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BIOMECHANICAL ANALYSIS OF HEEL PAD TISSUES

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ABSTRACT

The biomechanical behaviour of the heel pad tissues, including the calcaneal fat pad and skin, is investigated by means of a combined experimental and computational approach, for the definition of numerical models of the biological structures. The constitutive models are formulated starting from the analysis of the complex micro-structural configuration of the tissues and evaluating the relationship between tissue histology and mechanical properties. A visco-hyperelastic model is formulated with regard to the calcaneal fat pad. The model takes into account the typical features of the mechanical response, such as large displacement and strain, non-linear elasticity and time-dependent effects. A fiber reinforced hyperelastic model is provided to interpret the skin mechanical response, such as anisotropic configuration, geometric non-linearity, and non-linear elasticity. In order to define the numerical models, experimental data from mechanical tests are analyzed to achieve information about the tissue mechanical response and to evaluate the constitutive parameters. The definition of constitutive parameters is performed using a specific procedure based on the comparison of experimental data and model results through a cost function. The minimisation of the cost function is performed by a stochastic-deterministic optimization algorithm, leading to the definition of the optimal set of parameters. The first step is performed by considering data from in vitro and in vivo experimental tests, in order to evaluate a preliminary set of constitutive parameters that describe the stress-strain relationship under uni-axial tests. The second step involves the analysis of an in situ test on the heel pad, with and without skin. The comparison of data from in situ tests and numerical results leads to an optimal domain of parameters which can interpret the mechanical response of real heel pad tissues. The procedure is validated by considering experimental data from additional in situ and in vivo experimental tests. The numerical models developed represent the basis for the interpretation of the physiological behaviour of the heel pad, considering the effects induced by specific loadings, and for the evaluation of the interaction phenomena between the foot and footwear.
SOMMARIO

Lo studio del comportamento biomeccanico dei tessuti molli del retro piede, intensi come tessuto adiposo plantare e pelle, ha richiesto un approccio fortemente integrato di tipo computazionale e sperimentale. La valutazione del comportamento meccanico di tale regione è ottenuta in considerazione dell’analisi della configurazione micro strutturale dei tessuti e valutando la correlazione tra configurazione istologica e comportamento meccanico. Al fine di analizzare aspetti tipici della risposta meccanica del tessuto adiposo calcaneare, come la non linearità per geometria e materiale, e la dipendenza dal tempo è stato utilizzato un modello costitutivo di tipo visco-iperelastico. Gli aspetti tipici della risposta meccanica della pelle, ovvero la non linearità per geometria e per materiale, e la configurazione anisotropa sono stati descritti mediante la formulazione di un modello costitutivo di tipo iperelastico fibro-rinforzato. La valutazione dei parametri costitutivi ha richiesto lo sviluppo di modelli analitici e numerici capaci di interpretare le prove sperimentali considerate. I risultati di modello ed i risultati sperimentali sono stati confrontati attraverso una funzione costo, la cui minimizzazione ha portato alla definizione dei parametri oggetto dello studio. La procedura è stata conseguita in step successivi. Una prima valutazione dei parametri costitutivi è stata ottenuta considerando prove monoassiali, sviluppate su campioni dei tessuti in esame. Successivamente sono state considerate delle prove sperimentali su piede cadaverico in assenza ed in presenza della pelle. Questa analisi ha portato alla definizione di un dominio di parametri costitutivi tutti in grado di interpretare in maniera adeguata la risposta meccanica dei tessuti in esame. La procedura adottata è stata infine validata considerando ulteriori prove sperimentali in situ ed in vivo. I modelli numerici sviluppati permettono di valutare la risposta meccanica dei tessuti in condizioni fisiologiche, come ad esempio durante la camminata, nonché di valutare i fenomeni di interazione tra piede ed elementi biomedicali, come solette e calzature.
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INTRODUCTION

The musculoskeletal system is made up by the skeleton, muscles and joints. The foot provides a stable basis for standing, dampens the stress and adapts to uneven ground in bare foot and shod conditions.

