysis. These included the selection of appropriate material models and damping parameters, the application of representative loads
and boundary conditions and a meshing strategy that provides an accurate but computationally efficient solution to an analysis.

The preliminary FE model presented was driven with 3-D angular displacements biomechanically representative of a running footstrike and exhibited some clear advantages over the traditional mechanical testing of footwear. The predictive capacity of the model was demonstrated by comparing predicted model output for two different footwear constructions but validation of these outputs was not possible and the results must thus be interpreted with an understanding of this limitation. The development and evaluation of more complex and thoroughly validated FE footstrike analyses are presented in the subsequent chapters of this thesis. To aid comprehension of the various boundary conditions, loads, constraints and contact interactions employed in each modelling approach, simplified 2-D diagrams of each method are presented here in Figure 4.16, Figure 4.17, Figure 4.18 and Figure 4.19. Each diagram represents a model of a running footstrike performed in the Reference Shoe.

**Pedobarographic Model (Chapter 5)**

---

**Key**
- Kinematic boundary condition
- Kinetic load condition
- Reference point
- Kinematic constraint
- Contact interaction

![Pedobarographic Model](image)

Figure 4.16 - A 2-D exploded diagram of the Pedobarographic footstrike model presented in Chapter 5.
**Kinematic Model (Chapter 6)**

![Kinematic Model Diagram]

- **Key**
  - Kinematic boundary condition
  - Kinetic load condition
  - Reference point
  - Kinematic constraint
  - Contact interaction

**Figure 4.17** - A 2-D exploded diagram of the Kinematic footstrike model presented in Chapter 6.

---

**Combined Load Model (Chapter 7)**

![Combined Load Model Diagram]

- **Key**
  - Kinematic boundary condition
  - Kinetic load condition
  - Reference point
  - Kinematic constraint
  - Contact interaction

**Figure 4.18** - A 2-D exploded diagram of the Combined Load footstrike model presented in Chapter 7.
**Figure 4.19** - A 2-D exploded diagram of the Kinetic footstrike model presented in Chapter 8.
5. Pedobarographic Footstrike Modelling

The method described in Section 3.3, provided a means of recording the triaxial GRF components and distribution of plantar pressure exhibited during a shod human footstrike for a range of different footwear conditions. This chapter presents and evaluates an FE footstrike model which uses location specific pedobarographic force profiles determined from these trials to deform a virtual footwear assembly and thus allow its response to loading to be evaluated.

Initial model development was performed with the Reference Shoe before the modelling approach was validated by comparing the predicted deformations of the Titan Bounce TPU structures under a more complex loading scenario to digitised HSV footage. Finally, the predictive capacity of the model was evaluated by comparing the predicted deformation of the TPU structures output for a more compliant material to experimental trial data unused in the determination of model input loading conditions.

5.1. Preliminary Modelling

5.1.1. Model Overview

The Pedobarographic footstrike model was developed to provide a simple and computationally cheap tool with which to investigate footwear performance during the early stages of product development. Initial development of the modelling approach was performed with the Reference Shoe as it has the simplest geometry of any of the footwear types used during experimental testing. The same EVA and rubber material models were used as detailed in Section 4.2.3 with linear tetrahedral C3D4 elements used for meshing the midsole and outsole geometries. Vertical loading was applied to the top surface of the midsole via 10 discrete rigid surfaces representing the plantar aspect of the human foot, with each assigned an independent loading profile. The geometries of these surfaces were taken from a 3-D scan of a foot prosthesis (Otto Bock, Duderstadt, Germany) and meshed with rigid triangular R3D3 elements.
**Plantar mask**

As discussed in Section 3.3.3, the Pedar-x pressure measurement system was used to record the distribution of plantar loading across the 99 sensors contained within a standard insole. In order for these data to be analysed and used to provide loading conditions in a subsequent FE footstrike model it was necessary to divide the plantar aspect of the foot into a number of discrete sub-regions. These are commonly based around anatomical landmarks such as the eight segment approach adopted by Bontrager et al. (1997) or the five segment mask preferred by Akins et al. (2011). In order to ensure straightforward load calculations, it was decided that a simple 10 segment mask would be created for this study with the foot split into five equally sized sections along its longitudinal axis, each containing a medial and lateral compartment. A diagram of the segmentation approach used to divide the 99 insole sensors into 10 discrete regions is shown in Figure 5.2:

*Figure 5.2 - 10 segment plantar foot mask.*
**Constraints and loading conditions**

A link between the rigid foot segments and the top surface of the midsole was assigned by defining tie constraints such that all exterior midsole nodes lying within 5 mm of a foot segment master surface were constrained. Initial modelling indicated that excessive local loading would occur in the footwear under the interfaces between the foot segments. As seen in Figure 5.3, this necessitated tie constraints to also be defined between adjacent foot segment surfaces to prevent the occurrence of intersegmental separation. Whilst necessary, this resulted in the entire foot behaving as a single rigid plate, thus reducing the accuracy of the local loads applied. It was however considered a suitable solution at this stage of model development.

![Figure 5.3 - Highlighted nodes tied to prevent separation of plantar foot segments.](image)

Vertical loads were applied to the individual foot segments with concentrated forces distributed across all nodes contained in each plantar segment. Load amplitudes were determined from analysis of an experimental running trial performed by Subject B (see Table 3.1) in the Reference Shoe by distributing the vertical GRF measured by the force platform proportionally to the amount of plantar pressure measured over each foot segment at any particular moment during the footstrike. The concentrated forces exerted upon each plantar foot segment can be seen in Figure 5.4. Finally, all nodes on the bottom surface of the outsole were subjected to a full six DOF kinematic constraint such that no translation or rotation was possible. The analysis was submitted to the Abaqus/Explicit solver.
Figure 5.4 - Concentrated force amplitudes applied to each plantar foot segment.

5.1.2. Results and Discussion

As is to be expected, very good agreement can be seen in Figure 5.5 between the magnitude of the vertical GRF recorded in the experimental trial and that output with the FE model. A maximum discrepancy of 141 N was observed at the passive force peak with an average discrepancy of 27 N recorded across the duration of the footstrike (1.6 % of the experimental peak). A discussion on the possible reasons for this discrepancy is presented for the modelling methodology as a whole in Section 5.2.7.
Figure 5.5 - A comparison of experimental and simulated vertical GRFs.

Figure 5.6 shows the predicted von Mises stress in the footwear at various stages of the footstrike with von Mises stress (also known as the equivalent tensile stress) calculated from the three principle stresses such that

\[ \sigma_v = \sqrt{\frac{(\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_1 - \sigma_3)^2}{2}} \]

*Equation 5.1*

where \( \sigma_v \) is the von Mises stress and \( \sigma_1, \sigma_2 \) and \( \sigma_3 \) are the principal stresses. Abaqus computes the von Mises stress at each node based on the method of Segalman et al. (2000) for calculating the RMS von Mises stresses for linear structures in a randomly vibrating environment.
Figure 5.6 - Von Mises stress in footwear assembly at specific time intervals through a footstrike.

The predicted distribution of stress in the footwear components is logical with high localised stresses occurring in the rearfoot during the passive force peak and more evenly distributed loading predicted in the forefoot during the propulsive phase of gait, particularly under the first metatarsal head. However, it should also be noted that this modelling approach only applies normal loading to the footwear assembly. Loading that occurs due to bending and torsion as discussed in Section 4.2.2 is not represented in the analysis.

Furthermore, unrealistically high levels of stress were seen at the very anterior edge of the footwear throughout the footstrike due to the reduced thickness of the midsole in this area but, despite these limitations, the loading of the model was considered to be generally realistic. Validation of the distribution of predicted loading could however not be achieved so it was decided that a different shoe would be evaluated under a similar, but more complex loading scenario.
5.2. Triaxial Loading

A similar model with loading conditions determined from pedobarographic data of a trial performed by Subject A (see Table 3.1) in the *Titan Bounce* footwear was developed which included the anteroposterior and mediolateral components of loading, in addition to the vertical loads applied in the model described in Section 5.1. The *Titan Bounce* was selected as the deformation of the individual TPU structures was considered far simpler to quantify than that of a traditional EVA midsole, thus allowing the footwear’s response to loading to be more easily validated. Furthermore, each rigid foot segment was sub-divided into a medial and lateral section such that the plantar aspect of the foot was now discretised into 20 rather than 10 sub-regions. This was performed to provide a higher spatial resolution and thus more accurate distribution of plantar loading. All footwear components other than the sockliner and upper were included in the model with the undeformed assembly shown in Figure 5.7:

![Figure 5.7 - 20 segment Pedobarographic model for applying triaxial loading to the Titan Bounce.](image)

5.2.1. Determining Load Conditions

Force platforms and pressure insoles are the two technologies commonly employed to measure footstrike kinetics but both impose constraints on the complexity of analysis possible. Pressure insoles can only measure the normal component of the forces applied whilst force platforms cannot determine the distribution of the force components exerted across the foot. Liedtke et al. (2007) proposed footwear instrumented with dual force transducers to overcome these limitations but an alternative approach is to estimate the triaxial loads exerted on the sub-areas of the foot by combining the data output by a force platform and a pressure insole
or mat. This approach allows the foot to be discretised into a far greater number of regions and has been used in a number of reported studies (e.g. Abuzzahab et al. 1997; Giacomozzi et al. 2006; Giacomozzi & Macellari 1997; Giacomozzi et al. 2000; MacWilliams et al. 2003; Saraswat et al. 2010).

The method is based on an assumption of proportionality between the global GRF components measured by a force platform and the local distribution of plantar pressure as recorded simultaneously by a pressure mat. Bruening et al. (2010) evaluated this underlying “proportionality assumption” finding peak absolute errors for the shear forces of up to 12 % BW during normal gait and thus concluded that the method could lead to the some loss of information on foot function. Despite this assessment, and in the absence of a suitable alternative, it was decided that this method would allow largely representative triaxial loading conditions for the 20 segments of the foot to be determined.

5.2.2. Material Modelling

The Titan Bounce footwear model contains a number of different components and required the use of four different material models - EVA for the central forefoot segment, blown rubber for the outsole and TPU for the 12 distinct structures linked by a polyamide topplate. The EVA and rubber material models employed in the Titan Bounce model were as described in Section 4.2.3 whilst the material models developed for the TPU structures and polyamide topplate are reported hereafter:

**Thermoplastic polyurethane (TPU)**

The response of the TPU material to loading was evaluated with uniaxial tensile, planar shear and equibiaxial tensile tests and was found after extensive testing by the project’s industrial collaborator to be best characterised by a hyperelastic material model employing a first order Mooney-Rivlin strain energy function such that

\[
U = C_{10}(\bar{I}_1 - 3) + C_{01}(\bar{I}_2 - 3)
\]

Equation 5.2

where \(U\) is the specific strain energy, \(C_{10}\) and \(C_{01}\) are material parameters and \(\bar{I}_1\) and \(\bar{I}_2\) are the first and second deviatoric strain invariants (Abaqus 2012a). A comparison of the material’s response to experimental and simulated loading is shown in Section 5.3.1.
**Polyamide**

The polyamide used in the topplate of the *Titan Bounce* was assigned an isotropic linear elastic material model with the values for the Young’s modulus and Poisson’s ratio taken from a data sheet provided by the material’s supplier (Evonik Industries, Essen, Germany). A linear elastic model was considered sufficient to accommodate the relatively small strains (less than 5%) typically experience by the topplate during this type of analysis.

### 5.2.3. Element Selection

In order to ensure the optimum compromise between solve time and model accuracy, the outsole, topplate and EVA forefoot segment geometries were meshed with linear tetrahedral C3D4 elements. The deformation of the TPU structures during the simulated footstrike was the primary output of interest when developing this model so they were assigned modified second-order tetrahedral C3D10M elements to ensure an accurate response to loading. Modified continuum elements are typically more computationally expensive than their unmodified counterparts but are the only second-order tetrahedral elements available when using the Abaqus/Explicit solver.

### 5.2.4. Output Requests

Connector elements can be used in Abaqus to link two nodes, specify the mechanical relationship that exists between them and limit the degrees of freedom available for relative motion. As shown in Figure 5.8, unconstrained axial connectors were created across each TPU structure so that the relative displacement of the two attached nodes could be requested and the deformation of each TPU structure easily output.

![Axial connector elements](image)

*Figure 5.8 - Titan Bounce model with connector elements defined on each TPU structure*
A script written in the Python programming language was used to automatically write all history outputs contained within an output database to a CSV file. This was then imported into a pre-formatted spreadsheet to generate a standardised report file for each analysis and allow for a speedy comparison of footwear models to be made.

5.2.5. Mass scaling

As was discussed in Section 4.1.2, the solve time for an analysis is proportional to the stable time increment which can be approximated from the shortest transition time of a dilational wave across any element in the mesh (Abaqus 2012d). Mass scaling can be employed to reduce the solve time for an analysis by adding mass to any elements that cause the stable time increment to fall below a user-defined threshold. It must however be used with caution as this added mass can unduly change the dynamics of a system if defined inappropriately. An appropriate value for the desired stable time increment must therefore be selected for each problem type to ensure a satisfactory compromise between accuracy and computational efficiency.

The complex geometry of the Titan Bounce necessitated the use of a relatively fine mesh and this, coupled to the stiff materials used in the assembly meant that the initial time increment for the analysis reported in this chapter was small, approximately 0.1 µs. Furthermore, the assembled model was very large, containing approximately 270 000 nodes and required nine hours to solve, even when employing a computer fitted with eight parallel processors. A solve time this large is clearly not ideal for footwear development and reduces the practicality of the model as an industrial design tool. As a result, variable mass scaling was defined for all elements in the analysis to ensure that the stable time increment never fell below 0.3 µs, thus reducing the solve time by a factor of three. A comparison of model output from analyses with and without mass scaling defined is presented hereafter.

5.2.6. Results and Validation

Mass scaling

Defining mass scaling in the model resulted in a 5 % increase in the peak value output for the total strain energy and a 4.4 % increase in the maximum deformation of the lateral posterior TPU structure. This loss of accuracy was considered acceptable in exchange for such a significant reduction in analysis solve time so the mass scaling definition as reported in
Section 5.2.5 was maintained in all subsequent analyses of this type, including those used to output the results discussed in this remainder of this chapter.

**Kinetics**

As was reported in Section 5.1.2 for the Reference Shoe model, good agreement was seen between the experimental and simulated GRF magnitudes for all three components although it is clear that similar outputs to those assigned as the model input loading conditions are to be expected. Maximum errors of 108 N, 58 N and 186 N were observed for the anteroposterior, mediolateral and vertical GRF components respectively with average errors over the entire footstrike of 20 N, 12 N and 43 N calculated in comparison to the experimental trial.

![Graph showing GRF components](image)

**Figure 5.9 - A comparison of experimental and simulated GRF components.**

**Deformation of TPU structures**

The strain predicted to occur in the TPU structures is shown in Figure 5.10 but, whilst very useful for evaluating the design of the footwear, it was not possible to validate this output against data recorded from the same experimental trial due to a failure of the HSV cameras at the time of testing. The validity of the loading applied with the FE footstrike model was therefore examined by comparing the predicted deformation of the TPU structures to data obtained by digitising HSV footage of five experimental running trials performed by the same subject in the Titan Bounce footwear at a later date. This was achieved by manually tracking circular black marks drawn on to the TPU structures to match the locations of the nodes linked with connector elements in the FE model (see Section 5.2.4) using commercially available
image analysis software (Image-Pro, Media Cybernetics, Rockville, USA). The process of manually tracking each mark had an estimated error of ± 0.5 mm for each frame. An illustration of this process and results of the analysis are presented hereafter:

![Image](image.png)

**Figure 5.10 - Titan Bounce** under triaxial pedobarographic loading with maximum principal logarithmic strain visualised for TPU structures only. a) Passive force peak, medial view, b) passive force peak, lateral view, c) active force peak, medial view, and d) active force peak, lateral view.
Figure 5.11 - Digitisation of HSV footage. a) Heelstrike, b) mid stance, and c) toe-off.

Figure 5.12 - A comparison of experimental and simulated values for maximum deformation of the Titan Bounce TPU structures during a running footstrike.
Figure 5.13 - A comparison of experimental and simulated values for maximum deformation of the Titan Bounce TPU structures. Error bars represent one standard deviation.

Figure 5.14 - A comparison of experimental and simulated deformation of the Titan Bounce TPU structures. a) L1, b) M1, c) L3, d) M3, e) L5, and f) M5.
Figure 5.12 and Figure 5.13 show that, whilst generally underestimating the maximum deformations of each TPU structure, a similar trend was predicted by the FE model to that measured from the experimental trials with the maximum deformations highest at the rearfoot structures and decreasing towards the forefoot. The average maximum deformation of the 12 TPU structures was measured as 2.45 mm from the digitised HSV footage and predicted as 1.95 mm by the FE model with an average absolute residual of 0.88 mm.

Figure 5.14 shows a comparison of experimental and simulated deformations for six of the 12 TPU structures. It is apparent that the deformation profiles from the FE model are largely representative of those that were measured from the five experimental trials but that the modelling approach is unable to replicate highly dynamic loading as seen at the lateral rearfoot (Figure 5.14a). It is also evident that the general issue of the FE model underestimating the deformations that occur in the TPU structures is particularly significant at the medial forefoot.

5.2.7. Discussion

It is clear that the footstrike modelling methodology discussed in this section was able to apply triaxial load magnitudes to an FE footwear assembly largely representative of those that occur in an experimental trial. However, despite displaying generally realistic trends, the model was found to underpredict the maximum deformations of footwear structures measured in HSV footage. There are a number of possible reasons for this underestimation which are discussed hereafter.

The most obvious issue is that there was significant error associated with digitising the HSV footage of the five footstrike trials. This was estimated as ± 0.5 mm and this error, coupled with the variation that naturally occurs between trials, meant that standard deviations for maximum structure deformation of up to 1.2 mm were observed. When determining the maximum deformation of an individual TPU structure it is clear that the error associated with this measurement would lead to an overestimation of the true value.

The material models employed in the analysis were a simplistic representation of the complex behaviour that each material actually exhibited in response to loading. For example, the TPU structures modelled had no defined viscoelastic properties and thus no means for energy dissipation. The development of appropriate material models was not a focus of this thesis but it is clear that the limitations of the models used would affect the accuracy of model output.
There is also a degree of uncertainty in the representativeness of the loading conditions used in the analysis. Firstly, the load conditions were calculated for a single trial but model output was compared to average values calculated across five trials. Small variations in the variables that would affect the magnitude and distribution of loading such as peak GRF and sole angle at heelstrike were observed between the five trials and it clear that this would affect the pedobarographic loads exerted. Furthermore, the validity of the “proportionality assumption” used to determine triaxial load conditions has been questioned for normal gait (Bruening et al. 2010) and no analysis of its validity in running studies has been performed.

The accuracy of the loads calculated was also dependent on the fit of the pressure insole as it was difficult to consistently locate the insole within the forefoot segment of the footwear due to the tight-fitting nature of the elastane upper. Similar issues were encountered by Boyd et al. (1997) with poor reliability of pressure measurement reported at the hallux, an area which demonstrated comparably poor results for the Pedobarographic model presented in this section.