This work focuses on a particular anatomical region of the foot: the heel pad. The heel pad tissues, composed of calcaneal fat pad and skin, surround the calcaneus in the rear foot plantar region. Their conformation and mechanical properties strongly influence body posture during static standing and the gait cycle. Specifically, the human heel pad acts as an interface between the calcaneus and the ground at the heel strike during the gait cycle and running, absorbing shock and distributing deformational effects in the surrounding foot tissues. A specific knowledge of the mechanical behaviour of these tissues contributes to an investigation of the mechanical function of the foot, by considering the interaction phenomena between the foot and surrounding structures, such as orthosis and/or footwear products.

A biomechanical investigation of the heel pad can be performed through experimental and numerical approaches. Detailed information about the structural and micro-structural configuration of the tissues, together with data from mechanical tests, is necessary to define the proper constitutive formulation. Such data are important for developing and validating computational models that entail an appropriate constitutive formulation for heel pad tissues.

In Chapter 1, details of the anatomic configuration of the foot and its movements are presented. Attention is focused on the heel pad tissues, including the calcaneal fat pad tissue and skin, which determine the mechanical properties of the heel region. A review of the experimental tests computed on the heel region is reported to define the overall mechanical response of the tissues under different loading conditions, such as weight bearing and walking.

In Chapter 2 the histology and morphometry of soft connective tissues, together with information regarding mechanobiological models and general mechanical properties, are reported. Attention is focused on the calcaneal fat pad tissue and skin, as a basis for the
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constitutive analysis presented in this thesis. As mentioned, data from experimental tests are necessary to validate and enforce the mechanical assumptions from histological and morphometric analysis. This involves experimental activities that investigate the structure of the heel pad tissues and data from in vitro, in situ and in vivo mechanical tests. Numerical models allow to evaluate phenomena that are not accessible with corresponding accuracy by experimental tests. To define a numerical model it is necessary to create virtual solid models of the structures of interest and to formulate the constitutive models of the different tissues as presented in Chapter 3. Virtual solid models are defined by the analysis of biomedical images and represent the basis for the development of finite element models. The application of mathematical and physical laws, such as the axiomatic theory of constitutive relationships, are used to formulate constitutive models. Here, hyperelastic and visco-hyperelastic formulations are proposed. Following the mathematical definition of a constitutive relationship, constitutive parameters must be evaluated. The strong non-linearity of the soft tissue mechanical behaviour requires the evaluation of constitutive parameters by directly comparing the model results with the experimental data. The discrepancy between experimental data and model predictions is usually defined by a cost function, whose minimization leads to the definition of parameters. Stochastic-deterministic optimization algorithms are often adopted for this purpose. In Chapter 4 a specific procedure for the evaluation of constitutive parameters is reported and applied to the calcaneal fat pad tissue and skin. The procedure for the identification of the constitutive parameters is described in detail and involves the evaluation of a preliminary set of values that consider in vitro and in vivo tests, to be subsequently updated with additional in situ experimental data. This is achieved by an algorithm capable of identifying the region of minimal discrepancy between experimental and numerical model results. The validation of the constitutive model and its parameters must be performed by developing the numerical analysis of experimental situations that are not considered in the evaluation of the parameters. Indeed, experimental data from further in situ and in vivo tests are considered and compared with numerical results.

In Chapter 5 numerical analyses of the heel region are presented. The experimental approach can be investigated and integrated with a computational model of the heel pad. Numerical analyses are developed by considering in vivo tests on subject-specific heel pad. The accordance between the numerical and experimental results confirms that the
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The constitutive analysis proposed is accurate and reliable. Finally, some notes are reported about the investigation of the interaction phenomena occurring between the foot and footwear during the heel strike phase of the gait cycle. An evaluation of the mechanical behaviour of the heel pad tissues at the heel strike during bare and shod conditions is developed.

Some notes about the structural conformation and the mechanical properties of the calcaneal fat pad tissue are reported in Annex A. The mechanical properties of the tissue are determined by the dimension of adipose chambers, the thickness of connective septae walls and the mechanical properties of the different soft tissues. Different numerical models are provided accounting for the variation of chambers dimensions, septae thickness and tissues characteristics, in order to evaluate the influence of each component on the overall mechanical response.
CHAPTER ONE

MORPHOMETRIC AND BIOMECHANICAL ANALYSIS
OF THE FOOT

1.1. INTRODUCTION

This chapter provides general information about the anatomic configuration of the foot and its movements. The foot is the terminal portion of the limb which bears weight and allows locomotion. The human foot and the ankle are strong and complex mechanical structures containing more than 26 bones, 33 joints (20 of which are actively articulated), and more than a hundred muscles, tendons, and ligaments. This complex structure allows balance during standing posture and acts as a lever during locomotion. The analysis of the gait cycle makes it possible to evaluate the movements of the foot and the reaction force generated during walking and running.

Attention will be focused on plantar soft tissues, and in particular on heel pad, or soft tissues surrounding the calcaneus. The mechanical response under different loading conditions is affected by the properties of tissues and their structural organization and interaction. The skin binds up the adipose tissues and represents an important term in the evaluation of the deformation induced. The mechanical role of adipose tissues is to withstand load transmission phenomena, such as damping shocks generated during walking or running and to guarantee a smooth distribution of plantar pressure. The mechanical behaviour of rear foot is investigated by experimental tests which define the overall response of the tissues under different loading conditions, such as weight bearing or walking. It has been shown that the structural behaviour of the heel pad varies widely across individuals. These differences may be caused by differences in the thicknesses of the heel pad and/or by material properties alterations caused by aging or degenerative effects. Ultimately, this chapter reports an analysis of the factors which influence the mechanical properties of the tissues.
1.2. NOTES ON THE ANATOMY OF THE FOOT

The human foot is a complex multi-articular mechanical structure consisting of bones, joints and soft tissues that plays an extremely important role in the biomechanical function of the lower extremity and is controlled by both intrinsic and extrinsic muscles.

As reported in Figure 1.1 the foot can be divided into three main regions: the forefoot (and phalanges), the midfoot and the hindfoot.

![Figure 1.1. Main region of the foot: the forefoot and phalanges (1), the midfoot (2) and the hindfoot (3)](image)

1.2.1. Bones of the hindfoot

The articulations of the human hindfoot play an integral role in gait and other movements of the foot. The hindfoot is actually composed of multiple joints that combined, allow for movements such as dorsiflexion, plantarflexion, inversion, and eversion. The talocrural joint (or ankle) is the articulation between the tibia, fibula and the talar dome, and is mainly responsible for plantar/dorsiflexion. Another joint of the hindfoot is the sub-talar joint, which is the articulation between the talus and the calcaneus that provides the inversion and eversion motions. In addition to the bony
support, the talocrural and the sub-talar joint also receive substantial support from ligaments and other soft tissue structures.

The calcaneus (Figure 1.2) is located under the talus and has six top faces. Behind the veneer postero-lateral, the upper surface of the bone becomes irregularly cylindrical. The underside is irregular, has a rear tuberosity of the calcaneus that presents two tubercles, medial and lateral. At its front end face it is bound by another tubercle. On the frontal side there is the trochlear process, which is known as the union of the anterior third with the middle third, above and below which there are two grooves for the passage of the peroneal tendons. The medial surface is characterized by the presence of a long shower crossed by tendons, blood vessels and nerves, which begin on the leg and finish on the foot. It is bounded in the back of the medial tubercle of the
calcaneus, in front of a strong capital known as the sustentaculum, because on it rests the medial talus. The base of the sustentaculum is cut by a groove for the passage of the flexor hallucislongus tendon. The anterior articular surface has a vertical concave and convex transversely, so as to comply with the surface articulations of the cuboid. The face is tilted back and corresponds to the projection of the heel. At the bottom it is rough and gives insertion to the calcaneal tendon (Achille’s), while at the top it is smooth and is separated by a grant from the synovial tendon.

The talus (or astragalus) is one of the most important bones of the foot. It rests on top of the calcaneus and is an irregular cuboid. Its dome-shaped upper surface articulates with the distal ends of the tibia and fibula. The talar head articulates with the navicular bone.

1.2.2. Bones of the midfoot

As reported in Figure 1.1 the midfoot is composed of five of the seven tarsal bones, the navicular, cuboid, and three cuneiform bones. These can be thought of as being arranged in two irregular rows, with the cuboid occupying space in both rows. The proximal row contains the navicular on the medial side of the foot and the cuboid on the lateral side. The distal row contains the three cuneiforms (medial, intermediate, and lateral) and the cuboid. The boundary between the midfoot and forefoot consists of five tarsometatarsal (TMT) joints, the joints between the distal row of the midfoot and the bases of the metatarsals. The medial, intermediate, and lateral cuneiforms articulate with the first, second, and third metatarsals, respectively; the cuboid articulates with the fourth and fifth metatarsals. There are also multiple joints within the midfoot itself. The distal row of the midfoot has two intercuneiform joints (between adjacent cuneiforms) and a cuneocuboid joint (between the lateral cuneiform and the cuboid). Proximally, the three cuneiforms articulate with the navicular bone (the cuneonavicular joints). In some individuals, there is also a small articulation between the cuboid and navicular.