The interface between the footwear and the segments representing the human foot is another area that contained a number of significant simplifications. The geometry of the foot segments used in the analysis was taken from a prosthesis as opposed to a human foot and demonstrates a poor fit with the geometry of the footwear, particularly at the problematic medial forefoot region. Whilst not investigated, it is hypothesised that improved results could be achieved with a deformable, well-fitting foot geometry as opposed to the rigid segments used in the current methodology.

Finally, the exclusion of the sockliner from the FE model could have resulted in reduced loading of the forefoot TPU structures (M3-6 and L3-6). As shown in Figure 5.15, the sockliner overhangs the topplate in the Titan Bounce assembly, allowing loading to be applied directly to the TPU structures. The omission of this component from the analysis is another potential reason for the model’s underestimation of the maximum deformations experienced by the TPU structures.
Despite the reported limitations of the methodology employed, the results predicted with the analysis show similarities to those measured from biomechanical trials with comparable trends in the deformation of the footwear’s TPU structures observed. Due to equipment failure, the model was not validated against deformation data from the same trial used to determine the input load conditions. Whilst not ideal, the model was still considered suitable for use as a comparative tool, to evaluate the stresses and strains that are predicted to occur in prospective footwear designs. The predictive capacity of the model was investigated in the subsequently reported case study.

5.3. Case Study: *Titan Bounce* TPU Structure Material

5.3.1. Overview

The FE footstrike model discussed in Section 5.2 has been shown to demonstrate a similar response to loading as that measured during an experimental trial performed in the modelled footwear. The aforementioned process of evaluating the output against an experimental trial and thus validate the modelling methodology is necessary to ensure that an analyst can have confidence in the results predicted. The ultimate purpose of developing a model is not to replicate an experimental trial however, but rather to predict footwear performance without the need for experimental testing to be performed. A well-validated, predictive model thus negates the need for physical prototypes to be manufactured.

The predictive capacity of the Pedobarographic model was investigated by comparing data from five experimental running trials to model output. This was performed for two differing types of footwear - the standard *Titan Bounce* as presented in Section 5.2, and an unreleased model with more compliant TPU structures. Other than the TPU material models used, the
two analyses were identical with loading conditions determined from a single trial performed in the standard *Titan Bounce* construction.

As discussed previously in Section 5.2.2, both TPU variants were represented with hyperelastic material models employing a first order Mooney-Rivlin strain energy function. The response to uniaxial tensile loading for the two material models is presented in Figure 5.16 with the results of two analyses reported thereafter.

![Figure 5.16 - TPU material models used in Titan Bounce assembly to evaluate the predictive capacity of the Pedobarographic modelling approach.](image)

### 5.3.2. Results

When switching to the compliant TPU material, maximum deformation was increased for all TPU structures in both the experimental trials and FE analyses, with higher levels of strain predicted (see Figure 5.17). However, the increase in deformation was significantly greater in the experimental trials than predicted with the FE analysis with the extent of this underprediction particularly noticeable at the rearfoot TPU structures. The maximum deformation of each TPU structure increased by an average of 105 % in the experimental trial but only 43 % between the two FE analyses. The issues with the Pedobarographic modelling approach discussed in Section 5.2.7 obviously apply in both cases but the additional limitations of using the model predictively are considered here in Section 5.3.3.
Figure 5.17 - Titan Bounce assembly with standard and compliant TPU structures. Maximum principal logarithmic strain visualised for TPU structures only. a) Passive force peak, medial view, b) passive force peak, lateral view, c) active force peak, medial view, and d) active force peak, lateral view.
5.3.3. Discussion

The advantages of using an FE footstrike model to predict the performance of a prospective footwear design are clear but one must interpret the results of any such analysis with an appreciation of the limitations of such an approach. The five experimental trials performed in the Titan Bounce construction which employed a more compliant TPU material were used to evaluate the validity of the deformations predicted but were not used to determine any model inputs. When using this approach, the assumption must therefore be made that the loads applied to the footwear are identical between footwear conditions. The validity of this assumption is examined hereafter with all statistical comparisons made using a two-tailed unpaired Student’s t-test.
Predicted model output was compared to digitised HSV footage of five running trials performed with the compliant *Titan Bounce* footwear and, as reported in numerous studies (Clarke et al. 1983; e.g. Nigg et al. 1987; Kaelin et al. 1985), no statistically significant change in the magnitude of either force peak was observed between these trials and those completed in the standard *Titan Bounce* construction. Modulation of impact kinetics was achieved by the subject by increasing the sole angle (defined as the angle between the floor and a line drawn between the posterior calcaneal and hallux markers) by an average of 2.5° immediately prior to heelstrike and by increasing vertical touchdown velocity by an average of 0.19 m/s when wearing the compliant *Titan Bounce* construction. Despite the relatively small sample size, both of these changes were found to be statistically significant (*p* < 0.05).

These adaptive strategies for modulating impact kinetics were also observed in a variety of biomechanical studies employing both human subjects and musculoskeletal models of shod runners used to investigate the relationship between midsole hardness and the magnitude of the passive force peak exerted at heelstrike. As observed in the experimental trials completed in the standard *Titan Bounce* footwear, increased plantarflexion at the ankle and reduced touchdown velocity, as well as increased knee flexion were all reported to reduce the magnitude of the passive force peak (Cavanagh & Lafortune 1980; Gerritsen et al. 1995; Nigg et al. 1987; Wright et al. 1998).

Gerritsen et al. (1995) stated these initial kinematic conditions are primarily responsible for the modulation of impact forces whilst Nigg et al. (1988) concluded that, whilst changes in midsole hardness do not affect the magnitude of the passive force peaks, the point of application of these forces is changed. The kinematic adaptations observed from the experimental trials performed in the two footwear conditions indicate that greater loading would be applied to the rearfoot structures at heelstrike due to increases in the sole angle and touchdown velocity when wearing the compliant *Titan Bounce* footwear. Whilst the magnitude of loading applied to the compliant *Titan Bounce* model in the FE model was accurate, changes in the distribution of loading were not represented. This factor would certainly contribute to the relatively poor correlation between the deformations of the rearfoot TPU structures predicted by the FE model and measured from the experimental trials for the compliant *Titan Bounce* footwear.

This is a fundamental issue when using an FE footstrike model predictively as it is clear that kinematic adaptations to changes in the mechanical properties of the footwear affect the
distribution of loading applied during experimental trials. The neuromuscular control mechanisms employed by a runner to modulate impact force peaks are not represented in the modelling methodology and the results of any analysis must thus be interpreted with an understanding of this limitation. Whilst it is quick and simple to apply this modelling methodology to new footwear designs, the case study presented in this section has demonstrated the dangers of assuming that loading will remain consistent between footwear conditions in real use. A more sophisticated approach to replicate the loading conditions applied to footwear by a human subject is clearly required.

5.4. Chapter Summary

An FE footstrike modelling methodology employing load conditions determined from experimental measurements of plantar pressure was developed with a preliminary analysis of the Reference Shoe first performed such that the modelling approach and validity of predicted loading could be evaluated. As expected, good agreement between experimental and predicted load magnitudes for the vertical GRF was observed with a maximum error of 141 N and an average error of 27 N throughout the footstrike. The distribution of loading was also considered to be largely realistic although it was not possible to validate model strain output against an experimental trial.

A more complex model of the Titan Bounce footwear was developed and loaded with triaxial loading conditions determined using an assumption of proportionality between the distribution of normal loading and the shear forces exerted between the foot and footwear. As before, good agreement was seen between experimental and predicted GRF components with average errors of 20 N, 12 N and 43 N calculated for the anteroposterior, mediolateral and vertical components respectively.

The distribution of predicted loading was validated by comparison to digitised HSV footage of five trials performed in the modelled footwear and, whilst deformation of the TPU structures was generally underestimated by the model, patterns of deformation were largely consistent. The main reasons for this general underprediction of deformation were hypothesised to be measurement error associated with the digitisation of trial HSV footage, difficulties consistently locating the pressure insole within the footwear during experimental testing and an oversimplified contact interface between the foot and footwear.
The predictive capacity of the model was subsequently evaluated by applying identical loading conditions as those used in the standard model to a construction of the *Titan Bounce* employing more compliant TPU structures. Experimental and predicted deformations both increased although the average increase was found to be significantly greater in the experimental trials than predicted by the model with the greatest underestimations observed at the rearfoot TPU structures. This was primarily attributed to the fact that the loading conditions applied in the experimental trial were not identical to those exerted in the standard footwear construction due to kinematic adaptations employed by the subject to ensure consistent GRF magnitudes. These biomechanical adaptations resulted in a change in the distribution of loading not represented in the load conditions defined in the analysis. Whilst the model could still act as a useful comparative tool in the early development of an item of footwear, it was however concluded that a more sophisticated approach would be required to allow biomechanically representative loading to be applied to a virtual footwear assembly.
6. Kinematic Footstrike Modelling

A major limitation of the Pedobarographic foot strike model reported in Chapter 5 was that deformation of the footwear assembly due to relative motion of the sub-regions of the foot was not considered in the model. This chapter details the development of a fully dynamic foot strike model driven with six DOF kinematics determined for a multi-segment foot model which is then validated against experimental foot strike kinetics and HSV footage. The Kinematic foot strike model is more computationally expensive than the Pedobarographic model reported in Chapter 5 but was developed to provide a fuller representation of the loading that occurs during a human foot strike. It is thus better suited to the latter stages of product development.

The scale of uncertainty involved in the measurement of foot segment kinematics is discussed with the impact of these uncertainties on model output then investigated via a sensitivity analysis in which the location and orientation of the force platform is altered. The results of this sensitivity analysis then enable conclusions about the predictive capacity of a kinematically driven foot strike model to be reported.

6.1. Model Overview

The development and evaluation of a dynamic FE foot strike model replicating a single experimental trial performed in the Feather Zero footwear construction by Subject B (see Table 3.1) is reported in this chapter. Six DOF kinematic boundary conditions for a three segment foot model were determined for the running trial with the lowest measured aggregate segment residual value (rearfoot: 2.0 mm, midfoot: 2.3 mm, forefoot: 2.4 mm) relative to the GCS origin using the methodology detailed in Section 3.2. A detailed description of all relevant model features is presented hereafter.

6.1.1. Orientating Footwear Components

In order to achieve meaningful output from the FE foot strike model it was necessary to orientate the virtual footwear assembly such that it matched the position assumed at a selected frame prior to contact in the experimental trial. This was achieved by first capturing the 3-D geometry of the lasted footwear and its attached markers with an ATOS I 800 Digitizer scanner (GOM, Braunschweig, Germany) after the completion of all experimental trials. As
shown in Figure 6.1, fringe projection scanners work by projecting a collimated sinusoidal fringe pattern onto an object placed in the capture volume (Gorthi & Rastogi 2010). The pattern displacement is then recorded with stereo 800 000 pixel cameras (GOM 2005). The position of each point relative to the camera can then be calculated as follows:

\[
h(x, y) = \frac{BD}{\sin \alpha} = \frac{p}{\sin \alpha} \frac{\varphi(x, y)}{2\pi} = k\varphi(x, y)
\]

*Equation 6.1*

where \( h \) is the distance of the point from a reference plane, \( p \) is the fringe period, \( \alpha \) is the angle between the projection and detection directions, \( k \) is an optical coefficient related to the configuration of the optical measuring system and \( \varphi(x,y) \) is a phase modulated by the surface profile (Quan et al. 2004).

![Fringe projection method](image)

*Figure 6.1 - Fringe projection method. Recreated from Quan et al. (2004).*

Specialist application software can then be used to build a three-dimensional point cloud of the object being scanned from the two recorded images (Zhang 2010). With this technique, the three dimensional geometry of a shoe and its attached markers was captured at a resolution of 0.12 mm and output as an STL point cloud (GOM 2005).

Inspection and further manipulation of the scan geometry was performed with Rhino, a commercial 3-D modelling tool (McNeel, Seattle, USA). An appropriate frame from the experimental trial occurring shortly before contact was identified with points created in the Rhino workspace at the centre of each marker contained in the rearfoot and forefoot.
segments. The point cloud representing the *Feather Zero* footwear geometry could then be transformed to match these locations. This process is illustrated in Figure 6.2.

![Figure 6.2 - Orientating scan geometry (grey) such that attached markers (orange) match locations output during biomechanical trial for the forefoot (red) and rearfoot (green) segments.](image)

Perfect alignment between the centre of the scanned markers and the locations recorded during the experimental trial was not achievable as the scan geometry differs from the geometry of the footwear at this instant in the biomechanical trial for a number of reasons. One such reason is that the shoe upper deforms around the foot and the position of the markers is in part a function of foot geometry. With the equipment available, it was not however possible to scan the footwear as it was being worn by the experimental subject. A last was therefore used to represent the geometry of the human foot.

Differences in the geometry of the last and the subject’s foot therefore influence the position of the markers and introduce error into this procedure. A second source of error is that the subject’s foot is not in a neutral position immediately prior to contact and this causes a deformation of the footwear that cannot be replicated during the scanning procedure. As a result of these errors, a best fit was sought when aligning the scanned geometry to the marker locations with an average residual of 4.2 mm recorded at each marker location and a maximum residual of 10 mm observed at the hallux.

As seen in Figure 6.3, CAD geometries of the footwear components provided by the industrial collaborator were then imported and aligned to the scan geometry. Known differences between the geometries of components in the assembled footwear and the CAD geometries taken from component tooling profiles meant this procedure again required an alignment best
fit to be attempted. The magnitude of error involved in this procedure was more difficult to accurately quantify but can be considered minimal in comparison to the other sources of error discussed previously in this section. Once orientated in the Rhino workspace, component geometries were then exported as IGES files to be meshed for use in the FE footstrike model.

![Figure 6.3 - Alignment of footwear component geometries to scanned geometry of assembled footwear. Topplate (blue), midsole (yellow) and outsole (purple).](image)

**6.1.2. Constraints and Boundary Conditions**

As seen in Figure 6.4, the three functional segments of the foot were represented by plates tied to the top surface of the *Feather Zero* topplate and constrained to behave as rigid bodies. The dimensions of each of these plates were determined from a sagittal MR scan of the foot-ankle complex.

![Figure 6.4 - Kinematic footstrike model including *Feather Zero* footwear components and the three rigid foot segments.](image)
The rigid body kinematics of each of these plates is governed by that of its associated rigid body reference node which was positioned at the local origin of each foot segment as defined in the biomechanical model described in Section 3.2.2. Transient six degree of freedom displacement boundary conditions were applied to the three rigid body reference nodes in order to drive each segment independently through the global coordinate system. The procedure for determining the amplitudes used in the displacement boundary conditions was detailed in Section 3.2. Initial velocity conditions were calculated from the biomechanical trial data and applied to the assembly so that the elements would not be accelerated from a velocity of zero in the first increment of the analysis. Finally, the rigid geometry of the force platform was also created in the model and constrained in all six spatial DOFs.

Preliminary analyses indicated that further constraint was required at the midfoot and metatarsophalangeal intersegmental joints and this was achieved by introducing a representation of the human foot to the assembled model. The aim of the model was not to investigate internal loading within the foot itself, but rather to ensure that biomechanically representative loading was applied to a virtual footwear design. It was thus decided that the approaches to human foot modelling discussed in Section 2.4.1 were unnecessarily complex for the goals of the model and that coupling a homogenous 3-D structure with the geometry of a foot prosthesis (Otto Bock, Duderstadt, Germany) to the motion of the rearfoot and forefoot plates would prove sufficient. The complete assembled FE model including the foot geometry and force platform is shown in Figure 6.5:

![Figure 6.5 - Undeformed state of kinematic footstrike model including foot and force platform.](image-url)
Further analysis of preliminary models indicated that unrealistic deformation of the footwear components was occurring, particularly at the metatarsophalangeal joint. This was due to a degree of overconstraint caused by relative motion of the three foot plates and the tie constraints defined between them and the footwear components. To overcome this, the tie constraint between the midfoot plate and Feather Zero topplate was replaced with a coupling defined to apply no constraint between the two bodies in the local A-P direction. This alteration had a minimal effect on overall footwear kinematics but as seen in Figure 6.6, prevented the occurrence of unrealistic local deformations in the model.

![Figure 6.6 - Comparison of approaches for linking rigid midfoot segment to Feather Zero topplate. a) Tie constraint, and b) coupling with local A-P direction unconstrained.](image)

6.1.3. Material Modelling

**Footwear components**

The material models required to represent the Feather Zero were polyamide for the topplate, EVA for midsole and blown rubber for the outsole. All three materials were defined as described previously in this thesis with the polyamide material model as presented in Section 5.2.2 and the EVA and blown rubber materials as reported in Section 4.2.3.

**Human foot**

An incompressible second-order hyperelastic material model employing a Yeoh strain energy function (see Equation 4.5) was used to characterise the behaviour of the homogenous foot geometry with material parameters reverse engineered to provide sufficient constraint at the midfoot and metatarsophalangeal joints included in the model. As discussed in Section 6.1.2, this approach was considered sufficient as the purpose of the model was not to determine the biomechanics of the foot tissue itself, but rather to ensure that appropriate loading was
applied to the footwear geometries. Foot density was determined from body segment mass data published by de Leva (1996).

6.1.4. Contact

When modelling the interaction of a footwear component with any other body, it is important to select an appropriate contact enforcement algorithm. Two contact algorithms are available in Abaqus/Explicit - the kinematic contact method and the penalty contact method (Abaqus 2012d). The kinematic contact method works by disregarding any defined contact controls at the start of each increment and advancing the kinematic state of a model into a predicted configuration. The depth of penetration and associated mass of the penetrating nodes is then measured and used to calculate the force required to impart the acceleration necessary to oppose this penetration. The increment is then recalculated with the opposing forces applied to ensure that penetration does not occur. Alternatively, the penalty contact method searches for nodal penetration in the current configuration and then applies a “penalty stiffness” to overcome this penetration in subsequent increments. The scale of this penalty stiffness is selected by the software so as to have a limited effect on the analysis stable time increment but also prevent significant penetration from occurring in most analyses.

Hard kinematic contact control completely prevents penetration but is not as robust as the penalty contact method which can be used to model a wider variety of contact types, e.g. contact between two rigid bodies. In Abaqus/Explicit, general contact can also be defined to quickly apply contact controls to all bodies in an analysis as opposed to having to manually define all surface pairs that may potentially come into contact. General contact can only be used with the penalty contact enforcement algorithm so was thus defined for the model presented in this chapter. Frictionless tangential behaviour was also defined for all bodies coming into contact in the analysis.

6.1.5. Element Selection

The Feather Zero topplate component is suitably thin to be modelled by computationally cheap linear shell elements with an assigned thickness determined by taking the average thickness from the 3-D CAD geometry for the component. The rigidity of both the force platform and the plates representing the functional segments of the human foot was ensured by meshing the plates with rigid triangular R3D3 elements.
Due to their relatively complex geometry, the foot, midsole and outsole components were best represented with tetrahedral continuum elements. Linear C3D4 elements were employed during initial model development to minimise computational expense with a comparison of model output obtained when employing quadratic C3D10M elements made with the final model. As seen in Figure 6.7, similar footstrike kinetics were output by the two modelling approaches with a reduction in the active force peak of only 92 N (4.9 %) observed when employing C3D10M elements.