In addition to its articular surfaces, each tarsal bone has specific features adapted for each function. For example, the medial surface of the navicular projects downward to form a tuberosity, which serves as an attachment for the tibialis posterior tendon. The lateral surface of the cuboid also has a tuberosity, which serves as a ligament
attachment. The cuboid bone has no major tendon attachments; however, the peroneus longus tendon passes across the cuboid tuberosity to run in a groove on the plantar surface of the bone. The peroneus longus tendon often contains a sesamoid bone, which articulates with a small facet on the tuberosity.

1.2.3. Bones of the forefoot and phalanges

The framework of the forefoot is formed by five metatarsal bones and 14 phalanges (the bones of the digits or toes). Each digit has three phalanges (proximal, middle, and distal), except for the big toe, which has only the proximal and distal one. The digits and their metatarsal rays are numbered from one to five, starting with the big toe. The metatarsals and phalanges are long bones. The proximal end or base of each bone has a smooth articular surface where it forms a joint with the adjacent bone. The distal end or head also has an articular surface, except for the distal phalanges, whose distal ends provide attachment for the soft tissue of the digit tips.

Of the metatarsal bones, the first bears the most weight and plays the most important role in propulsion and, therefore, it is the shortest. It provides attachment for several tendons, including the tibialis anterior and the peroneus longus. The fifth metatarsal has a protuberance on the lateral side of its base, to which the peroneus brevis tendon is attached. The second, third, and fourth are the most stable of the metatarsals, in part because of their protected position, but also because they have only minor tendon attachments, and therefore are not subjected to strong pulling forces.

Near the head of the first metatarsal, on the plantar surface of the foot, there are often two sesamoid bones. A sesamoid is a small, oval-shaped bone which develops inside a tendon, where the tendon passes over a bony prominence.

1.2.4. Ligaments

The function of foot ligaments is essentially to stabilize the joints and to maintain in the structural shape of the arch. The foot joints are complex and numerous. The ligaments of the hindfoot are very important because they stabilize and support the ankle and the sub-talar joints that have a fundamental role during the motion of the foot. These ligaments are responsible for ankle sprains that are the most common type of acute sport trauma (Fong et al. 2009).
In the ankle a superior articulation, talocrural or ankle joint, and an inferior articulation represented by a subtalar or talocalcaneal joint can be distinguished. The talocrural joint is composed of the tibia bone, the fibula bone and the talus bone underneath. The subtalar joint consists of talus bone on the top and calcaneus bone on the bottom.

![Figure 1.3. Ankle ligaments](image)

The major ligaments of the ankle are: the anterior tibiofibular ligament, which connects the tibia to the fibula; the lateral collateral ligaments, which attach the fibula to the calcaneus and gives the ankle lateral stability; and, on the medial side of the ankle, the deltoid ligaments, which connect the tibia to the talus and calcaneus and provide medial stability.

Other important articulations are the transverse tarsal joints, represented by talocalcaneonavicular and calcaneocuboid, and distal intertarsal joints, namely the cuneonavicular, cuboidonavicular and intercuneiform joints. The talocalcaneonavicular articulation is composed of the talus, which is received into the concavity formed by the posterior surface of the navicular, the anterior articular surface of the calcaneus, and the upper surface of the plantar calcaneonavicular ligament. The two ligaments in this joint are the articular capsule and the dorsal talonavicular. The ligaments connecting the calcaneus with the cuboid in the calcaneocuboid articulation are the articular capsule, the dorsal calcaneocuboid ligament, part of the bifurcated ligament, the long plantar ligament and the plantar calcaneocuboid ligament. In the cuneonavicular and cuboidonavicular articulations, the navicular and the cuboid are connected to the three cuneiform bones by dorsal and
Morphometric and biomechanical analysis of the foot

plantar ligaments. In the intercuneiform joints, the three cuneiform bones and the cuboid are connected together by dorsal, plantar, and interosseous ligaments.

The midfoot and forefoot articulations are tarsometatarsa (Lisfranc), intermetatarsal, metatarsophalangeal and interphalangeal joints. The tarso-metatarsal joints, which define the Lisfranc line, are arthrodial joints that connect the three cuneiform and the cuboid bones to the bases of the five metatarsal bones. The fibrous capsule is reinforced by dorsal, plantar and interosseous ligaments, which also connect the intermetatarsal articulation.

The joints between the metatarsals and the proximal phalanges are called the metatarsophalangeal (MTP) joints. Each digit also has two interphalangeal (IP) joints, proximal (PIP) and distal (DIP), except for the big toe, which has only one IP joint. Each MTP and IP joint is bound together by several ligaments, one on each side of the joint (medial and lateral collateral ligaments), and one along the plantar surface (plantar ligament). The metatarsophalangeal joints are important also because they are the attachment points of the plantar fascia (Figure 1.2). The plantar fascia consists in a thin fibrous band that spans between the medial process of the calcaneal tuberosity, diverging distally to form five bands to continuing up to metatarsophalangeal joints. It is considered to be one of the major stabilizing structures of the longitudinal arch of human foot (Wangdo et al. 1995).

The sesamoid bones articulate with the head of the first metatarsal, and function as part of the first MTP joint. They are held in place by their tendons, and are also supported by ligaments. These include the sesamoid collateral ligaments (which bind the sesamoids to the metatarsal head) and the intersesamoidal ligament (which connects the sesamoids to each other).

1.2.5. Muscles and tendons

The arches of the foot are maintained not only by the shapes of the bones as well as by ligaments. In addition, muscles and tendons play an important role in supporting the arches. The muscles can be classified as extrinsic muscles arising from the lower leg and intrinsic muscles which arise within the foot itself. These can in turn be divided into dorsal and plantar groups. All these muscles are actively providing stability during
locomotion and balance during standing and a strong lever arm effect during propulsion (Abbound, 2002).

The intrinsic muscles are located within the foot and cause movement of the toes. These muscles are flexors (plantar flexors), extensors (dorsiflexors), abductors, and adductors of the toes. Flexor hallucis brevis, whose tendon contains the sesamoid bones of the first metatarsal joint, is an intrinsic plantar flexor of the big toe.

The extrinsic muscles are located outside the foot, in the lower leg and have long tendons that cross the ankle, to attach on the bones of the foot and assist in movement (O’Connor and Hamill, 2004). Altogether there are thirteen tendons that cross the ankle. The gastrocnemius and soleus muscle are located in the posterior part of the lower extremity and are the main plantar flexor of the ankle. It allows a person to point the foot and stand on tip-toe. The two heads unite into a broad aponeurosis which eventually unites with deep tendon of the soleus to form the Achilles tendon. The peroneal muscles are on the outside of the legs; they provide stability and allow the foot to turn out. Other muscles are responsible for movements of the ankle, foot and toes and help to support the arches of the foot.

1.3. HEEL PAD TISSUES FUNCTIONALITY

The heel pad region, composed of calcaneal fat pad and skin, acts as an interface between the calcaneus and the ground. During the static standing and the stance phase of the gait cycle the heel pad tissues represent the weight-bearing and shock-absorbing structures of the foot. The heel pad tissues has the ability to dump impact forces at the heel strike during the gait cycle and running (Whittle, 1999), making it possible to achieve a smooth distribution of plantar pressure (Ledoux and Hillstrom, 2002). The mechanical response under different loading conditions is affected by the properties of the tissues and their structural organization and interaction. The skin binds up the calcaneal fat pad and represents an important term in the evaluation of the deformation induced (Figure 1.4).
In this way, their conformation and mechanical properties strongly influence body posture during static standing and the gait cycle. A specific knowledge of heel pad tissues mechanical behaviour contributes to an investigation of foot mechanical functionality, taking into account the interaction phenomena between the foot and surrounding structures, such as footwear products (Zheng et al., 2000; Rome et al., 2001). One of the most important factors which must be considered is the plantar pressure because this is associated with comfort and with the onset of foot pain and ulceration. In this way, the total thickness of heel pad tissue, i.e. calcaneal fat pad and skin, or the presence of diseases influences the plantar pressure, in particular under the heel and metatarsal heads (Rome, 1998; Cavanagh et al., 1999). Considering the heel pad, the highest thickness values are found where severe interactions develop, such as loading by impacts (Rome, 1998). The thickness ranges between 14.4 and 24.5 mm with an average value of 18 mm (Uzel et al., 2006). The thickness is measured by ultrasound and measurements from standardized foot radiographs, which represent objective and reliable techniques (Cavanagh, 1999; Prichasuk et al., 1994). The distribution of overall heel pad thickness varies according to local loading conditions.