![Figure 6.7 - A comparison of predicted vertical GRF output when employing linear (C3D4) and quadratic (C3D10M) tetrahedral continuum elements.](image)

As discussed in Section 4.2.6, a compromise between accuracy and computational expense must be reached when developing any FE model. The model employing quadratic continuum elements contained approximately 127 000 nodes in comparison to only 29 000 for the linear model and this resulted in a 251 % increase in the solve time for the analysis. It was thus decided that the linear C3D4 element provided a sufficiently accurate solution whilst remaining computationally inexpensive and should therefore be used for future footwear development in all but exceptional circumstances.

### 6.2. Results and Validation

The kinematic FE footstrike model was validated and evaluated by comparing requested analysis output to biomechanical data obtained from the experimental trial used to determine model boundary conditions.
6.2.1. HSV Footage

Whilst it is only possible to make a visual comparison, Figure 6.8 shows a clear resemblance between the model field output and synchronised high speed video footage although it should be noted that this is to be expected with a kinematically driven footstrike model.
Figure 6.8 - A comparison of model field output and experimental HSV footage. a) Medial view, and b) posterior view.
6.2.2. Vertical GRF

Analysis of the predicted GRF presented in Figure 6.9 shows that distinct passive and active force peaks are observable and that they are of a comparable magnitude to those measured in the experimental trial. However, the simulated impact force peak was found to be 26% lower than in the biomechanical trial whilst the passive force peak was 14% higher than during experimental testing. This indicates that the discrepancy between simulated and experimental loading profiles is not due to a systematic error.

![GRF Graph](image)

**Figure 6.9 - A comparison of experimental and predicted vertical GRFs.**

The total duration of the stance phase was also reduced by 0.034 s in the FE analysis due to the poor fit achieved for the forefoot segment when orientating the footwear geometries to the experimental marker locations (see Section 6.1.1). The lack of contact that occurs between the footwear and force platform during terminal stance can be observed in Figure 6.10:
Figure 6.10 - Model field output at 85% of experimental stance phase (0.21 s) indicating that no contact between the footwear and outsole occurs at this time.

6.2.3. Centre of Pressure

COP location is known to be highly sensitive to errors in the moment and force components measured by the force platform load cells when the vertical GRF is small (Kwon 1998). The first and last 0.01 s of the experimental footstrike was therefore ignored when validating the COP location predicted with the FE model to prevent erroneous trial data from being included in the comparison.
Figure 6.11 - A comparison of experimental and predicted COP location.

Figure 6.12 - A comparison of experimental and predicted COP location. a) A-P axis, and b) M-L axis.

Figure 6.11 and Figure 6.12 show fair agreement between experimental and predicted COP location. The average residual between experimental and simulated output location was 12.1 mm (maximum of 24.7 mm) in the anteroposterior direction and 7.8 mm (maximum of 13.8 mm) in the mediolateral direction. The period at the end of the simulated footstrike where there was no contact between the footwear and force platform could obviously not be included in this calculation.
6.2.4. Summary of Results

The kinematic footstrike modelling methodology presented in this section has been shown to demonstrate good agreement to both experimental HSV footage and COP output location. Experimental and simulated GRF traces were also found to be broadly comparable. However, despite similarities in the GRF traces, simulated loading was an inexact representation of that applied during the biomechanical trial. This can primarily be attributed to errors introduced when fitting the biomechanical foot model to the dynamic trial in order to determine foot segment kinematics and the methodology employed to determine the initial orientation of the footwear components in the FE analysis.

An example of how a kinematic footstrike model could be used to predict footwear performance despite the limitations of the method is presented in Section 6.3 whilst an investigation into the sensitivity of model output to these errors follows in Section 6.4. Finally, a full evaluation of the modelling methodology, its associated limitations and its predictive capacity is presented in Section 6.4.4.

6.3. Example Application: Footwear Stiffness

6.3.1. Background

Running footwear is typically engineered such that different areas of a shoe have different properties. A common example of this is when an item of footwear is designed to dissipate energy and attenuate impact loading at the rearfoot but minimise the energy losses that occur at the forefoot during the propulsive phase of gait (Chase 2009). This can be achieved by carefully selecting appropriate materials and engineering component geometries to achieve the desired amount of stiffness in the different regions of the footwear.

The cushioning properties of athletic footwear are traditionally evaluated with mechanical compression tests administered at the rearfoot and forefoot of the midsole. Load profiles representative of those that would typically occur during a running footstrike are applied with rigid stamps to allow the stiffness of the footwear to be determined by analysis of the force-displacement plot (ASTM 2006). The test apparatus used by the project’s industrial collaborator to administer these compressive tests is shown in Figure 6.13:
Figure 6.13 - Rigid stamps and compression tests in progress. a) Rearfoot, and b) forefoot.

A number of finite element models replicating these mechanical tests have been reported (e.g. Nakabe & Nishiwaki 2002; Gibbs 2006; Mara 2007) but suffer from similar limitations as the mechanical tests discussed in Section 4.2.1 in that the loading conditions applied are not necessarily representative of the complex loading that occurs during a typical running footstrike. A methodology developed to predict footwear stiffness properties from a dynamic model employing full footstrike kinematic boundary conditions is reported hereafter.

### 6.3.2. Model Overview

History output requests were created to allow the forces exerted on the Feather Zero midsole and the displacements predicted to occur as a result of these forces to be extracted from the analysis. Rearfoot and forefoot surfaces were generated and then used to define two “integrated output sections” (IOSs) which allow the average motion of a surface and the sum of forces acting across said surface to be easily output. The IOS surfaces were defined so as to replicate the both the geometry and location of the stamps used during mechanical testing and are shown in Figure 6.14:
6.3.3. Results

Figure 6.15 shows the predicted forces acting across the rearfoot and forefoot regions whilst Figure 6.16 shows the average deformation of the two regions. The force and deflection data is then combined in Figure 6.17 to allow the predicted stiffness of the two regions to be calculated. This shows that the model predicts the forefoot to be approximately 65.4% stiffer than the rearfoot when considering the maximum load cases for both regions. Figure 6.17 also shows excellent agreement between the stiffnesses measured with the mechanical tests and those predicted with the FE model but that the magnitude of loading applied was much smaller in the simulation.
6.3.4. Discussion

The results reported in Section 6.3.3 illustrate that the FE footstrike model presented in this chapter could be used to predict the rearfoot and forefoot stiffnesses of a prospective footwear design. This is however merely a function of the geometry and material properties assigned to footwear components. The novel feature of the reported model was the method of load application and this approach was found to be incapable of applying experimentally representative load magnitudes to the modelled footwear assembly. Rearfoot loading and deformation were lower because the model was shown to underestimate the experimental
passive force peak by approximately 26% (see Figure 6.9). Furthermore, whilst the active force peak was increased in the FE model relative to the experimental motion capture trial, it is likely that the loading that occurred across the forefoot IOS was underestimated as this surface spans the metatarsophalangeal joint, an area between the rigid midfoot and forefoot plates that remains unconstrained in the model.

Despite this, the level of agreement between the stiffnesses predicted with the FE footstrike analysis and those measured with mechanical testing suggests that the modelling approach could be used to virtually evaluate a footwear design under more complex loading conditions than typically used during traditional mechanical testing. The modelling approach could also be used to simultaneously characterise the stiffness of the footwear in any number of different sub-regions, not only for the rearfoot and forefoot segments typically evaluated with mechanical testing. Whilst this represents progress, it must be remembered that predicted stiffnesses are only valid for the range of loading and deformations simulated. The sensitivity of the stiffnesses predicted with the model to foot segment kinematics is investigated in Section 6.4.3:

6.4. Sensitivity Analysis

When fitting the three segment foot model to the experimental trial replicated in the kinematic footstrike model segment residual values of up to 2.4 mm were observed. The procedure employed to orientate the footwear geometries relative to the force platform in the first frame of the analysis also relied on a best fit procedure and led to differences in the location of the experimental and simulated marker centres of up to 10 mm. These values are very significant compared to the geometries of the footwear components modelled. It was therefore necessary to investigate the sensitivity of model output to the defined kinematic boundary conditions and thus evaluate the predictive capacity of such a modelling approach. This was achieved by altering the pose of the simulated force platform and comparing model output to the results obtained from the base model presented in Section 6.2.

6.4.1. Vertical GRF

The sensitivity of the vertical GRF predicted by the model to defined kinematic boundary conditions was examined by translating the force platform ± 2 mm in the vertical direction (along the D-V axis). A translation of 2 mm was selected to correspond with the minimum
segment residual value observed during biomechanical modelling as described in Section 3.2. The sensitivity of predicted vertical GRF to this alteration can be observed in Figure 6.18:

![Figure 6.18 - Variation of predicted vertical GRF as a result of vertically translating virtual force platform.](image)

A translation of + 2 mm along the D-V axis was found to increase the passive force peak by 52 % and the active force peak by 120 %. Similarly, translating the force platform - 2 mm was found to reduce the passive and active force peaks by 34 % and 44 % respectively. The results of this study clearly indicate that the magnitude of loading predicted by the model is highly sensitive to the initial orientation and positioning of footwear geometries. This could potentially be exploited to improve the agreement between the vertical load profiles measured during experimental testing and predicted with the FE model by adjusting the position of the force platform throughout the analysis.

**6.4.2. Centre of Pressure**

To serve as a useful tool in the development of novel footwear designs an FE footstrike model must not only be capable of applying realistic load magnitudes, but also be able to accurately represent the distribution of loading that occurs during a typical footstrike. It was therefore necessary to investigate the sensitivity of model COP output to the kinematics defined in the model. To achieve this, the base analysis was resubmitted with the virtual force platform translated and rotated into a number of new positions - firstly by translating ± 2 mm vertically, secondly by rotating ± 1° about the sagittal axis and finally, by rotating ± 1° about the frontal
axis. The results of these simulations and effect on model COP output location are shown in Figure 6.19:

![Figure 6.19 - Variation of COP output location as a result of repositioning virtual force platform.](image)

In comparison to the base model, COP output location was found to be most sensitive to rotation about the sagittal axis with a maximum change of 69 mm. This was largely due to an extended period of contact at toe-off. Output location was altered by up to 26 mm and 19 mm for vertical translation and rotation about the frontal axis respectively. Analysis of these results suggests that both the magnitude and distribution of predicted loading is highly sensitive to both the initial orientation of footwear components and the accuracy of the defined kinematic boundary conditions. Similarly to as was discussed in Section 6.4.1 for the
vertical GRF, adjusting the orientation of the force platform throughout an analysis could potentially improve the agreement between the experimental and predicted COP output location.

### 6.4.3. Footwear Stiffness

The kinematic FE footstrike model was shown in Section 6.3 to provide accurate predictions of the rearfoot and forefoot stiffness of the *Feather Zero* but it was conceded that these predictions would only be valid for the range of loading and deformations applied during the virtual footstrike. As the level of deformation that occurs is entirely dependent on the defined boundary conditions it was deemed important to investigate the sensitivity of the stiffnesses predicted with the model to these foot segment kinematics. The effects of translating the force platform ± 2 mm vertically on the simulated rearfoot and forefoot loading profiles, deformations and predicted stiffnesses are shown in Figure 6.20, Figure 6.21 and Figure 6.22:

![Graph showing the effect of vertical force platform translation on rearfoot and forefoot loading profiles and stiffness predictions.](image)

**Figure 6.20** - Variation of vertical force exerted across rearfoot and forefoot IOS as a result of vertically translating virtual force platform.
Figure 6.21 - Variation of average predicted deformation of rearfoot and forefoot IOS as a result of vertically translating virtual force platform.

Figure 6.22 - Variation of predicted footwear stiffnesses as a result of vertically translating virtual force platform. a) Rearfoot IOS, and b) forefoot IOS.
Vertical translation of the force platform acts to increase or decrease both the deformations that occur at the rearfoot and forefoot integrated output sections (Figure 6.20) and the vertical reaction forces acting across each IOS (Figure 6.21). However, as seen in Figure 6.22, the stiffness of the model is relatively unaffected with all three analyses predicting similar force-deflection profiles for both the rearfoot and forefoot integrated output sections. It is therefore possible to conclude that the loading applied to the footwear with an FE footstrike model is highly sensitive to the defined foot segment kinematics but that predicted footwear stiffnesses are relatively unaffected. A discussion on how this impacts the practicality of the FE footstrike model as a footwear development tool and an evaluation of the model’s predictive capacity is presented in Section 6.4.4.

6.4.4. Discussion

The FE footstrike model presented in this chapter has been shown to output comparable data to that measured from the experimental trial used to determine the kinematic boundary conditions defined in the model, particularly for the HSV footage and COP location. Despite this, model GRF and COP output was shown to be highly sensitive to the defined foot segment kinematics and, whilst predicted footwear stiffnesses were relatively unaffected, the range of loading for which these stiffnesses were calculated was also found to be similarly sensitive.

Model sensitivity was evaluated either by translating the force platform ± 2 mm vertically or by rotating it ± 1° about the sagittal or frontal axes. This scale of adjustment was selected as it was comparable to the level of uncertainty in the defined kinematic boundary conditions and was found to have a very significant impact on the outputs predicted with the model. 2 mm is equal to the smallest segment residual value calculated when determining foot segment kinematics whilst the procedure employed to determine the initial orientation of the footwear components resulted in an average residual of 4.2 mm at each marker location. The maximum deformation predicted with the model was 4.8 mm so with kinematic uncertainties comparable to this value it was possible to conclude that loading highly representative of the experimental trial was not achievable when using kinematic boundary conditions as reported in this chapter.

The aforementioned uncertainty in determining foot segment kinematics and the initial orientation of the footwear components are the primary limitations of the reported
methodology but further limitations of the approach also exist. These include a simplified representation of the foot-shoe contact interface, the use of rigid plates to simulate the complex anatomy of the human foot, discrepancies between the footwear geometries meshed and those of the assembled footwear and the homogenous material model assigned to a scanned foot prosthesis that does have the same dimensions as that of a human foot. It should also be noted that the marker system used to determine foot segment kinematics was developed for barefoot, as opposed to shod use. Improved results may have been achieved if a kinematic model developed to determine the kinematics of the foot-shoe complex rather than foot in isolation (e.g. Bishop et al. 2013) had been employed.

Aside from concerns about the accuracy and sensitivity of the analyses presented in this chapter, a second limitation of the reported methodology is that it is based on the reverse engineering of a previously performed biomechanical trial. In order to obtain input kinematics, the footwear must be constructed so that experimental trials can be performed. This approach clearly does not satisfy one of the primary goals of this thesis - to develop a modelling methodology that could be used as a predictive tool, allowing a footwear design to be evaluated virtually and negating the need for physical prototyping.

Despite these limitations, it is clear that the methodology presented allows for complex, multiaxial loads largely characteristic of a human footstrike to be applied to a proposed footwear design and could provide genuine insight into the performance of an item of footwear currently in development. However, the results of the sensitivity analysis presented demonstrate the sensitivity of model output to foot segment kinematics and raise concerns about the predictive capacity of a kinematically driven footstrike model.

A 2 mm adjustment in the position of the force platform resulted in a maximum increase in vertical loading of 120%. A comparable modification to footwear geometry could be expected to cause similarly unrealistic loading with changes in the footwear components’ material properties also resulting in unrepresentative load magnitudes. Despite the demonstrated similarities between model output and an experimental trial, it is therefore possible to conclude that the capacity of a purely kinematically driven footstrike model to predict the mechanical behaviour of a proposed footwear design is limited to cases where the geometry and properties of the novel design do not deviate significantly from the footwear used during the original experimental trial.
6.5. Chapter Summary

An FE analysis of a running trial performed in the Feather Zero footwear was developed and used to evaluate a fully dynamic footstrike modelling methodology employing six DOF kinematic boundary conditions. The initial orientation of footwear geometries was determined by performing a 3-D scan of the Feather Zero and its attached markers and then aligning the marker centres to positional data taken from a frame of the experimental trial occurring immediately prior to contact. This procedure required a best-fit to be sought and resulted in an average residual of 4.2 mm at each marker location. Rigid plates representing the three functional segments of the foot were then tied to the Feather Zero topplate and driven with displacement boundary conditions applied to each plate’s respective rigid body reference node. The defined boundary condition amplitudes were determined following the procedure reported in Section 3.2 with a minimum segment residual value of 2.0 mm observed.

Model output was validated by comparison to the data recorded from the corresponding biomechanical trial. An excellent representation of experimental HSV footage was achieved with model COP output readings found to be broadly comparable to experimental data for the trial modelled. However, there were significant discrepancies between the experimental predicted vertical GRF traces with the predicted passive force peak 26 % lower than that recorded during the corresponding experimental trial.

Despite this issue, the footstrike modelling methodology can still be said to have provided output somewhat representative of the running trial emulated. A methodology was also presented to determine the cushioning properties of the footwear, a key parameter in footwear development. Whilst rearfoot and forefoot stiffnesses predicted with the analysis were in good agreement with mechanical test data, the model was found to be incapable of applying biomechanically representative load magnitude at each of these locations.

A sensitivity analysis was subsequently performed to investigate the impact of the uncertainties introduced when calculating foot segment kinematics and determine the initial orientation of the footwear components on model output. This was achieved by resubmitting the model with the force platform translated ± 2 mm vertically or rotated ± 1° about the sagittal and frontal axes. Model output was found to be highly sensitive to these changes with predicted GRF peaks increasing by a maximum of 120 % and COP location changing by up to 69 mm. Predicted footwear stiffnesses were not significantly affected by the changes but it
was conceded that these stiffnesses could only be considered valid for the range of loading applied; a variable that was also found to be sensitive to the defined kinematics.

Whilst the kinematically driven footstrike model was considered to provide a reasonable representation of the experimental trial emulated, demonstrated sensitivity to the defined boundary conditions and the uncertainty involved in determining these values clearly limits the predictive capacity of such a model. It was therefore concluded that such an approach should only be employed when evaluating a prospective footwear design with very similar geometry and properties to that used in the experimental trial from which the boundary conditions had been determined.
7. Combined Load Footstrike Modelling

The kinematic modelling methodology reported in Chapter 6 was found to output results generally representative of an experimental trial but uncertainties in the measurement of foot segment kinematics and orientation of footwear geometries meant that highly representative load profiles could not be achieved. It was also concluded that the predictive capacity of the model was severely limited by the sensitivity of model output to the defined kinematic boundary conditions. The FE footstrike model presented in this chapter attempts to overcome these limitations by adopting a combined loading approach incorporating both kinematic and kinetic boundary conditions. As such, it is hereafter referred to as the “Combined” model with analysis output frequently compared to that obtained with the “Kinematic” model reported in Chapter 6.

The methodological differences between the Combined and Kinematic approaches are reported followed by a discussion on the validity of the output achieved with the Combined model. The development of a footstrike model representing a lateral cut and incorporating triaxial kinetic load conditions is then reported with this model subsequently used to evaluate the predictive capacity of the Combined approach.

7.1. Running Footstrike Model

7.1.1. Model Overview

The ability to apply appropriate loading to a virtual footwear assembly is absolutely essential if an FE footstrike model is to be used as a tool with which to evaluate prospective footwear designs. The foot segment kinematics defined in the Kinematic model presented in Chapter 6 are representative of an experimental trial but the residual in the measurement of these values and best-fit procedure required to orientate the footwear components relative to the force platform meant that highly representative loading of the footwear was not achievable. In order to overcome this, a concentrated force was applied to the force platform’s rigid body reference node with an amplitude equal and opposite to the vertical GRF measured during the experimental trial. The kinematic constraint on the vertical translation of the force platform was removed to allow motion of the force platform relative to the footwear to occur. The model was otherwise identical to the Kinematic model reported in Chapter 6 and is shown in Figure 7.1:
This Combined approach employing both kinematic and kinetic boundary conditions allows the position of the force platform to adapt such that an appropriate level of loading is applied to the footwear assembly at all times throughout the footstrike. The translation of the force platform is governed by Newton’s second law (Equation 7.1) in that the acceleration \( (a) \) is proportional to the net force \( (ΔF) \) of the system and inversely proportional to the mass assigned to the force platform \( (m) \). The net force is the difference between the force applied to the force platform purely via the foot segment kinematics and that defined as a load condition on the force platform’s rigid body reference node.

\[
ΔF = ma
\]

Equation 7.1

As such, the point mass assigned to the force platform’s rigid body reference node functions as a damping parameter for the system. If it were made to be too high the translational response of the system would be sluggish with the magnitude of applied loading lagging behind the prescribed amount. In contrast, selecting a value too low would result in the predicted GRF profiles being overly noisy. It was therefore necessary to perform a preliminary analysis to determine the optimum mass to be assigned to the force platform. The kinematic response of the force platform and vertical GRF profiles output are presented in Figure 7.2 and Figure 7.3 respectively:
Figure 7.2 - Vertical translation of force platform with different masses assigned to the force platform rigid body reference node. First 0.1 s of footstrike shown only.

Figure 7.3 - Predicted vertical GRF with different masses assigned to the force platform rigid body reference node.

Figure 7.2 and Figure 7.3 highlight the importance of selecting an appropriate mass for the force platform in order to avoid noisy model output but still ensure the desired loading profile is applied. The results of these preliminary analyses indicated that a mass of 2.5 mg provided the optimum compromise and was therefore assigned to the force platform’s rigid body reference node for all subsequent models reported in this chapter.
7.1.2. Results

As in previous chapters, model output was validated by comparison to data obtained from the experimental trial used to determine foot segment kinematics and the desired loading profile. Model output obtained with the Kinematic model is also presented in this section to allow for a thorough evaluation of the Combined methodology to be made.

**Vertical GRF**

![Vertical GRF Graph](image)

**Figure 7.4 - A comparison of experimental and predicted vertical GRF with Combined and Kinematic models.**

Figure 7.4 shows that a near perfect representation of experimental loading was achieved with the Combined model although this is to be expected as the experimental GRF is now being used as a model input. The average absolute residual observed over the duration of the footstrike was 6 N with a maximum residual of 33 N. This slight loss of accuracy can be attributed to the numerical damping techniques employed.

The contact time in the Combined model was also identical to that recorded in the experimental trial. This represents significant progress over the kinematic model which underestimated the contact period by 0.034 s. Figure 7.5 shows a comparison of model field output at 85 % of the experimental stance phase and illustrates that the footwear is still being loaded with the Combined model but that no contact is occurring at the equivalent moment with the Kinematic model.
Figure 7.5 - Model field output at 85% of experimental stance phase (0.21 s). a) Combined model contacting force platform, and b) Kinematic model not contacting force platform.

Centre of pressure

Figure 7.6 - A comparison of experimental and predicted COP location with Combined and Kinematic models. a) A-P axis, and b) M-L axis.

COP location predicted with the Combined model was very similar to the output obtained with the Kinematic model indicating that the distribution of simulated load was similar between the two models. Due to the extended contact time, COP location was however output through to toe-off for the Combined model and this resulted in a divergence of up to 64.9 mm from the experimental results in the A-P direction (Figure 7.6a). It is believed that this is due to the particularly poor fit achieved for the hallux marker when orientating the footwear geometries relative to the force platform, as reported in Section 6.1.1, and that this has consequently resulted in a poor representation of the distribution of loading during terminal stance.
Footwear stiffness

Figure 7.7 - A comparison of predicted rearfoot and forefoot loading profiles obtained with Combined and Kinematic models.

Figure 7.8 - A comparison of predicted rearfoot and forefoot deformations obtained with Combined and Kinematic models.
Combined Load Footstrike Modelling

Figure 7.9 - A comparison of predicted rearfoot and forefoot stiffnesses obtained with Combined and Kinematic models.

The rearfoot and forefoot stiffness profiles predicted with the Combined model are very similar to those achieved with the Kinematic model but, as was discussed in Section 6.3.4, stiffnesses predicted with the Kinematic model can only be considered to be valid for the loads and deformations applied. Figure 7.7 and Figure 7.8 show that the loads applied with the Kinematic model were too low at the rearfoot and too high at the forefoot whilst the Combined model ensures that representative load magnitudes are applied to the footwear throughout the entire duration of the footstrike. It can thus be concluded that the range of loading applied to the footwear and subsequently used to determine the stiffnesses of the rearfoot and forefoot regions is significantly more characteristic of a running footstrike with the Combined model than with the Kinematic model.

7.1.3. Discussion

The Combined running footstrike model ensured that a representative force profile was applied to a virtual footwear assembly by allowing vertical translation of the rigid force platform. The maximum translation that occurred during the analysis was 4.6 mm. In comparison, the maximum deformation that was predicted in any part of the footwear assembly was not much larger, only 6.6 mm, indicating the significance of enabling this vertical translation to occur. The distribution of loading achieved with the Combined model was also shown to be a good representation of the experimental trial although predicted COP output location diverged from experimentally measured values during terminal stance. This indicates
that, due to the best-fit procedure required to orientate the virtual footwear assembly, the
distribution of applied loading may not be as accurate with the Combined modelling approach
when the magnitude of loading and subsequent deformations are low.

Despite this limitation, it is clear that the Combined modelling approach provides a significantly
improved representation of the complex loading that occurs during a running footstrike and
thus represents considerable progress over the Kinematic model. The predictive capacity of
the Combined model is investigated in Section 7.3 of this chapter.

7.2. Lateral Cut Footstrike Model

A lateral cut is a sideward cutting movement used to allow an athlete to quickly change
direction and is very common in a number of sports such as basketball, tennis and handball
(Nigg et al. 1989; Stacoff et al. 1993; Suda & Sacco 2011). When performing this action it is
typical for the medial side of the sole to make first contact with the ground and this results in a
large moment lever relative to the subtalar joint axis and potentially injurious inversion torque
(Stacoff et al. 1994; Wright et al. 2000). As a result, lateral stability is an important footwear
property commonly evaluated during the footwear development process (Simpson et al. 1992).

Figure 7.10 - Lateral cutting movement with moment lever relative to the subtalar joint axis
highlighted. From Stacoff et al. (1994).

A primary aim of this thesis was to develop generic modelling approaches that could be applied
to any type of footstrike, not just rearfoot running. The capacity of the Combined model to
realise this goal was therefore investigated with a lateral cut footstrike model developed using
the same methodology as adopted for the running footstrike model but with the addition of
shear loading.
7.2.1. Model Overview

An FE footstrike model emulating an experimental trial of a lateral cut was developed employing a combination of kinematic and kinetic boundary conditions, as with the Combined running model in reported in Section 7.1. The experimental trial was performed with maximum effort in the Feather Zero footwear by Subject B (see Table 3.1) and involved the subject side stepping towards the force platform from the right, striking the force platform with the left foot and then side stepping back towards the start position. Foot segment kinematics were determined as detailed in Section 3.2 with the procedure used to orientate the footwear assembly relative to the force platform as reported in Section 6.1.1.

A lateral cut footstrike imparts a significant mediolateral GRF component so shear forces were included in the model in order to provide a better representation of experimental loading. In order to achieve this all three GRF components were applied to the force platform’s rigid body reference node as concentrated forces with force amplitudes determined from the experimental trial. Rotation of the force platform was prevented with a kinematic constraint but translation in all three spatial directions was allowed.

Tangential contact behaviour between the force platform and outsole was enforced with an isotropic Coulomb friction model employing a specified coefficient of friction. This model assumes that no relative motion of the two surfaces occurs until the equivalent frictional stress ($\tau_{eq}$) exceeds the critical stress ($\tau_{crit}$) which is equal to the product of the COF ($\mu$) and contact pressure ($p$);

$$\tau_{crit} = \mu p$$  \hspace{1cm} \text{Equation 7.2}

$$\tau_{eq} = \sqrt{\tau_1^2 + \tau_2^2}$$  \hspace{1cm} \text{Equation 7.3}

where $\tau_1$ and $\tau_2$ are the shear stresses in the first and second slip directions (Abaqus 2012c). No distinction between static and kinetic COFs was made as the observed static COF for rubber depends on dwell time and rate of starting and it is therefore difficult to define a single value (Persson & Volokitin 2002). The experimental methodology used to determine the employed COF value and a discussion on the limitations of this approach was presented previously in
Section 3.4. Model definition was otherwise as reported for the Combined running footstrike analysis presented in Section 7.1 with the analysis submitted to Abaqus/Explicit solver.

![Image: FE model of a lateral cut footstrike with triaxial GRF components applied to the virtual force platform.]

7.2.2. Results

As with previously reported analyses, the validity of results output with the lateral cut footstrike model was verified by comparison to the experimental HSV footage and GRF component profiles.

**HSV footage**

Figure 7.12 shows that, whilst unquantified, good agreement was observed between model field output and experimental HSV footage recorded of the experimental lateral cut trial. This is consistent with the results of previously presented models employing foot segment kinematic boundary conditions.
Figure 7.12 - A comparison of experimental HSV footage and model field output for the Combined lateral cut model. a) Posterior view, and b) anterior view.
**GRF components**

As was reported for the running footstrike model in Section 7.1.2, the Combined approach applied loading highly representative of the experimental lateral cut trial for all three GRF components. The average residuals between model and experimental GRF values were 12 N, 8 N and 4 N for the vertical, mediolateral and anteroposterior components respectively with maximum residuals of 45 N, 54 N and 33 N observed.

![Graphs showing GRF components comparison](image)

**Figure 7.13** - A comparison of experimental and applied vertical GRF components with Combined and Kinematic models. a) Vertical component, b) mediolateral component, and c) anteroposterior component.
7.2.3. Discussion

The results presented in Figure 7.12 and Figure 7.13 indicate that, as was reported for the running footstrike analysis, the Combined modelling methodology provided a significantly improved representation of a lateral cut footstrike in comparison to a purely kinematic modelling approach. This demonstrates that the methodology could be used to approximate any type of footstrike for which boundary conditions have been calculated, thus satisfying one of the primary goals of the thesis. Another key aim of the thesis was to develop footstrike modelling methodologies that do not just replicate an experimental trial, but that also demonstrate the capacity to be used predictively in the footwear development process. The predictive capacity of the Combined modelling approach was evaluated using the lateral cut model reported in this section with the results of this analysis presented hereafter:

7.3. Evaluation of Predictive Capacity

In Section 6.4.4 it was stated that the sensitivity of the outputs obtained with the Kinematic model to defined foot segment kinematics significantly limited the predictive capacity of such an approach. It has already been demonstrated that the Combined methodology provides an improved representation of an experimental trial but it was also necessary to evaluate the predictive capacity of the Combined approach. This was achieved by resubmitting the Combined lateral cut model and analysing the output obtained after modest geometric and material modifications had been made to the Feather Zero footwear assembly.

The results of these analyses are presented and compared to results obtained with the Kinematic modelling approach. The results reported for the Kinematic model are taken from an analysis in which kinematic boundary conditions were applied to the force platform’s rigid body reference node to match the 3-D translation output with the Combined model employing the unaltered Feather Zero assembly. This ensured that the same GRF profiles would be achieved for the base Kinematic and Combined models, thus allowing a better comparison of the two modelling approaches’ predictive capacity to be made.
7.3.1. Footwear Geometry

Model overview
Determining the optimum dimensions of a footwear component is a common task undertaken during the development of a new footwear design. As shown in Figure 7.14, this was represented by transforming the geometry of the Feather Zero midsole and outsole components such that they became 10% thicker in the vertical direction. This led to a maximum increase in thickness of 2.6 mm at the rearfoot of the Feather Zero assembly. The analysis was resubmitted with no further changes made from the base lateral cut footstrike model.

![Figure 7.14 - A comparison of footwear geometries used in analyses. a) Base Feather Zero assembly used in experimental trial, and b) Modified assembly with 10% increase in component thicknesses.](image)

Results
As illustrated in Figure 7.15, the increased thickness of the footwear geometry led to a significant increase in the predicted vertical GRF for the Kinematic model but loading representative of the experimental trial was still applied with the Combined model. This was achieved because the position of the force platform was able to adapt to the new footwear geometry, something that could not occur with the Kinematic model.
Figure 7.15 - A comparison of predicted vertical GRF profiles obtained for modified Feather Zero assembly with Kinematic and Combined models.

This is demonstrated in Figure 7.16, when employing the Combined approach the vertical position of the force platform was consistently lower for the thick Feather Zero geometry than for the standard geometry. The average difference in position between the two models was 2.7 mm, a value understandably very similar to the 2.6 mm maximum increase in the thickness of the footwear assembly.

Figure 7.16 - Vertical translation of force platform with different Feather Zero footwear geometries modelled.

Despite the improved results achieved with the Combined approach, it should however be noted that the experimental GRF profile was obtained with the standard Feather Zero
assembly, not the thicker geometry used in the FE model. To state that the loading applied to 
the thicker footwear assembly with the Combined model is a good prediction of that which 
would be applied during a hypothetical experimental trial can therefore only be an 
assumption. It is however necessary to make this assumption when using an FE footstrike 
model predictively, the validity of which is discussed in Section 7.3.4.

7.3.2. Footwear Materials

Model overview
As well as the geometry of footwear components, the materials employed in these 
components is another primary topic of interest when new footwear designs are being 
developed. The capacity of the Combined modelling approach to predict the behaviour of 
footwear components with a range of material properties was investigated by varying the 
shore hardness of the EVA material assigned to the Feather Zero midsole. The response to 
uniaxial tensile loading of the two EVA materials modelled is shown in Figure 7.17 with EVA 
45C more compliant and EVA 60C stiffer than the standard EVA 55C respectively. As before, no 
additional changes were made to the base lateral cut footstrike model.

![Graph showing stress-strain relationship for EVA materials](image)

**Figure 7.17** - Response to uniaxial tensile loading of EVA material models used to evaluate predictive capacity of the Combined modelling approach.

Results
Figure 7.18 shows that vertical loading representative of the experimental lateral cut trial was 
applied with the Combined model, irrespective of the EVA material model employed. However 
for the Kinematic model, increasing or decreasing the shore hardness of the EVA material 
employed in the midsole resulted in a corresponding change in the applied loading. The
application of experimentally representative loading meant that the Combined model could be used to predict differences in the deformations that occur in the Feather Zero assembly when employing different EVA materials. For example, the maximum deformation that was predicted to occur in the forefoot was 6.92 mm with EVA 45C, 6.25 mm with the standard EVA 55C and 5.88 mm with EVA 60C.

![Figure 7.18](image.png) 

**Figure 7.18** - A comparison of the vertical loading applied for different EVA shore hardnesses with Kinematic and Combined models.

This information would be of great interest to a footwear developer and is an example of how the Combined model could be used predictively to determine the optimum material to employ in a footwear construction. Contrastingly, deformations predicted with the Kinematic model were identical for all models as the position of the force platform was not able to adapt to the differing footwear constructions modelled. It should also be noted that as discussed previously, the assumption that similar loading would occur when modelling different footwear assemblies must be valid for results predicted with the FE footstrike model to be of any worth.

### 7.3.3. Footwear Loading

**Model overview**

The positive correlation between the body weight of a subject and the loading that they exert on their footwear during rearfoot running is well established (Frederick & Hagy 1986; Kinoshita et al. 1990) with both of linear and non-linear methods proposed to normalise peak GRF magnitudes (Wannop et al. 2012). Whilst these studies deal with rearfoot running as opposed
to a lateral cut footstrike, it is fair to assume that the loads exerted during a lateral cut would also increase with body weight. The ability of the Combined model to predict the response of the footwear assembly under different scales of loading was therefore investigated by adjusting the magnitudes of the three force components applied to force platform’s rigid body reference node to be 70 % and 150 % of the originally assigned values. The vertical loading applied in each analysis is shown in Figure 7.19:

**Results**

The maximum forefoot deformations predicted with the model were 5.77 mm for the 70 % load, 6.25 mm for the standard 100 % load and 6.63 mm for the 150 % load case. This clearly indicates that, under the assumption that foot segment kinematics remain unchanged, the Combined model can be used to predict footwear behaviour under a range of differing loading conditions. This would certainly be of interest during the footwear development process but could not be achieved with the Kinematic modelling approach.

![Figure 7.19 - Three vertical loading profile applied with Combined lateral cut model.](Image)

**7.3.4. Discussion**

It was demonstrated in Sections 7.1 and 7.2 that, in comparison to the Kinematic model, an improved representation of the experimentally measured load profiles can be applied to a virtual footwear assembly with the Combined modelling methodology. However, it has also been shown here in Section 7.3 that the ability of the force platform to translate and ensure that appropriate loading is applied greatly improves the predictive capacity of the Combined
Combined Load Footstrike Modelling

approach. The response of the footwear to experimentally representative loading could thus be predicted, even after changes to the modelled component geometries and assigned material models had been made. Furthermore, the Combined model also allows the response of the footwear under user-defined load conditions to be investigated, something that could not be achieved with the Kinematic model.

When interpreting results obtained with the Combined approach, one must however be aware of the limitations of using any FE footstrike model predictively. The Combined approach ensures that experimentally representative load magnitudes are always applied to a footwear assembly but these loads can only be said to be wholly representative for the footwear in which the experimental trial was performed.

Research into rearfoot running has shown that athletes adapt their footstrike kinematics between footwear conditions to ensure consistent force magnitudes are applied (Clarke et al. 1983; Kaelin et al. 1985; Nigg et al. 1987; Gerritsen et al. 1995) and the Combined approach can be said to somewhat represent this adaptation in that the position of the force platform adapts to ensure the intended load magnitudes are applied. However, this adaptation is simplistic and only involves translation of the force platform relative to the foot-footwear assembly. The complex rotational and inter-segmental translational changes in foot segment kinematics that occur in reality are not manifested. It can thus be concluded that the Combined approach allows experimentally representative load magnitudes to be applied to a prospective footwear design, but that potential changes in the distribution of loading are not represented. Despite this, the Combined modelling methodology has been clearly shown capable of predicting the behaviour of a prospective footwear design and, with this limitation in mind, could certainly be of use in the footwear development process.