1.3.1. Calcaneal fat pad

Within the heel pad region, particular attention is paid to the calcaneal fat pad, which is the portion of the plantar region interposed between the calcaneus and skin that plays a fundamental role in foot mechanics (Fig. 1.5(a)) (Natali et al., 2010). The calcaneal fat pad is mainly organized according to a honeycomb configuration (Jahss et al., 1992a-1992b; Snow and Bohne 2006). Fat tissue chambers are embedded and separated from each other by connective septa (Fig. 1.5(b-c)).
In different locations within the calcaneal fat pad structure, chambers and septa are characterized by specific dimensions and orientations (Hsu et al., 2007). These aspects are heavily influenced by the specific local loading conditions. The fibrous septa extending from the skin to the bones is characterized into small chambers connecting directly with the inside of the sub cutis and greater chambers situated deep in the small chamber stratum. Therefore, the calcaneal fat pad is anatomically divided into a superficial microchamber and a deep macrochamber layers (Hsu et al., 2007).

1.3.2. Skin

Skin is one of the most complex biological tissues. It consists of intertwined networks of collagen, elastic and nerve fibres, small blood vessels and lymphatics.
The arrangement organization of collagen fibres determines a strongly anisotropic behaviour (Reihsner and Menzel 1996). In general, the orientation of the collagen fibres, which affects the extensibility of the skin, depends on the location within the body and is described by cleavage lines (as reported previously by Langer 1978a, 1978b). On the plantar surface of the heel, different orientations of collagen fibres have been evaluated. Recent studies report that the cleavage lines have a predominant organization in concentric circles around the basis of the heel (Figure 1.6) (Andermahr et al. 2007; Natali et al., 2012a).

1.3.3. Heel pad properties

The structural configuration of a network of fibrous elements binding fat cells allows the heel pad to withstand impacts and prolonged pressure loads. Because of the high liquid content characterizing heel pad tissues and the complex arrangement of solid components influencing fluid movement, the heel pad is considered to have an almost incompressible behaviour, which is a typical feature of soft biological tissues (Natali et al., 2010a). The typical features of heel pad tissue mechanical response is characterized by large displacements and strains, almost incompressible behaviour, a non-linear stress-strain relationship, and time-dependent effects (Kinoshita et al., 1996; Hsu et al., 1998; Tong et al., 2003; Wearing et al., 2009). Initially the heel pad expands in the lateral and anterodorsal directions and the tissue has low stiffness in the normal direction. Subsequently, the collagen fibres of the calcaneal fat pad and skin come under tension, limiting the movement of the tissue and increasing the stiffness of the heel pad in the normal direction (Figure 1.7) (Natali et al., 2012a; 2012b).

Figure 1.7. Heel pad mechanical response under compressibility, from Hsu et al. (1998)
The mechanical properties of the heel pad are investigated using \textit{in vivo} or \textit{ex vivo} techniques. Different loading conditions are applied over the heel region and the corresponding displacement or strain is measured (Kinoshita et al., 1996; Hsu et al., 1998; Gefen et al., 2001a; Tong et al., 2003; Challis et al., 2008; Wearing et al., 2009), leading to a comprehensive characterization of the mechanical response of the plantar region. Some studies evaluate the mechanical response of the plantar tissue during walking. For example Gefen et al., (2001a) measured simultaneously the contact pressure and the heel pad deformation on a subject walking barefoot on a treadmill. In this way it is possible to obtain the overall heel pad stress-strain behaviour. However, this study does not report some important notes, such as impact velocity, the angle of impact or other boundary conditions. The specific stress-strain behaviour of the fat pad tissue is analyzed using \textit{in vitro} tests based on more simple geometry specimens, boundary conditions and loading protocols. Experimental data from \textit{in vitro} tests can be found in the literature (Ledoux et al., 2004). Miller-Young et al. (2002) developed compression tests on cylindrical specimens from the human fat pad under different loading conditions and evaluated stress relaxation phenomena that were also studied by Ledoux and Blevins (2007) on similar geometry specimens.