7.4. Chapter Summary

An FE footstrike modelling methodology employing a combination of kinematic and kinetic boundary conditions was presented and evaluated in this chapter with the results of an investigation into the predictive capacity of this “Combined” approach reported. Based on the “Kinematic” model discussed in Chapter 6, the Combined model employs the definition of experimentally representative load profiles to the force platform’s rigid body reference node in addition to six DOF foot segment kinematics.
This approach was shown to provide a significantly better representation of the experimentally measured GRF profile in comparison to a Kinematic model of a rearfoot running footstrike. The average residual between the experimental and predicted vertical GRF was only 6 N with an identical contact period observed between the experimental trial and analysis results. Despite evident progress, the Combined model was also shown to share some of the same limitations as its parent Kinematic model, namely uncertainty in the initial orientation of the footwear assembly which resulted in a divergence from the experimental results in the predicted location of the COP during terminal stance.

The Combined approach was also shown to be applicable to other types of athletic actions with the same methodology adopted to model a lateral cut footstrike incorporating shear loading via a tangential contact definition. This model also demonstrated excellent agreement to experimental GRF profiles with average residuals of 12 N, 8 N and 4 N for the vertical, mediolateral and anteroposterior components respectively.

This lateral cut model was subsequently used to evaluate the predictive capacity of the Combined modelling approach which was shown capable of applying experimentally representative loading to modified footwear assemblies featuring geometrical and material modifications from the standard Feather Zero construction. The ability of the force platform to translate in all three spatial dimensions was also shown to enable the behaviour of the footwear to be predicted under user-defined load profiles.

As has been discussed in previous chapters, interpretation of the results obtained with an FE analysis must be conducted with an appreciation of the limitations of the approach adopted. When reviewing results obtained with a Combined analysis of a novel footwear design it must therefore be considered that the experimental load conditions modelled were measured for an alternative footwear construction. There is hence an assumption that the applied loading would be similar to that which was measured in the experimental trial. Kinematic adaptations employed by an athlete to ensure consistent force magnitudes are somewhat represented by the translation of the force platform but complex adjustments in the foot segment kinematics and subsequent changes in the distribution of loading were not manifested. Despite this limitation, the Combined approach has demonstrated significant potential as a predictive FE footstrike model to be used in the development of modestly revised athletic footwear designs.
8. Advanced Kinetic Footstrike Modelling

It was demonstrated in Chapter 7 that, whilst the Combined approach was effective in applying experimentally representative load magnitudes, the inability of the defined foot segment kinematics to adapt after alterations were made to the modelled footwear assembly meant that it was less successful in ensuring representative load distributions were exerted. This limitation restricts the predictive capacity of such an approach and would pertain to any footstrike model employing rigidly defined kinematic boundary conditions. The FE footstrike model presented in this chapter was subsequently developed with this limitation in mind and therefore employs exclusively kinetic load conditions determined through the inverse dynamic analysis of an experimental trial.

The fundamental principles of inverse dynamics are first discussed with the biomechanical foot model developed to determine three-dimensional ankle joint reaction forces and moments during rearfoot running detailed thereafter. The “Kinetic” footstrike model employing these load conditions is then presented and validated against experimental trial data. The ability of foot kinematics to adapt after changes to the footwear construction have been made in the model is then assessed by replicating a published clinical trial before the predictive capacity of the model is evaluated by applying the kinetic load conditions determined in the base model to the as yet unmodelled Springblade footwear construction.

8.1. Inverse Dynamics

When the kinematics and inertial properties of a body and external forces acting upon it are known, inverse dynamics can be used to indirectly determine the kinetics that are responsible for the observed motion of the body (Zatsiorsky 2002). The first analyses of human movement employing inverse dynamics was reported by Braune and Fischer (1895-1904) with a similar approach adopted by Elftman for his investigations into walking (1939a; 1939b) and running (1940). The first 3-D gait analysis employed Newton-Euler equations of motion and was reported by Bresler and Frankel (1950). The arrival of commercial force platforms and inexpensive computers in the 1970s spurred further research in the field with joint kinetics now reported for a wide range of human movements (Robertson et al. 2004). The fundamental principles of inverse dynamics including all necessary assumptions are presented
in this section with the single segment foot model used to determine the dynamic 3-D loading conditions employed in the Kinetic FE footstrike model also reported.

8.1.1. Fundamental Principles

Inverse dynamics is an established biomechanical technique which uses Newton’s Second Law and its angular equivalent to estimate the net forces and moments that occur at each of the joints in the body (Robertson et al. 2004). Whilst only a single segment foot model was employed for the analysis reported in this thesis, linked-body models with frictionless pin joints can also be used to determine joint kinetics in more complex multi-segment analyses.

The first stage of an inverse dynamic analysis is to use the principle of superposition to transfer all anatomical forces acting across a body to a common axis located at the most proximal point of the segment. These forces and the moments that occur as a result of them can then be summed to produce a single net force and net moment acting at the proximal joint centre for the defined segment. This process is illustrated hereafter with Figure 8.1a showing a free body diagram (FBD) of the anatomical forces acting on the foot and Figure 8.1b showing these forces resolved as a single net force and net moment acting at the ankle joint. The vertical force acting on the foot due to gravity and external GRF acting at the COP are also shown.
Assuming the inertial properties of the segment are known, Newton’s Second Law (Equation 8.1) can then be used to determine the magnitude of the net force acting at the joint

\[ \Sigma \vec{F} = m \vec{a} \]

*Equation 8.1*

where \( \vec{F} \) is the net force acting on the body when also considering the force due to gravity and external GRF, \( m \) is the mass of the segment and \( \vec{a} \) is the rigid body acceleration of the segment’s centre of mass measured with synchronised kinematic data acquisition techniques.

The net moment acting about the joint axis can be similarly determined with the angular equivalent of Newton’s Second Law (Equation 8.2)

\[ \Sigma \vec{M} = I \vec{\alpha} \]

*Equation 8.2*

where \( \vec{M} \) is the net moment acting about the segment’s proximal joint, \( I \) is the moment of inertia of the segment and \( \vec{\alpha} \) is the angular acceleration of the body about the joint centre.

The net force and moment acting at the defined segment joint centre can hence be estimated
although it is not possible to determine the contribution of each anatomical structure to these values purely with inverse dynamics. Further analysis using techniques such as coupled EMG can however provide some insight into what these values may be (Robertson et al. 2004).

Estimating the net force and moment that occur at a joint with inverse dynamics requires a number of assumptions to be made. These include the idealisation of all joints to be frictionless and that each segment behaves as a rigid body such that its length, distribution of mass and inertial properties remain the same (Zatsiorsky 2002). The validity of each these assumptions with respect to the model developed for this thesis is discussed in Section 8.1.2.

**8.1.2. Biomechanical Modelling**

A single segment foot model was defined in the Visual3D Standard biomechanical modelling software (C-Motion, Inc., Rockville, USA) from a shod static trial with the distal point defined as a marker placed on the second metatarsal head and the proximal ankle joint centre defined as a point halfway between the medial and lateral malleoli as suggested by Nair et al. (2010). Depending on the technique used, foot segment inertial parameters can vary greatly (El Helou et al. 2012) but generally do not have a significant impact on the estimated joint kinetics (Robertson et al. 2004). Segment inertial properties were thus defined by scaling the body weight of the subject following guidelines published by de Leva (1996) with the mass of the Reference Shoe worn during the modelled dynamic trial added to the foot segment mass.

Using this model, the triaxial net force and net moment components were output at the ankle for a rearfoot running trial performed in the Reference Shoe by Subject A (see Table 3.1) relative to the global coordinate system. Whilst the mechanical principles detailed previously in Section 8.1.1 were only presented for a 2-D planar analysis, full 3-D dynamic analyses can also be performed. Although, Alkjaer et al. (2001) reported near identical results for 2-D and 3-D methods when estimating joint moments during level walking, it was determined that a 3-D analysis would be necessary as no such evidence could be found for a running analysis.

When performing a 3-D analysis, joint kinetics can be computed using four different numerical methods - a Newton-Euler approach (Bresler & Frankel 1950), a Lagrangian approach (Whittlesey & Hamill 1996), with wrenches and quaternions (Dumas et al. 2004) or using generalised coordinates and forces (Silva & Ambrósio 2002) although Dumas et al. (2007) found only a 2 % difference between the methods at the ankle during walking gait. As such,
3-D ankle joint kinetics were output using the most commonly used method in biomechanical analyses, the Newton-Euler approach (Robertson et al. 2004), and can be seen in Figure 8.2:

![Figure 8.2](image)

**Figure 8.2** - Estimated ankle joint kinetics for running trial performed in *Reference Shoe*. a) Net force components, and b) net moment components.

In order to output the kinetics that occur at the ankle joint it was necessary to idealise the foot as a single rigid segment which rotates about an ankle joint with no rotational stiffness. It is necessary to make these assumptions when performing an inverse dynamic analysis but in reality the ankle is not frictionless (Günther & Blickhan 2002) and the deformation of the foot that occurs during the footstrike, primarily bending at the MPJ, indicates that the foot is neither rigid nor of constant length (Robertson et al. 2004). Furthermore, the presence of deformable soft tissue in the foot also means that the segment’s inertial properties are not constant. The strategy employed to overcome the limitations of the approach adopted to determine load conditions for the Kinetic FE footstrike model are presented in Section 8.2.1.
8.2. Finite Element Footstrike Modelling

The development of an FE footstrike model driven with exclusively kinetic load conditions is detailed in this section with model output validated against biomechanical data obtained from the experimental trial used to determine the defined load conditions.

8.2.1. Model Definition

Overview

The Kinetic footstrike model reported in this Section is in many ways similar to the Kinematic and Combined footstrike models presented in Chapters 6 and 7 respectively in that it contains a homogenous foot structure attached to the midsole and outsole of the Reference Shoe. The initial orientation of the foot and footwear geometries prior to contact were determined using the same methodology as reported in Section 6.1.1 and the same material models, contact definition and meshing strategies were all employed unless otherwise stated. The primary differences between the modelling approaches were in the representation of the human foot, foot-shoe contact interface and the defined loading conditions, each of which are detailed hereafter.

Constraints and loading conditions

The human foot model was meshed with linear tetrahedral elements that do not have rotational DOFs. To allow rotational load conditions to be applied it was therefore necessary to create a 6 DOF reference node in the assembly to represent the ankle joint centre. As in the biomechanical model used to estimate the loading that occurred at the ankle during the modelled experimental trial (see Section 8.1.2), this was located halfway between the medial and lateral malleoli and coupled to all surface foot nodes located within 30 mm of its location. To be consistent with the principles of inverse dynamics, no rotational stiffness was applied to the ankle joint. The plantar surface of the foot was tied to the midsole directly beneath the heelpad and MPJ.

In order to model the periods both immediately prior to and following contact with the force platform, the footstrike was modelled in three distinct phases - terminal swing (pre-contact), support, and initial swing (post-contact). During the support phase, the net force and moment components determined from the inverse dynamic analysis were applied to the reference node representing the ankle, whilst during the two swing phases, kinetic loads were applied to
all interior and exterior foot nodes so as to replicate the kinematics of the foot measured during the experimental trial. Gravity was also applied to the foot and footwear assembly throughout the analysis but was relatively insignificant in comparison to the kinetic loads applied at the foot and ankle. The three distinct phases of the analysis are illustrated in Figure 8.3 to demonstrate the relative weight of the foot and ankle loads applied during each footstrike phase.

![Figure 8.3](image)

**Figure 8.3 - Illustration of the relative weight of loading applied during each of the three phases of the Kinetic footstrike model.**

When using finite element theory to model a structural system it is often necessary to tune model inputs such that the result of analyses accurately match those measured during an experimental trial (DeVore et al. 1986). This was necessary when developing the Kinetic model and can be attributed to the limitations of the biomechanical foot model (see Section 8.1.2), sources of error in the 3-D calculation of ankle joint kinetics (i.e. errors in joint centre of rotation locations, force platform measurements, motion capture system measurements, and segment angle calculations due to skin movement artefact) (Riemer et al. 2008; Zatsiorsky 2002), and limitations in the FE model itself such as the homogenous foot structure, rudimentary plantar contact interface and simplified material models employed.

Whilst some manual tuning of the load conditions defined in an FE model is acceptable, some strategic rules were established to ensure that the loads applied were still recognisable as having been derived from an inverse dynamic analysis. The load applied to a node at any point during an FE analysis is the product of the load magnitude constant and the associated load amplitude value specified for that moment in the analysis (Abaqus 2012d). It was therefore decided that the magnitude defined for each load component could be altered but that the load amplitudes would be left unchanged. This meant that the loads applied were identical to
those estimated with the inverse dynamic analysis but multiplied by a scale factor where necessary. A single exception to this rule was made in that that the amplitude for the vertical ankle force could be adjusted. This was decided because applying a representative vertical load to the modelled footwear is of the utmost importance when evaluating a prospective footwear design. With these rules established, the loads defined in the model underwent an iterative tuning process to minimise the differences between the footstrike kinematics and kinetics output with the model and those measured from the experimental trial.

**Contact**

As was discussed for the Combined model in Section 7.2.1, the tangential COF assigned to contact interface between the footwear outsole and force platform was determined through experimental traction testing as described in Section 3.4. However, in addition to this a tangential COF of 0.6 (Zhang & Mak 1999) was assigned to the plantar foot-midsole interface to ensure that the shear stresses that occur at these surfaces were represented where not tied.

**Human foot modelling**

The foot model initially employed in the Kinetic footstrike models was identical to that used in the Kinematic and Combined approaches in that the geometry was taken from a 3-D scan of a foot prosthesis with the homogenous material model reverse engineered to provide sufficient constraint to the system. A more complex model of the human foot was not included in the analyses in order to maintain manageable solve times. Preliminary kinetic models however indicated that this approach was insufficient as the forefoot would plantarflex relative to the rearfoot during the first half of the stance phase, an action which led to the forefoot of the footwear impacting the force platform much earlier than observed in the experimental trial.

To overcome this issue a simple mechanical system was incorporated into the FE foot model. As seen in Figure 8.4, rigid plates were created at the rearfoot and forefoot to match the underlying foot geometry and tied to the outer surface of the foot model. The two rigid body reference nodes were then linked with a bushing connector that allows six DOF motion between the two bodies relative to one another. A connector stop condition was then applied in a locally defined coordinate system such as to prevent all plantarflexion of the forefoot relative to the rearfoot. A comparison of the forefoot-rearfoot joint angle with and without
the connector stop criterion to that measured in the experimental trial is also shown in Figure 8.5:

**Figure 8.4** - Simple mechanical system employing a connector stop condition to prevent unrealistic plantarflexion of the forefoot relative to the rearfoot.

**Figure 8.5** - A comparison of the sagittal plane joint angle of the forefoot relative to the rearfoot in the experimental trial and FE model with and without the connector stop criterion.

As seen in Figure 8.5, with the exception of the first 0.05 s of the footstrike, the angle of the forefoot relative to the rearfoot was similar in the FE model and experimental trial with maximum dorsiflexion angles of 8.7° and 7.7° respectively. This indicates that despite its simplicity, the mechanical system incorporated into the foot model was successful in preventing unrepresentative plantarflexion and that the stiffness of the material model assigned to the foot was appropriate. Figure 8.5 does however show that, as was remarked
upon when discussing the procedure used to orientate the foot and footwear relative to the force platform in Section 6.1.1, the subject’s forefoot was dorsiflexed relative to the rearfoot during the first 0.05 s of the experimental trial. This was not observed in the FE footstrike model as the modelled foot is passive rather than active and begins the simulation in a neutral position.

8.2.2. Results and Validation

The Kinetic FE footstrike model was validated against the foot marker kinematics, COP location and vertical GRF measured in the experimental trial used to determine the kinetic load conditions applied in the analysis. All results are presented for the support phase of the analysis only.

Foot kinematics

Predicted foot kinematics were compared against the sagittal plane kinematics of four markers attached to the foot during the experimental trial. These were located at the posterior calcaneus, ankle (halfway between the lateral and medial malleoli), second metatarsal head (MTH) and distal hallux. The results of this comparison are shown in Figure 8.6:
Figure 8.6 - A comparison of experimental and predicted foot kinematics in the sagittal plane. a) Vertical position against A-P position, b) time against A-P position, and c) vertical position against time.

Figure 8.6 shows that the FE model was generally successful in replicating the sagittal plane kinematics of the experimental trial for the four analysed markers, particularly in the vertical direction. The calcaneal and ankle rearfoot markers were however shown to progress too far in the anterior direction during the propulsive phase of gait, with a maximum discrepancy of 36.8 mm observed for the calcaneal marker after 0.18 s of the footstrike. Similar discrepancies were not seen at the forefoot indicating that the modelled foot was unrepresentatively compressed in the A-P direction. This can be attributed to the use of an overly compliant homogenous foot volume that does not contain any internal rigid structures to represent the bones of the foot.
The frontal plane kinematics of the foot were also validated by comparing the predicted and experimental rearfoot eversion angle throughout the footstrike. The results of this analysis are shown in Figure 8.7:

![Figure 8.7 - A comparison of experimental and predicted rearfoot eversion angle.](image)

Figure 8.7 illustrates that the Kinetic FE footstrike model was able to demonstrate good agreement to the experimentally calculated rearfoot eversion angle with a maximum discrepancy of $0.96^\circ$ observed.

**Vertical GRF**

It was discussed in Section 8.2.1 that the vertical ankle force was the only amplitude that was tuned during development the Kinetic footstrike model. This was permitted to ensure that an experimentally representative vertical load would be applied to the modelled footwear and Figure 8.8 shows that this was achieved with an average discrepancy between the experimental and predicted results of 84 N. The maximum discrepancy was 216 N which occurred between the passive and active force peaks.
Figure 8.8 - A comparison of experimental and predicted vertical GRF.

Centre of pressure

Figure 8.9 shows that the Kinetic footstrike model was capable of predicting COP output locations that demonstrated fair agreement to those observed during the experimental trial. The average residual between experimental and simulated output location was 12.6 mm (maximum of 28.5 mm) in the anteroposterior direction and 14.2 mm (maximum of 23.2 mm) in the mediolateral direction. As was discussed in Section 6.2.3, the first and last 0.01 s of the experimental footstrike were not included in the analysis due to the large measurement errors that can occur when the vertical GRF is small.

Figure 8.9 - A comparison of experimental and predicted COP location. a) A-P axis, and b) M-L axis.
8.2.3. Discussion

A novel FE footstrike model employing exclusively kinetic loading conditions has been presented in this section with load amplitudes determined from an inverse dynamic analysis of an experimental running trial. The FE model of the human foot was more complex than in previously reported models (see Chapters 6 and 7) in that a simple mechanical system was employed to prevent plantarflexion of the rearfoot relative to the forefoot not observed during experimental running trials. Whilst successful, this mechanical system would however be inappropriate for any footstrike action during which such flexion actually occurs. When considering frontal and sagittal plane foot kinematics, vertical GRF and COP location, the model demonstrated generally good agreement to experimental data although the distribution of loading was not quite as representative as achieved with the Combined modelling approach.