It is possible to evaluate the variation of the heel pad compressibility, considering the different loading conditions. In detail, the heel pad compressibility index (HPCI) is defined as the ratio of the heel pad thickness in loaded conditions (LHPT) to unloaded positions (ULHPT) (Figure 1.8).

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{heel_pad_compressibility.png}
\caption{Heel pad compressibility index, from Matteoli et al. (2009)}
\end{figure}

This index expresses the ability of the heel pad to be compressed. If HPCI is close to 1 then the heel pad presents a high stiffness. The table 1.1 reveals the experimental data obtained from different experimental tests in normal conditions.
Morphometric and biomechanical analysis of the foot

Biomechanical analysis of heel pad tissues

<table>
<thead>
<tr>
<th>Authors</th>
<th>Age</th>
<th>Tools</th>
<th>Loading rate</th>
<th>ULHPT (sd) mm</th>
<th>LHPT (sd) mm</th>
<th>HPCI (sd)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Uzel et al. (2006)</td>
<td>20-30</td>
<td>sonography</td>
<td>weight bearing</td>
<td>18.6 (2.5)</td>
<td>11.3 (1.9)</td>
<td>0.61 (0.06)</td>
</tr>
<tr>
<td>Tong et al. (2003)</td>
<td>26-75</td>
<td>test rig</td>
<td>quasi static</td>
<td>15.5 (2.4)</td>
<td>10.1 (1.8)</td>
<td>0.65 (0.077)</td>
</tr>
<tr>
<td>Wearing et al. (2009)</td>
<td>34-61</td>
<td>lateral radiographs</td>
<td>walking</td>
<td>19.1 (1.9)</td>
<td>8.8 (1.5)</td>
<td>-</td>
</tr>
<tr>
<td>Prichasuk et al. (1994)</td>
<td>20-60</td>
<td>lateral radiographs</td>
<td>weight bearing</td>
<td>18.70 (2.46)</td>
<td>9.97 (2.33)</td>
<td>0.53 (0.09)</td>
</tr>
</tbody>
</table>

Table 1.1. Survey of ULHPT, LHPT, HPCI for the normal heel pad

Uzel et al. (2006) reported a study about the heel pad thickness and the HPCI comparing sedentary subjects with different athletic subjects. The HPCI was calculated measuring the heel pad thickness in loading and unloading conditions via sonography. The measurements were computed during a standing position and a no stress-strain curve could be calculated with the tools available.

However, the experimental conditions, such as tools or strain rate loading conditions, lead to differences in measured heel pad properties. In addition it has been shown that the structural behaviour of the heel pad varies widely across individuals. These differences may be due to a number of factors, including differences in the thickness of the heel pad, the geometry of the calcaneus and the material properties of the relevant tissues due to individual variability, as well as age, injury and disease (Alcantara et al., 2002; Ozdermir et al., 2004; Natali et al., 2008, 2012; Matteoli et al., 2009). For example, the unloaded heel pad thickness increases with age and with body weight. In general, people have the tendency to increase their weight with age because of increased body fat. Increased body fat content in the heel pad leads to increased pressure in a closed space with an inevitable results of the reduction of elasticity (Alcantara et al., 2002). This factor is highlighted by an increase in the compressibility index from 0.52 (±0.08) for normal weight to 0.58 (±0.11) for overweight subjects (Pirchasuk et al., 1994). The increase in thickness may also hinder the afferent sensory pathways and thus reduce the sensory perception. This means that, as in the case of diabetic neuropathy, the subject continues to load the weight on the foot even in the presence of lacerations, thereby aggravating the situation.
Some notes about the correlation between the mechanical response of calcaneal fat pad tissues and the meso-structural conformation, with regard to adipose chambers and connective septae, are reported in Appendix A. Numerical analyses must be adopted to evaluate the influence of the meso-structural configuration on the calcaneal fat pad mechanical response. This approach represents the basis for considering the influence of degenerative/pathological factors on the mechanical response of the heel pad region (Natali et al., 2012).