The development of the Kinetic footstrike model required an iterative tuning approach to be undertaken before the results reported in Section 8.2.2 could be achieved. A coarse mesh was employed on all foot and footwear structures to minimise the time it took to achieve a solution for each iteration of the analysis but this procedure remained time-consuming and would not be well suited to the short development timelines typical in the footwear industry if required for each new footwear design. The Kinetic footstrike model was however developed under the assumption that employing purely kinetic load conditions in an analysis would allow the kinematics of the foot and footwear to adapt to changes in the footwear assembly, thus negating the need to perform manual tuning of the model for each new footwear design. If successful, this would ensure that an accurate representation of experimental load magnitudes and distributions would always be applied, drastically improving the predictive capacity of such an approach. This hypothesis is evaluated in the remaining sections of this chapter.

8.3. Case Study: Motion Control Footwear

The predictive capacity of the Kinetic footstrike model was evaluated by determining if similar patterns in the changes to foot kinematics were predicted with the model as those published for experimental trials after orthotic intervention. The introduction of medially posted insoles designed to reduce foot eversion was modelled with the results primarily compared to average population data reported for a clinical trial performed by Mündermann et al. (2003) although the results of other similar studies are also considered.
8.3.1. Literature Review

Whilst controversial (see Section 2.2.4 for a full discussion), excessive foot eversion has long been associated with an increased risk of injury (e.g. James et al. 1978; Messier & Pittala 1988) with motion control devices such as medially posted insoles and orthotics commonly prescribed to control and limit this eversion. As seen in Figure 8.10, Mündermann et al. (2003) reported that the introduction of a 6 mm medially posted insole significantly reduced both the maximum eversion angle and maximum eversion velocity that occurred during experimental running trials. These results were supported by similar findings reported by Rodrigues et al. (2013) for a 4° medially wedged insole and by Milani et al. (1995) and O’Connor & Hamill (2004) for varus wedge footwear designed with midsoles 30 mm and 20 mm thick at the medial and lateral heel respectively.

![Figure 8.10 - Average experimental foot eversion angle for a single subject with and without medially posted insoles. Recreated from Mündermann et al. (2003).](image)

The study by Mündermann (2003) also investigated other kinematic variables, stating that the effects of medial posting were consistent across subjects but did not explicitly state that this was the case for maximum eversion angle and maximum eversion velocity. Rodrigues et al. (2013) were however able to confirm that the peak eversion angle was reduced for all 33 subjects tested and also reported that the time to maximum eversion angle was not significantly changed between footwear conditions, an assertion supported by O’Connor & Hamill (2004). Differences in the effect of medially wedged footwear on frontal plane touchdown angle were however reported with no change observed by O’Connor & Hamill (2004) but an increased inversion angle of the ankle joint centre measured by Rodrigues et al.
(2013). Milani et al. (1995) also reported a similar increase in the touchdown inversion angle but this was not found to be statistically significant.

8.3.2. Model Overview

As shown in Figure 8.11, the 6 mm medially posted wedge insole used in experimental trials by Mündermann et al. (2003) was replicated by defining skin reinforcements to the top surface of the footwear midsole. These were divided into two regions with the medial and central surfaces assigned shell elements 6 mm and 3 mm thick respectively. The lateral aspect of the midsole was unaltered with all shell elements assigned the standard EVA material model used in the footwear midsole. The relative coarseness of the midsole mesh meant that the medial post geometry could not be perfectly replicated but this approach was considered to provide a sufficient approximation. The foot was not repositioned in the model assembly and the model was otherwise identical to that described in Section 8.2.1.

Figure 8.11 - Illustration of skin reinforcements defined on the top surface of the midsole to approximate a 6 mm medially posted insole.

8.3.3. Results

The results of the analysis including the medially posted insole were compared to those predicted for the standard Reference Shoe assembly.

Sagittal plane kinematics

Sagittal plane marker kinematics were largely unaffected by the introduction of a medially posted insole with a maximum deviation from the results obtained for the standard footwear assembly of only 2.4 mm.
**Vertical GRF**

Vertical GRF was similarly unaffected with a 21.9 N maximum change in the predicted loading of the force platform, a value only 1.4 % of the maximum GRF observed.

**Foot eversion**

Significant changes in frontal plane foot kinematics were however predicted after the introduction of a medially posted insole with a 2.1° reduction in the maximum foot eversion angle and 12.6 % reduction in the maximum eversion velocity. This compares favourably with the experimental results published by Mündermann et al. (2003) who reported a 2.3° reduction in maximum foot eversion angle and 15.5 % reduction in maximum eversion velocity. Furthermore, in line with the data published by Mündermann et al. (2003), Rodrigues et al. (2013) and O’Connor & Hamill (2004), the predicted time to peak eversion was also unaffected.

![Figure 8.12 - A comparison of predicted foot eversion angle with and without medially posted insole.](image)

A comparison of the reductions in maximum eversion angle and maximum eversion velocity predicted to those published in the four experimental trial previously discussed in this section is presented in Table 8.1. Differences in the scale of the reductions observed could potentially be attributed to variation in the methodologies used to define the eversion angle of the foot and differences in the types of orthotics used.
<table>
<thead>
<tr>
<th></th>
<th>Max. Eversion (°)</th>
<th>Max Eversion Velocity (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>FE Simulation</td>
<td>2.1</td>
<td>12.6</td>
</tr>
<tr>
<td>Mündermann et al. (2003)</td>
<td>2.3</td>
<td>15.5</td>
</tr>
<tr>
<td>Rodrigues et al. (2013)</td>
<td>3.6</td>
<td>20.6</td>
</tr>
<tr>
<td>O’Connor &amp; Hamill (2004)</td>
<td>1.1</td>
<td>13.9</td>
</tr>
<tr>
<td>Milani et al. (1995)</td>
<td>5.0</td>
<td>25.0</td>
</tr>
</tbody>
</table>

Table 8.1 - A comparison of predicted and published reductions in maximum eversion angle and maximum eversion velocity in medially posted footwear.

In both the FE simulation and reported experimental trials (Mündermann et al. 2003) the introduction of a medially posted insole led to the foot being placed in a more inverted position during the first half of the footstrike. Whilst predicted with the model, a similar trend was however not observed in the experiment trials for the second half of the footstrike with no increase in the maximum inversion angle reported between footwear conditions. Mündermann et al. (2003) suggest that this may be because the contact area between the foot and insole is small during this phase of gait, thus limiting the effectiveness of the medial post. The foot-footwear contact interface in the FE model is simplistic with ties employed at both the rearfoot and forefoot and this might explain why an increased maximum inversion angle was predicted with the model.

8.3.4. Discussion

The FE footstrike model has been shown to well replicate the changes in frontal plane kinematics reported in published experimental trials with comparable reductions in maximum eversion angle and maximum eversion velocity predicted. Despite the changes made to the footwear assembly, the model was still able to ensure that experimentally representative load magnitudes were applied to the footwear with sagittal plane kinematics also remaining largely unchanged. This supports the hypothesis that an FE modelling approach employing exclusively kinetic load conditions has a greater predictive capacity than an alternative approach with defined kinematic boundary conditions. This was because the kinematics of the unconstrained foot were able to adapt to a newly defined footwear condition.

It must however be noted that the kinematic adaptations that occurred in the model were passive and significantly less complex than the active adaptations employed by a human subject to modulate the loads exerted in differing footwear constructions. For example, both Mündermann et al. (2003) and O’Connor & Hamill (2004) reported that ankle inversion
moments were significantly reduced in the medially posted condition whilst a change in the touchdown inversion angle was also reported by some authors (see Section 8.3.1). These active changes were not manifested between footwear conditions in the FE model as identical load conditions were always defined. Despite this limitation, the capability of a kinetically driven footstrike model to adapt its kinematics to different footwear conditions and the associated improvement in the predictive capacity of such an approach is clear.

8.4. Case Study: Springblade

The Springblade (see Section 3.1.2) was a novel footwear design launched by adidas in August 2013 which replaced the traditional foam midsole with a topplate formed of 16 individual leaf springs. During product development the response to loading of each of these leaf springs was typically predicted with an FE analysis in which each individual leaf spring was deformed via contact with a rigid plate on to which a vertical displacement boundary condition had been prescribed (see Figure 8.13). Whilst undoubtedly providing some insight into the performance of the footwear, a limitation of this approach is that the magnitude of the displacements prescribed is somewhat arbitrary. This means that the scale of loading that each leaf spring undergoes is not necessarily representative of an experimental trial.

![Figure 8.13 - Displacement driven FE analysis typically used by industrial collaborator to evaluate Springblade design with logarithmic strain visualised for the topplate only. a) Unloaded, and b) loaded.](image-url)
The ability of the Kinetic footstrike model reported in Section 8.3 to adapt to an entirely new footwear construction was therefore investigated in this section by applying the same load conditions determined for the Reference Shoe to an analysis performed with the Springblade footwear. The predicted deformation of each leaf spring was then compared to digitised HSV footage of experimental trials performed in the Springblade footwear with the predictive capacity of the model subsequently evaluated after changes to the topplate’s material stiffness had been made.

8.4.1. Model Definition

The topplate of the Springblade is constructed of a Nylon 12 polyamide supplied by Evonik Industries (Essen, Germany). A linear elastic material model is typically only valid for small elastic strains of up to 5 % (Abaqus 2012d) and, since strains larger than this were anticipated, a first order hyperelastic material model employing a Neo Hookean strain energy potential was used to characterise the material’s response to loading:

$$U = C_{10}(I_1 - 3) + \frac{1}{D_1}(J_{el} - 1)^2$$

Equation 8.3

where $U$ is the strain energy per unit of reference volume, $C_{10}$ and $D_1$ are material coefficients, $I_1$ is the first deviatoric strain invariant and $J_{el}$ is the elastic volume ratio (Abaqus 2012a). The material coefficients used to represent the topplate material where determined such that

$$C_{10} = \frac{4}{E(1 + \nu)}$$

Equation 8.4

$$D_1 = \frac{6(1 - 2\nu)}{E}$$

Equation 8.5

where $E$ is the Young’s modulus and $\nu$ is the Poisson’s ratio (Abaqus 2012a) with values for both parameters taken from a data sheet provided by the material’s supplier for its conditioned state. The blown rubber used in the outsole was represented with a hyperelastic third order strain energy function as reported in Section 4.2.3.
The rearfoot and forefoot of the Springblade topplate is manufactured in two halves so it was therefore necessary to apply a tie constraint to the adjacent surfaces of the two parts where they overlap at the midfoot. The Springblade model was then orientated to match the plantar surface of the foot with the whole assembly then translated vertically such that there was the same amount of clearance between the footwear and force platform as observed in the Reference Shoe model reported in Section 8.2. All loads were otherwise unchanged.

Initial analyses indicated that further tuning of the model would be required when modelling the Springblade footwear to ensure that kinematics and kinetics representative of an experimental footstrike would be simulated. As before, load magnitudes and initial conditions were adjusted but, with the exception of the vertical ankle force, the load amplitudes defined were as determined from the inverse dynamic analysis detailed in Section 8.1.2. In comparison to the development of the initial Reference Shoe Kinetic model, fewer iterations of the analysis were needed to achieve acceptable model output but each iteration took significantly longer to reach a solution due to the finer mesh required to characterise the complex geometry of the Springblade topplate.

8.4.2. Results

Model field output was compared to experimental HSV footage of a running trial performed in the Springblade footwear with the predicted deformations of each individual leaf spring validated against average data taken from the digitisation of footage recorded for five different trials. The methodologies employed to output the predicted deformation of each individual leaf spring with connector elements and determine the experimental equivalent through the digitisation of HSV footage are as detailed for the Titan Bounce TPU structures in Sections 5.2.4 and 5.2.6 respectively. Whilst impossible to experimentally validate, the predicted load exerted on each individual leaf spring was also output by requesting the total force due to contact pressure for the outsole surfaces attached to each leaf spring.

HSV footage

Generally good agreement was observed between model field output and the foot kinematics recorded in the HSV footage of each experimental trial performed in the Springblade footwear. Despite this, and as was also reported for the Reference Shoe model in Section 8.2.1, dorsiflexion of the forefoot in the first 0.05 s of the experimental trial was not replicated in the
FE model as the human foot model is passive and located in a neutral position. This discrepancy can be best observed in the first frame of Figure 8.14a. It should also be noted that the footstrike contact times recorded in the five experimental trials were similar to those predicted with the model with an average experimental contact time of 0.227 s (s.d. = 0.012 s) compared to 0.23 s in the FE model.
Figure 8.14 - A comparison of model field output and HSV footage to a single experimental trial. a) Lateral view, and b) posterior view.
**Leaf spring deformation**

Figure 8.15 and Figure 8.16 illustrate that the maximum leaf spring deformations predicted with the Kinetic FE footstrike are generally similar to the values determined through digitisation of the five experimental trials, particularly for the forefoot leaf springs. The average maximum deformation across the 16 measured TPU structures was 2.98 mm with the average predicted with the FE model only slightly less at 2.83 mm although an experimental value for the heel leaf spring could not be determined as it was not possible to accurately measure its deformation with the medially and laterally positioned cameras. The average absolute residual between experimental and predicted maximum deformations was 0.47 mm with 11 of the 16 predicted maximums falling within one standard deviation of the experimentally measured values.

Both the experimental HSV footage and FE model identified the trend that the lateral rearfoot leaf springs underwent the greatest deformations whilst the medial leaf springs deformed the most of those located at the forefoot. This indicates that the distribution of loading applied to the Springblade footwear with the Kinetic footstrike model can be generally said to be representative of that which would be applied during an experimental trial. This represents significant progress over the standard displacement driven analysis typically used by the industrial collaborator of this thesis and presented in the introduction to this section (see Figure 8.13).

![Figure 8.15 - A comparison of experimental and simulated values for maximum deformation of the Springblade leaf springs during a running footstrike.](image-url)
Figure 8.16 - A comparison of experimental and simulated values for maximum deformation of the Springblade TPU structures. Error bars represent one standard deviation.

A comparison of the experimental and simulated deformations is shown for six of the 16 measured leaf springs in Figure 8.17 which demonstrates that whilst not perfect, the deformation profiles predicted with the FE model are largely representative of those measured in the experimental trials. Potential explanations for the discrepancies that do exist are discussed in Section 8.4.4.
Figure 8.17 - A comparison of experimental and simulated deformation of the Springblade leaf springs. a) L1, b) M1, c) L3, d) M3, e) L6, and f) M6.
Leaf spring loading

Whilst no experimental trial data exists against which to validate the predicted loading of each individual leaf spring, Figure 8.18 and Figure 8.19 show that greatest loads were predicted to be applied at the heel, lateral rearfoot and medial forefoot leaf springs of the Springblade footwear. This was similarly reflected in the larger deformations predicted for these leaf springs as presented in Figure 8.15 and Figure 8.16.

Figure 8.18 - Area chart showing the predicted loading of each individual leaf spring and cumulative total load. Boxed numbers represent medial and lateral leaf spring pairs.
8.4.3. Evaluation of Predictive Capacity

It was shown in Section 8.4.2 that, after some tuning of load magnitudes, the Kinetic model was capable of applying experimentally representative loading to the Springblade footwear. As reported hereafter, the predictive capacity of this approach was subsequently investigated by increasing the stiffness of the material model assigned to the topplate and leaf springs.

Model overview

The revised Kinetic footstrike model was identical to the standard Springblade model reported in Section 8.4.1 except that the Young’s modulus of the material assigned to the rearfoot and forefoot topplate components was increased by 50%. Hyperelastic material parameters were recalculated accordingly (see Equation 8.4 and Equation 8.5).

As introduced in Chapter 1 and discussed throughout this thesis, to be of genuine use in the footwear industry an FE footstrike model must not only emulate the loading that occurs during an experimental trial and demonstrate predictive capacity, but also allow a footwear developer to quickly compare two footwear assemblies and make informed design decisions. A methodology was therefore developed to generate report files allowing for the quick comparison of the results predicted with the Kinetic footstrike model for two different footwear designs. This was achieved by first writing a custom Python script to extract all...
history output from the results file for each analysis and write the data to a plain text CSV file. Any two CSV files could then be imported into a pre-formatted Excel spreadsheet to automatically generate a wide variety of graphs comparing the performance of the individual leaf springs on each shoe and of the shoes as a whole. Key parts of the report file comparing the standard Springblade model and that with the increased topplate stiffness are presented hereafter:

**Results**

When designing the Springblade it was considered important to ensure that the topplate material was not strained above its elastic limit as this would lead to fatigue of the material and ultimately to failure. Figure 8.20 shows that the Kinetic model could be used to predict the strains that occur in the Springblade during a typical footstrike and thus determine if the magnitudes and distribution of material strain satisfy the design criteria. As is to be expected, higher levels of strain were predicted in the topplate for the standard material than when the stiffer material was employed.
Figure 8.20 - A comparison of logarithmic strains predicted to occur in the topplate of the *Springblade* footwear with the Kinetic footstrike model. a) Passive force peak, and b) active force peak.
A second design parameter of particular interest is the response to loading of each of the individual leaf springs. Combining predicted data for the deformation and loading of each leaf spring and plotting a linear trend line using a least squares method allows the linear stiffness of each leaf spring to be characterised under experimentally representative loading. This is illustrated for a single leaf spring in Figure 8.21 with a comparison of the stiffnesses for all the leaf springs in the standard and stiff analyses presented in Figure 8.22:

![Graph showing force-deformation profiles for M7 leaf spring](image)

**Figure 8.21 - A comparison of the standard and stiff force-deformation profiles for the M7 leaf spring located at the medial forefoot.**
Figure 8.22 - A comparison of leaf spring stiffnesses predicted with the Kinetic model for standard and stiff material models.

Figure 8.21 shows the typical relationship between loading and deformation for an individual leaf spring with the area contained within each curve equal to the amount of energy lost during the footstrike, primarily due to the material damping defined in the analysis. A linear trendline employing a least squares approach is also plotted for each leaf spring with the gradient of the line used to characterise its stiffness. Whilst the geometry of each leaf spring means that its response to loading is actually non-linear, this simple method for characterising the stiffness of each leaf spring still serves as a quick and easy technique for a designer to evaluate the behaviour of a prospective Springblade footwear design under experimentally representative loading.

8.4.4. Discussion

The inverse dynamic analysis performed for a running footstrike enabled Kinetic boundary conditions to be determined and experimentally representative loading to be applied to the Reference Shoe footwear used during experimental trials. It was shown in Section 8.3 that passive adaptation of the footstrike kinematics can occur in the model after simple changes to the footwear assembly have been made but further tuning is evidently required when applying the same loading conditions to an entirely new footwear construction such as the Springblade.
The development of a Kinetic FE footstrike model with generic load conditions capable of being applied to any footwear assembly was hence not achieved with the reported approach. This tuning procedure is time-consuming due to the increased analysis solve times associated with the complex geometries of the *Springblade* components and can thus be said to somewhat restrict the predictive capacity of the Kinetic modelling approach.

Despite this, when working under the assumption that a subject would exhibit a similar loading profile wearing the *Springblade* footwear to that measured when wearing the *Reference Shoe*, it was shown that the Kinetic footstrike model could be used to predict footwear performance after some tuning of load magnitudes. This was evidenced by an average leaf spring maximum deformation residual of only 0.47 mm in comparison to digitised HSV footage of experimental trials.

Aside from the error inherent in the digitisation of trial footage, discrepancies that do exist between the experimentally measured data and model output can primarily be attributed to the fact that model load conditions were determined from a trial performed in the *Reference Shoe* and then applied to the *Springblade* footwear. One must appreciate this limitation when using the Kinetic footstrike model predictively, as is the case for all *Springblade* analyses reported in this section.

After having determined that the Kinetic footstrike model is capable of applying experimentally representative loading to the *Springblade* footwear, it was also demonstrated that the model could be successfully used to compare the performance of two footwear designs. Important design criteria such as strain in the topplate during footstrike loading and the stiffness of each individual leaf spring were successfully output and compared. This represents significant progress over the traditional displacement driven FE analysis typically employed by the industrial sponsor of the research when evaluating prospective designs of the *Springblade*.

### 8.5. Chapter Summary

An FE footstrike modelling methodology employing exclusively kinetic load conditions was developed to avoid the limitations of previously reported models employing rigidly defined kinematic boundary conditions, namely the inability to adapt foot kinematics to changes in the modelled footwear assembly. The 3-D loads that are applied at the ankle during a running
footstrike were estimated through inverse dynamic analysis and used to load the foot and footwear during the support phase of gait.

The final stages of terminal swing and first stages of initial swing were also included in the analysis by applying kinetic load conditions to the entire foot such that the experimental kinematics of the foot were emulated. A connector stop condition was used to link the forefoot and rearfoot model and prevent plantarflexion of the forefoot that was not observed during the experimental trial. The reported modelling approach also required iterative tuning of the applied load conditions to be undertaken before satisfactory loading of the Reference Shoe footwear could be achieved but this procedure was not particularly time consuming due to the relatively coarse mesh employed in the model.

Predicted model output was validated against data obtained from the emulated experimental trial with good agreement observed, particularly for the vertical GRF which had an average residual of only 84 N, although this value was higher than the equivalent for the Combined model. Predicted model COP output locations were largely in line with the experimental readings whilst a maximum residual in the rearfoot eversion angle of 0.96° was observed. The maximum discrepancy between experimental and predicted sagittal plane marker kinematics was 36.8 mm although some unrepresentative compression of the modelled foot was observed along its longitudinal axis due to the simplicity of the homogenous structure employed.

The improved predictive capacity of the Kinetic footstrike modelling approach was demonstrated by resubmitting the analysis with the inclusion of a medially posted insole modelled with skin reinforcements defined on the top surface of the midsole. In comparison to a published clinical trial (Mündermann et al. 2003), similar changes in the reduction of the maximum eversion angle and maximum eversion velocity were predicted although it was conceded that the kinematic adaptations observed between footwear conditions were passive and did not wholly replicate the active adaptations employed by human subjects.

The predictive capacity of the Kinetic model was further investigated by applying the same load conditions to the Springblade footwear construction. Further time-consuming tuning of the applied loads was required before satisfactory model output could be achieved, indicating that it was not possible to develop a generic footstrike model applicable to all footwear.
designs with the reported methodology. Despite this limitation, the deflection of the Springblade leaf springs predicted with the model after tuning had been undertaken showed good agreement to digitised HSV footage obtained of experimental running trials performed in the Springblade footwear.

A methodology to quickly write report files comparing two prospective footwear designs was also demonstrated for a Springblade construction employing a stiffer topplate material with footwear strains and individual leaf spring stiffnesses output. In comparison to the kinematic models typically used to evaluate Springblade footwear designs, this approach was considered to better represent the loads applied to the shoe during a running footstrike. It can subsequently be concluded that the predictive capacity of the Kinetic footstrike model presented in this chapter is significantly greater than that of the models employing kinematic boundary conditions previously reported.
9. Optimisation

A number of footstrike modelling methodologies have been reported in this thesis which allow experimentally representative loading to be applied to a virtual footwear assembly, thus allowing informed design decisions to be made. The use of these models in the footwear development process would reduce the industry’s traditional dependence on physical prototyping but would still require a time-consuming manual design approach to be adopted. Automated optimisation techniques act to shorten development cycles by determining the optimum solution for a given set of loads, constraints and objectives (Abaqus 2012a). The potential value of implementing such procedures in the footwear development process is clear with a number of alternative approaches detailed and evaluated in this chapter.

Based on the Pedobarographic footstrike model reported in Chapter 5, a topology optimisation approach is employed to output novel lightweight footwear designs with optimum results presented for a number of different target volumes and load conditions. Modified designs output after the definition of geometric constraints are also discussed. An alternative shape optimisation approach used to minimise peak stresses in an updated Springblade construction, the Springblade Pro, is also detailed thereafter.

9.1. Introduction to Optimisation

9.1.1. Overview

A traditional, manual product development approach is time consuming and limits the number of novel designs that can be created due to the iterative nature of the methods employed. Furthermore, the designs generated are limited by the experience, ability and imagination of the developer (Altair 2010). The purpose of an automated design process is to use optimisation algorithms to determine the optimum structural configuration for a component given a specified set of loads and constraints (Ravindran et al. 2006). Provided the model and optimisation objective have been correctly defined, no further human intervention is required, thus allowing component designs to be quickly created and evaluated 24 hours per day. This results in a wider range of designs being investigated, improved product quality and reduces the time required to develop a new product (Altair 2010).
Shimoyama et al. (2011) reported a procedure developed to determine the optimum combination of materials employed in the midsole of an athletic shoe but, despite obvious applicability, no further studies on footwear optimisation have been reported to the author’s knowledge. Two novel methodologies developed to automatically output optimised footwear geometries are therefore presented in this chapter.

9.1.2. Optimisation with FEA

The automated design of optimised components is commonly achieved with the combination of optimisation algorithms and computerised numerical techniques such as FEA (Rao 2011). Component optimisation is achieved with Abaqus using ATOM (Abaqus Topology Optimisation Module), an optimisation plug-in which offers additional functionality and utilises the Abaqus/Standard non-linear solver (Abaqus 2012b). According to accompanying documentation (Abaqus 2012a; Abaqus 2012d), the following analysis components must be defined to refine an original design and output an optimised component geometry:

- **Design area**: The region of the model that can be optimised. This can be the entire model or a subset containing only selected regions.

- **Design responses**: The outputs from the base Abaqus analysis used as the inputs for the optimisation. These can either be read directly from the analysis report file or calculated from other data contained within it.

- **Objective functions**: The objective of the optimisation, typically to minimise or maximise a defined design response. Multiple objective functions can also be defined with a relative weight of importance for each objective specified.

- **Constraints**: Restrictions placed on a specified design response such as always being a certain value or not exceeding a defined threshold. Satisfying these constraints takes priority over the maximisation or minimisation of the objective function.

- **Geometric restrictions**: Geometric constraints independent of the optimisation defined to ensure that the optimised component geometry could be feasibly manufactured.
• **Stop conditions**: Determines when an optimisation is considered complete. Global stop conditions specify the maximum number of design iterations to be performed whilst local stop conditions instruct an optimisation to terminate once a specified minimum (or maximum) has been reached.

• **Optimisation task**: The optimisation task contains all of the optimisation components detailed in this section and specifies the optimisation algorithm to be used.

ATOM allows two different types of structural optimisation to be performed - topology optimisation and shape optimisation, each of which is introduced hereafter.

**Topology optimisation**

Topology optimisations utilise the “Material Distribution Method” (Bendsøe & Sigmund 2003) to determine the optimum distribution of material in a structure and are typically employed to output components optimised to be stiff but lightweight. The optimisation process acts to optimise the material properties of the elements in the design area such that the design objective is achieved whilst continuing to satisfy any defined constraints. The mass and stiffness of each element is coupled to its density meaning that the optimisation effectively removes some elements from the analysis by assigning them a near-zero density and therefore ensuring that they no longer contribute to the overall stiffness of the structure (Abaqus 2012a). The use of topology optimisation to output novel, lightweight footwear designs is detailed in Section 9.2 of this chapter.

**Shape optimisation**

Shape optimisation is used to refine the surface geometry of a component so as to reduce the occurrence and severity of local stress concentrations and thus achieve stress homogenisation. Adjusting the location of the structure’s surface nodes can cause the component mesh to become distorted so a mesh smoothing strategy must also be defined to maintain the quality of the mesh. Only the corner nodes of each element are modified with the displacement of midside nodes interpolated from the movement of each corner node (Abaqus 2012a). The use of shape optimisation to reduce stress concentrations and refine the geometry of a Springblade footwear design currently under development is detailed in Section 9.3 of this chapter.
9.2. Topology Optimisation

9.2.1. Background
First reported by Frederick et al. (1983), a number of studies have shown that adding mass to a runner’s shoes increases submaximal oxygen uptake by approximately 1% per 100 g per shoe. However, further studies have also suggested that shod running may provide an energetic advantage over a barefoot condition for habitual rearfoot runners (e.g. Kerdok et al. 2002). This suggests that the advantages of shoe cushioning may counteract the negative effects of the added mass (Franz et al. 2012). As a result of these findings, lightweight performance footwear that still offers some degree of cushioning represents a large segment of the running shoe market (Nigg 2009; Asplund & Brown 2005).

9.2.2. Optimisation Objective
Mass customisation is the concept of “producing goods and services to meet individual customer’s needs with near mass production efficiency” (Tseng & Jiao 2001) and was embraced relatively early by the footwear industry (Piller et al. 2012). Whilst brands have offered customers the ability to customise both the fit and aesthetic design of their footwear, little opportunity to similarly optimise footwear function has been made available (Pandremenos et al. 2010).

As such, this section details the use of topology optimisation techniques developed to automatically generate lightweight midsole geometries customised for an individual runner’s measured footstrike loading pattern. The objective of each optimisation was to achieve a design which maximised the cushioning offered for a runner’s personal load profile whilst ensuring that the volume of the midsole did not exceed a defined threshold, thus remaining lightweight.

9.2.3. Model Overview

Base model
A generic midsole geometry was created, meshed with linear hexahedral elements and assigned an isotropic linear elastic material model with a similar stiffness to the EVA hyperfoam model employed in previously reported models. 3-D footstrike load profiles were defined in a similar manner to that reported for the Pedobarographic model in Section 5.2.1
with triaxial loads determined using the “assumption of proportionality” (Bruening et al. 2010) and applied to 20 rigid plates tied to the top surface of the midsole. The rigid plates representing the sub-sections of the foot were constrained to prevent any translation in the A-P and M-L directions with the bottom surface of the midsole also constrained in all spatial DOFs.

Only the end state of each analysis step is considered in the optimisation when using ATOM so the maximum loads calculated for each foot region were applied with a ramp amplitude. This approach minimised the solve time for each individual analysis, something that is particularly important when developing a model for optimisation as the base analysis is run many times in succession. ATOM is only compatible with Abaqus/Standard so each analysis was submitted to the implicit solver.

![Figure 9.1 - Base model used in footwear topology optimisations. a) Undeformed, and b) deformed.](image)

**Optimisation task**

The objective function of the optimisation was to minimise strain energy whilst satisfying a constraint defined to ensure the midsole volume remained lower than a specified proportion of its original value. Before being input into the optimisation algorithm both the strain energy and volume design responses must be calculated as scalar values. The sum of the strain energy of all elements in the design area is a measure of the overall stiffness of the structure and can be calculated for linear models with Equation 9.1:

$$\Sigma u^Tku$$

*Equation 9.1*
where $u$ is the displacement vector and $K$ is the global stiffness matrix. The volume design response value is simply calculated by summing the individual volumes of all elements in the design area (Abaqus 2012a).

Figure 9.2 shows the typical variation of these two design response values over the course of an optimisation and indicates that the volume of the midsole remains approximately equal to the defined constraint value, in this case 0.3, whilst strain energy is consistently reduced until convergence is achieved and the optimisation terminates. This approach results in the development of midsole geometries optimised to be stiff and lightweight.

![Figure 9.2 - Variation of scalar design response values for an optimisation defined to minimise strain energy and maintain a relative volume of less than 0.3.](image)

ATOM allows two different types of optimisation algorithm to be used in a topology optimisation - general (sensitivity-based) and condition-based. The general algorithm is best described by Bendsøe and Sigmund (2003) and relies on the “method of moving asymptotes” (Svanberg 1987; Svanberg 2002) which breaks the optimisation problem down into a series of separable and convex sub-problems. Alternatively, the condition-based optimisation algorithm described by Bakhtiary et al. (1996) uses nodal strain energy and stresses as input data and is more efficient as the local stiffnesses of the design variables do not need to be determined. Despite this, the general optimisation algorithm was selected for the topology midsole optimisation reported in this section as it is more robust and can be used for a wider variety of problems (Abaqus 2012a).
The initial relative density of all elements in the design area was set to match the specified relative volume constraint with a standard density update strategy and maximum change per design iteration of 0.25 defined. Since the modelled problem is quasi-static, a solid isotropic material interpolation (SIMP) technique was defined which specifies an exponential relationship between the density and stiffness of an element (Abaqus 2012d).

The design area was identified to contain all elements contained within the midsole with the default values selected for all further optimisation parameters. No strategy for the deletion of elements was included in the optimisation approach reported so all elements in the design area therefore had a relative density of between 0.0 and 1.0. Global stop conditions were defined such that the optimisation would not exceed 50 design iterations although convergence was typically achieved after 35-45 designs had been simulated.

**9.2.4. Results**

A number of optimised midsole geometries are presented in this section with only elements of a relative density above 0.3 visualised.

**Volume constraint**

The volume constraint defined in an optimisation determines the maximum volume (and thus mass) of the component designs generated, with more lightweight structures output for a lower specified volume. A relative volume constraint of 1.0 would result in a midsole mass of 272 g assuming a standard EVA 55C was employed. Figure 9.3 shows the optimised midsole geometries output with three different relative volume constraints for a typical rearfoot running footstrike.
Figure 9.3 - Optimised midsole geometries output with different relative volume constraints. a) 70 % (190 g), b) 50 % (136 g), and c) 30 % (82 g).

Figure 9.3 shows that the optimisation strategy that results in the greatest minimisation of strain energy and thus the stiffest structure involves removing material in areas of low strain whilst retaining material in the regions of the midsole that experienced greater loading. Irrespective of the volume constraint defined, it is clear that material has been primarily retained under the calcaneus and metatarsal heads whilst the most aggressive removal of material was observed at the medial midfoot. For the optimisation in which a 30 % relative volume constraint was defined, this resulted in a 88 % reduction in predicted strain energy compared to the homogenous distribution of material defined for the base model.
Load conditions

Midsole structures output with a relative volume constraint of 30 % and optimised for different footstrike load profiles are shown in Figure 9.4. Load conditions were determined from experimental trials performed by Subject A (see Table 3.1).

Figure 9.4 - Optimised midsole geometries output with different footstrike load conditions. a) Rearfoot running, b) forefoot running, and c) lateral cut.

As before, the optimum distribution of material for each loading scenario was found to be achieved with an approach that retains material in highly loaded regions but removes it in areas that experienced lower strains. This is highlighted by the difference in optimised geometries output for rearfoot (Figure 9.4a) and forefoot (Figure 9.4b) running. Material is primarily retained at the heel and under the metatarsal heads for rearfoot running whilst
almost no material remains at the rearfoot for the forefoot striking load condition. The application of triaxial load conditions is particularly reflected in the geometry output for the lateral cut footstrike with the optimised structural members created displaying a clear medial-lateral skew to resist the significant lateral loads applied.

**9.2.5. Applying Geometric Restrictions**

As discussed in Section 9.1.2, ATOM allows geometric constraints to be applied to an optimisation to ensure that the structure output could be feasibly manufactured. Figure 9.5 shows a range of optimised geometries output for the rearfoot running model with a 30% volume constraint (Figure 9.4a) after a number of different geometric constraints had been defined:
Figure 9.5 - Optimised midsole geometries output with geometric constraints defined.  

- a) No geometric constraints, 
- b) top surface frozen, 
- c) vertical forge control, 
- d) mediolateral forge control, 
- e) 8 mm minimum member size control.
Figure 9.5b shows a midsole geometry generated after a frozen area had been defined to remove the top surface of the midsole from the design area and ensure the optimised midsole geometry generated remained as a single part. Figure 9.5c and Figure 9.5d show the geometries output after a vertical and mediolateral forge control constraint had been defined to prevent the creation of undercuts that would stop a structure from being removed from a mould although it is particularly clear in Figure 9.5d that this constraint has not been rigidly adhered to. Finally, minimum member size control was defined to generate the geometry shown in Figure 9.5e and prevent the generation of thin truss members.

Topology optimisation can be used to develop completely novel footwear designs but this process could not be considered as anything other than an interesting exercise to stimulate design ideas unless the geometries generated are actually manufacturable. The unconstrained structure presented in Figure 9.5a is clearly of too great a complexity to be manufactured with traditional moulding techniques but each of the midsole geometries shown in Figure 9.5b - Figure 9.5e have geometric constraints applied upon them and are simplified in some way. Whilst further work would certainly have to be done in this area before the midsoles created with automated topology optimisation procedures could be incorporated in a footwear design, the geometries presented in Figure 9.5 provide an insight into the type of structures that could be achieved with the implementation of geometric constraints.

9.2.6. Multistep Analyses

The models presented in Section 9.2.4 and Section 9.2.5 are optimised to minimise strain in the midsole under the triaxial loads that occur during a footstrike but do not consider loading that occurs due to relative motion of the different foot segments. To overcome this, a multistep analysis was created with additional steps defined to simulate the bending and torsion that occurs during a typical rearfoot running footstrike. This was achieved by applying moments at the midfoot and rearfoot rigid plates of a sufficient magnitude to emulate the kinematic deformations that were measured during the modelled experimental trial. As shown in Figure 9.6, each of the three steps modelled represented a different mode of deformation - triaxial plantar loading, torsion and bending.
Figure 9.6 - Different load conditions simulated in a multistep analysis. a) Unloaded, b) triaxial plantar loading, c) torsion, and d) bending.

A multi-step approach resulted in a “multi-objective optimisation” (Bartholomew 2012) and, whilst it is possible to define an objective function such that the loading at the end of each step is weighted differently, it was however decided the minimisation of strain under each mode of deformation would be granted equal priority.

A seen in Figure 9.7, the optimised midsole structures output with torsion and bending loads were largely similar to those without, indicating that plantar loading was the most dominant mode of deformation. Despite this, a notable difference between the two optimised structures was the additional material retained at the midfoot to provide extra torsional stiffness in the multistep analysis. A brief discussion on the appropriateness of maximising the bending and torsional stiffnesses of athletic footwear is presented in Section 9.2.7.
9.2.7. Discussion

Lieberman et al. have conducted a significant number of studies into different running footstrike styles and concluded (2010) that “…runners who forefoot or midfoot strike do not need shoes with elevated cushioned heels to cope with [the] sudden, high transient forces that occur when you land on the ground.” This suggests that footwear could be optimised to a runner’s individual footstrike pattern, only providing support under the areas of the foot that are loaded, with material omitted elsewhere. This could potentially result in the associated energetic advantages of a lightweight footwear design (see Section 9.2.2) whilst still providing cushioning where required.

The methodology presented in this section has demonstrated that topology optimisation could potentially be used to achieve such a goal, automatically generating novel footwear geometries optimised for a subject’s individual footstrike loading pattern but still remaining lightweight. Geometric constraints were also subsequently implemented in the optimisation procedure to encourage the generation of geometries that could be feasibly manufactured although significant further work would still be required in this area before manufacture with traditional moulding techniques could be achieved.
Furthermore, a fundamental drawback of applying topology optimisation to the design of footwear midsole components is that the stiffest structure is considered optimal (Nishiwaki et al. 1998), thus neglecting the primary role of the midsole, to dissipate footstrike impact forces. For the multi-step analysis presented in Section 9.2.6, this approach also resulted in a geometry being generated which was optimised to maximise the rigidity of the midsole to midfoot torsion and bending at the MPJ, not properties that are generally considered desirable.

Based on the work of Sigmund (1997) and Frecker et al. (1999) into the design of compliant mechanisms with topology optimisation, a more advanced midsole optimisation methodology was developed to overcome these issues with minimum displacement constraints defined for each of the 20 foot regions. The requirement of satisfying 20 displacement constraints and one volume constraint whilst simultaneously optimising for stiffness however caused the analysis solve time to become unmanageable. Further work is clearly required in this area to allow for the development of optimised custom midsoles that still maintain some degree of engineered compliance.

**9.3. Shape Optimisation**

**9.3.1. Optimisation Objective**

Shape optimisation is used to refine the surface geometry of a component to prevent the occurrence of stress concentrations and limit the effects of material fatigue. A shape optimisation procedure was developed and applied to the topplate component of the Springblade Pro, an under development update of the Springblade footwear constructed with only 14 leaf springs. A 3-D rendering of the Springblade Pro is shown in Figure 9.8:
9.3.2. Model Overview

**Base model**

The loading applied to the single piece *Springblade Pro* topplate was similar to that modelled for the topology optimisation presented in Section 9.2.3 with 3-D concentrated forces calculated for a rearfoot running footstrike and applied over 20 discrete regions of the foot with a ramp amplitude. The outsole component was not modelled but the part of each leaf spring where the outsole piece would be attached was constrained such that no vertical translation or rotation about the mediolateral and anteroposterior axes was permitted.

The topplate geometry was assigned the same Neo Hookean hyperelastic material model as reported for the standard *Springblade* geometry in Section 8.4.1 and meshed with linear tetrahedral elements. As with the topology optimisation, the base model was solved with the iterative Abaqus/Standard solver. Figure 9.9 shows the model in its unloaded and loaded states. From this it is apparent that the shorter forefoot leaf springs are somewhat overconstrained as relatively little deformation occurs in this area and this must be taken into consideration when analysing the optimised topplate geometry presented in Section 9.3.3.
**Optimisation task**

The objective function of the optimisation was to achieve stress homogenisation and prevent stress concentrations by minimising the peak von Mises stress predicted. The volume of the structure was constrained to remain constant to prevent an increase in component mass. The top surface of topplate was removed from the design area but all other nodes were defined as eligible to have their spatial location refined by the default condition-based optimisation algorithm (Bakhtiary et al. 1996).

The model contained approximately 55 000 nodes and 220 000 elements so a constrained Laplacian mesh smoothing algorithm was selected as it is most efficient for large models (Abaqus 2012d). A high target mesh quality was requested with the mesh smoothing region specified to encompass all elements in the model. Finally, global stop conditions were defined such that the optimisation would terminate after no more than 30 design iterations although an optimised solution was typically achieved after 10-15 design loops.

**9.3.3. Results**

The refined Springblade Pro geometry output with the shape optimisation approach had the same mass as the original geometry modelled but exhibited a 25 % reduction in the peak von Mises stress predicted. A 20 % and 31 % reduction in peak logarithmic strain and overall strain energy were also predicted respectively. Whilst difficult to visualise, this was achieved by increasing the fillet radii of the leaf springs and redistributing material such that the leaf springs were thicker towards the top, where the largest von Mises stresses were predicted.
Figure 9.10 - An example of a leaf spring geometry optimised to minimise the peak von Mises stress. Refined geometry is shown translucent.

The refinement of the topplate geometry resulted in global stiffening of the Springblade Pro’s individual leaf springs with, for example, the predicted deformation of the heel leaf spring being reduced from 4.0 mm to 3.5 mm under the same loading conditions. As can be seen in Figure 9.11 for the heel leaf spring, the stress homogenisation and general stiffening of the model achieved with the shape optimisation resulted in a clear reduction in the strains predicted for the optimised geometry.

Figure 9.11 - A comparison of logarithmic strains predicted to occur in the of the Springblade Pro topplate. a) Original geometry, and b) optimised geometry. Heel leaf spring shown only.
9.3.4. Discussion

Whilst the *Springblade Pro* topplate geometry output with the reported shape optimisation procedure was not particularly aesthetically pleasing, the optimised geometry successfully reduced stress concentrations and would therefore be predicted to experience less material fatigue under typical rearfoot running footstrike loading. The forefoot leaf springs were however overconstrained in the model and this resulted in the refinement of the topplate geometry being more pronounced at the rearfoot leaf springs due to the higher stresses and deformations predicted here. To overcome this issue a more complex modelling approach would be required to constrain the bottom to the *Springblade Pro* topplate in a more experimentally representative manner although this would have an associated impact on the computational expense of the optimisation.

A further drawback of the reported approach was that the predicted deformation of the individual leaf springs was decreased with the optimised geometry, thus indicating that the overall compliance of the topplate had been reduced and that it would be less successful at attenuating footstrike loading. As was the case for the topology optimisation reported in Section 9.2, the definition of additional minimum displacement constraints would ensure that a desired level of engineering compliance was maintained in the optimised structure whilst still minimising the occurrence of stress concentrations. This approach could also have applications in the automated functional grading of footwear, the currently manual process based on anthropometric data used to ensure that footwear of different sizes have appropriate mechanical properties (Kleindienst et al. 2006).

9.4. Chapter Summary

Based on the Pedobarographic footstrike model reported in Chapter 5 of this thesis, two automated optimisation methodologies were developed and used to output optimised footwear component geometries. Implementing the use of such automated procedures in footwear development would help with the creation of novel footwear designs and reduce the lengthy and therefore expensive timelines traditionally associated with traditional, iterative footwear development processes.

A topology optimisation approach utilising the “Material Distribution Method” was first reported and shown to generate lightweight midsole geometries optimised for different
footstrike loading conditions. Structures obtained after the specification of simultaneous geometric constraints defined to ensure the feasible manufacture of the novel geometries generated were also presented. Finally, a novel midsole geometry created with a multi-objective optimisation incorporating plantar loading, torsion and bending was created with equal weighting defined for each mode deformation.

A shape optimisation methodology was also developed and used to refine the surface geometry of an updated Springblade footwear construction currently under development, the Springblade Pro. Stress homogenisation was achieved with a 25% reduction in the peak von Mises stress for a model of the same mass although the predicted deflection of the forefoot leaf springs was low due to overconstraint of the model.

Despite representing obvious progress in the area of automated footwear development, a drawback of both the topology and shape optimisation procedures presented was that no control of the optimised structures’ global stiffness was facilitated, thus neglecting one of the primary functions of the components modelled, to attenuate footstrike loading. It was therefore discussed that the implementation of additional displacement constraints in future models would enable a desired degree of compliance in the optimised geometries to be engineered. This has potential further applications to the functional grading of athletic footwear.
10. Conclusions and Further Work

10.1. Thesis Conclusions

A range of FE footstrike models have been developed and shown to be capable of applying experimentally representative load profiles to prospective footwear designs, thus allowing product performance to be evaluated in a virtual environment. A number of case studies have been reported to demonstrate the feasibility of using these models in the footwear development process with techniques for automated product optimisation also presented. The novel footstrike modelling methodologies reported in this thesis are now being integrated into the footwear development processes employed by the project’s industrial collaborator, indicating the industrial value of the models created.

The criteria for evaluating each of the developed footstrike models were first introduced in Section 1.1. These are the biomechanical representativeness of the loads applied, the predictive capacity of the model, the feasibility of applying the methodology to other footstrike types and the ease with which the methodology could be implemented in the footwear development process. A direct comparison of the four reported models is presented in Table 10.1, with the extent to which each modelling approach satisfies these criteria discussed thereafter.
### Conclusions and Further Work

#### Input Boundary Conditions
- Pedobarographic Model (Chapter 5)
  - Plantar pressure distribution
  - Triaxial GRFs
- Kinematic Model (Chapter 6)
  - 6 DOF foot segment kinematics
- Combined Load Model (Chapter 7)
  - 6 DOF foot segment kinematics
  - Vertical GRF
- Kinetic Model (Chapter 8)
  - Ankle loads determined from inverse dynamics analysis

#### Primary Validation Data
- Pedobarographic Model (Chapter 5)
  - Deformation of TPU structures
- Kinematic Model (Chapter 6)
  - HSV footage
  - Vertical GRF
  - COP location
  - Footwear stiffness
- Combined Load Model (Chapter 7)
  - HSV footage
  - COP location
  - Footwear stiffness
- Kinetic Model (Chapter 8)
  - HSV footage
  - COP location
  - Sagittal plane kinematics
  - Frontal plane kinematics
  - Deformation of *Springblade* leaf springs

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**Table 10.1** - A comparison of the reported footstrike modelling methodologies.
10.1.1. Pedobarographic Model

The Pedobarographic model was reported in Chapter 5 and is the simplest of the FE footstrike models presented in this thesis but still capable of applying triaxial load magnitudes highly representative of a shod human footstrike. New footwear designs can be quickly implemented in the model whilst load profiles representative of alternative footstrike types could also be applied with ease. The simplicity of the methodology and its relatively short solve time also mean that it is the most appropriate model to be used with automated optimisation procedures such as those reported in Chapter 9.

Despite these strengths, as was reported in Sections 5.2 and 5.3 when modelling the Titan Bounce footwear, highly accurate predictions of experimental deformation could not be achieved with the Pedobarographic model. Furthermore, no bending or torsion of the footwear is included in the analysis. As a result of these limitations it can thus be concluded that the Pedobarographic model is the least biomechanically representative of the modelling approaches reported and is best used as a quick and convenient comparative tool for evaluating footwear performance.

10.1.2. Combined Load Model

The Kinematic approach introduced in Chapter 6 evolved into the Combined model reported in Chapter 7 with some of the major limitations of the Kinematic model overcome by defining simple kinetic load conditions. As such, only the Combined footstrike modelling approach is considered here.

The Combined modelling methodology was found capable of applying highly representative triaxial load magnitudes although uncertainty in the initial orientation of the footwear geometry led to divergence from the experimentally measured COP values during terminal stance. Despite this, the Combined model is the strongest of the methodologies reported at applying loading representative of a previously performed biomechanical trial.

Furthermore, whilst the time required to calculate foot segment boundary conditions and prepare the model for each new footstrike type is significant, the lateral cut analyses reported in Section 7.2 demonstrate that the methodology can be applied to footstrike types other than rearfoot running. The predictive capacity of the model was also investigated in Section 7.3 and
found to be considerable, provided that alterations to the modelled footwear were modest. However, complex adaptations to foot kinematics that occur as a result of changes in the footwear construction cannot be represented with the reported methodology.

10.1.3. Kinetic Model

As was reported in Chapter 8, development of an FE footstrike model employing exclusively kinetic load conditions required a significant period of manual tuning before experimental kinematics and load profiles could be emulated. Further tuning of the model was also required when adapting it for use with the Springblade footwear. Despite this limitation, loading applied with the model was shown to be largely representative of that measured during experimental trials and, whilst a significant period of model pre-processing would be necessary, the methodology could feasibly be applied to footstrike types other than rearfoot running.

The primary advantage of the Kinetic footstrike model is that the absence of kinematic boundary conditions allows the kinematics of the foot and footwear to adapt to changes in the modelled footwear construction and ensure that biomechanically representative loading is applied. It can thus be concluded that the predictive capacity of the Kinetic footstrike model is the greatest of those reported in this thesis.

10.2. Suggestions for Further Research

Further areas of study that it is felt would most contribute to achieving the research objectives of this project are presented here including a discussion of any preliminary work completed in these areas where appropriate.

10.2.1. Human Foot Modelling

As was discussed in Section 2.4.1, a significant body of research exists on the topic of finite element modelling of the human foot. As the focus of this project was on evaluating footwear performance, no consideration of the internal loads that occur in the foot was necessary. Despite this, the homogenous foot structure utilised in the reported footstrike models proved to be a limitation, particularly for the Kinetic modelling methodology. Preliminary work on a more complex human foot model developed to overcome this limitation is briefly discussed hereafter:
As shown in Figure 10.1, the preliminary human foot model is based on the geometry of a last and contains three rigid segments - the rearfoot, midfoot and forefoot. Flexion would be possible between the forefoot and midfoot segments whilst torsion would be defined to occur between the midfoot and rearfoot segments. Rotational stiffness behaviour would be determined either from literature (Stefanyshyn & Nigg 1997; Oleson et al. 2005) or inverse dynamic analysis. Furthermore, vertical translation of the rearfoot relative to the midfoot would also be enabled with the stiffness of the interface reverse engineered to emulate the longitudinal arch deformation measured by Bandholm et al. (2008) during balanced standing and walking gait. It is believed that the implementation of a more biomechanically representative foot such as this would improve the validity of the results predicted with subsequent footstrike models.

10.2.2. Full-Field Strain Validation

As was demonstrated in Sections 5.2 and 8.4, the deformation predicted to occur with an FE footstrike model can be validated by comparison to digitised HSV footage. However, only the deformation that occurs at each of the individual points selected can be quantified with this technique as the distribution of strain across the footwear components is not captured during the experimental trial. A number of optical techniques such as digital speckle pattern interferometry (DSPI) (Gröning et al. 2012) and digital image correlation (DIC) (Tarrier et al. 2010) have been used to validate finite element output with quantitative full-field strain comparisons. Applying such techniques to the development of athletic footwear would allow the distribution of strain predicted with an FE footstrike model to be more thoroughly validated than with the methods reported in this thesis.
10.2.3. Modelling of the Plantar Contact Interface

The contact interface between the footwear and the plantar surface of the foot is of key interest to footwear designers as the perception of comfort has been well correlated to the minimisation of plantar peak pressures (Hennig et al. 1996; Shorten 2009). Whilst not previously reported in this thesis, the Kinetic model was used to predict the plantar pressures that occur under the MPJ when wearing the Reference Shoe during rearfoot running. Results were promising with the peak pressure of 289 kPa predicted with the model only 5 % greater than the maximum value of 275 kPa measured during the modelled experimental trial. Literature sources also reported comparable plantar peak pressures in this region of 369.6 kPa (Chuckpaiwong et al. 2008), 362.0 kPa (Tessutti et al. 2010) and 369.7 kPa (Hong et al. 2012) and whilst these values are higher than those predicted with the Kinetic model, this can be at least partially attributed to the relatively low mass of the subject used during experimental testing.

The potential of using the Kinetic model to predict plantar peak pressures was further investigated with the Springblade Pro footwear construction by increasing the thickness of the modelled topplate using skin reinforcements. Although no experimental data exists against which to validate model output, as expected peak plantar pressure was found to be inversely proportional to topplate thickness with a 28.6 % reduction in the maximum value predicted when increasing the topplate thickness by 2 mm. Whilst these results are promising, the capacity to use the Kinetic model to predict plantar pressures is limited by the simplicity of the tie constraint linking the foot and footwear and is a poor representation of the complexity of this interface. Replacing these tie constraints with the definition of more representative contact behaviour or by including the upper in the footwear model would significantly increase the potential of using such an analysis to better understand the interaction that occurs at this interface.

10.2.4. Neuromuscular Control

The Kinetic footstrike model reported in Chapter 8 demonstrated the ability to adapt foot and footwear kinematics to the footwear conditions defined in the model. It was however conceded that whilst a human subject actively alters the loads that occur within the body when wearing different footwear (e.g. Mündermann et al. 2003; O’Connor & Hamill 2004) the loads defined in the FE footstrike model remained unchanged. The kinematic adaptations that are
predicted to occur with the model as a result of changing footwear conditions can therefore only be considered to be passive.

A musculoskeletal model incorporating adaptive neuromuscular control based on the principle of optimising the metabolic efficiency of locomotion was developed by Ackermann & van den Bogert (2010). This model was subsequently coupled to a 2-D plane strain FE model of the human foot by Halloran et al. (2010) with convergence to realistic gait kinematics achieved when also defining the minimisation of foot soft tissue deformation as an objective of the movement optimisation. A later study by van den Bogert et al. (2012) also reported that changes in muscle peak force and activation patterns were predicted by the musculoskeletal model when adding mass at the foot. Coupling a similar optimal control system to the Kinetic footstrike model would allow active adaptation of the simulated loads to occur as a response to altered footwear conditions. This would significantly improve the predictive capacity of the Kinetic footstrike model as reported in this thesis.

10.2.5. Footwear Optimisation

The development and implementation of automated optimisation techniques remains in its infancy (Bartholomew 2012), particularly in the sporting goods industry. It is however anticipated that these techniques will be employed ever more frequently in the footwear development process as the methods are refined and become increasingly effective. One such refinement that could be used with the optimisation methodologies reported in Chapter 9 of this thesis would be to replace the 20 segment foot mask employed to apply plantar loading with the definition of an analytical mapped pressure field. This allows the loading that occurs at any number of points in the 3-D workspace to be defined in analysis. As shown in Figure 10.2, the loading applied in between each of these defined points is then determined through interpolation, resulting in the definition of a continuous pressure field. In comparison to the 20 plate mask used in the optimisation models reported in Chapter 9 of this thesis, this would result in a much higher resolution of loading and ultimately lead to the improved generation of optimised footwear designs.
10.3. Final Statement

This thesis is the first to report dynamic FE footstrike models that are able to apply highly representative kinematics and kinetics to a prospective footwear design. All model outputs were validated against experimental motion capture trials, thus ensuring the appropriateness of the loads applied. The analysis methods reported in this thesis have since been adopted in the footwear development processes of the thesis’s industrial sponsor. This has led to a reduction in the traditional dependence on physical prototyping and will ultimately contribute to an improvement in the quality of the footwear products created.
11. References


Asics, 2010. Running Footwear. , 2010(24th March). Available at:  


Bandholm, T. et al., 2008. Foot Medial Longitudinal-Arch Deformation During Quiet Standing and Gait in Subjects with Medial Tibial Stress Syndrome. The Journal of Foot and Ankle Surgery, 47(2), pp.89–95. Available at:  

Bartholomew, P., 2012. The state of current practice in engineering design optimisation, NAFEMS.


GOM, 2005. ATOS User Information, Braunschweig: GOM mbH.


Lafortune, M.A., 2008. The role of research in the development of athletic footwear. Journal of Foot and Ankle Research, 1 (Suppl.).


References


References


Vicon, 2007. *Vicon MX Hardware System Reference Revision 1.7*,