**Ageing**

The shock absorbing characteristics of the heel pad *in vivo* were examined by Kinoshita et al. (1996) in different groups of elder and young adults. A free-fall impact testing device which consisted of an instrument shaft, accelerometer and position detection transducer was used to obtain deceleration and deformation of the heel pad during impact. In order to simulate walking and running, 0.57 m/s and 0.94 m/s impact velocities were chosen, respectively (Figure 1.9). The force-displacement curves showed that the energy absorbed was less for the elderly than for the younger adults, suggesting that the capacity for shock absorbency of the heel pad declines with age.

![Figure 1.9. Force-displacement curves for young (a) and old (b), from Kinoshita et al. (1996)](image)

Hsu et al. (1998) compared the mechanical properties of the heel pad in young and aged adults. Using a compression relaxation device with a push-pull system, the thickness of heel pad was determined under different loading conditions. The heel pad was compressed by serial increments of 0.5 Kg to a maximum of 3 Kg and then relaxed sequentially. The ultrasound traducer was moved manually step-by-step at a speed of approximately 0.06 cm/s. The average unloaded heel pad thickness was 1.76 (±0.20) cm in the young group and 2.01 (±0.24) cm in the elderly group. The average
compressibility index was 0.53 (±0.077) in the young group and 0.613 (±0.055) in the elderly group. These results, according to Kinoshita et al. (1996) indicated the loss of the elasticity of the heel pad in aged adults.

The effect of aging on the mechanical properties of plantar soft tissues was evaluated also by Kwan et al. (2010). In this study, the thickness and the stiffness of the plantar soft tissues were examined using tissue ultrasound palpation system. The measurements were made under the big toe, the first, the third and the fifth metatarsal heads and the heel. The results obtained among different age groups pointed out that the mean stiffness and thickness increased with age. The stiffened soft tissue may reduce the adaptability of the tissue to respond to sudden or repetitive stress, which may lead to foot problems in elderly people (Hsu et al., 2005). Degenerative effects cause a gradual loss of collagen, a decrease in the elastic fibrous tissue and a reduction of water content, as well as distortion and rupture of the fibrous-tissue strands with spilling of the fat cells and local loss of fat pad or rupture of the fibrous tissue septa (Prichasuk et al., 1994). These changes contribute to the instability of the foot in stance phase of walking.

**Heel pain**

Pirchasuk et al. (1994) and Pirchasuk (1994) studied the heel pad thickness and compressibility in normal subjects, using loaded and unloaded lateral radiographs. The compressibility index increased with age, but that there was no significant difference with gender. In the second study, the same methodology is adopted to compare the HPCI between normal heel and heel pain. The results showed an increase in the compressibility index for patients with heel pain. The same considerations were obtained by the study of Kanatli et al. (2001) which aimed to determine the effect of heel pad thickness and its compressibility in normal subjects and in patients with heel pain. Both heels of the control subjects and painful heels of the patients were radiographed with and without weight bearing.

The study of Tong et al. (2003) reports a preliminary development of a test rig and the acquired mechanical behaviour of the human heel pad. The plantar thickness and the HPCI of normal, plantar heel pain and diabetic subjects were measured. The results revealed that patients with plantar heel pain and diabetic patients were less compressible compared with those of healthy individuals.
The dynamic lateral foot radiographs was adopted by Wearing et al (2009) to measure the bulk compressive properties of the heel pad in individuals with and without plantar pain while walking. Plantar pressure data were simultaneously acquired during the experimental tests. In this way, the energy dissipation and the peak strain were estimated from the subsequent stress-strain curves. The analyses revealed an increase in the compressibility index in the patients with plantar heel pain, as reported by other authors (Pirchasuk (1994); Kanatli et al. (2001); Tong et al. (2003)).

A specific analysis relevant to the mechanical properties of the heel pad of patients with heel pain was described by Özdemir et al. (2004). They showed a relationship between the thickness and elasticity of the heel pad with age, sex, obesity and more. An increase in heel fat pad thickness with ageing and an increase in body weight, caused the HPCI to rise, as measured with a lateral radiographs. In addition, the establishment of subcalcaneal spurs increased the stiffness of the heel pad and played a role in the formation of heel pad.

Table 1.2 compares the results of different experimental tests computed on normal subjects and on patients with heel pain. In general it is possible to observe an increment of the heel pad thickness and of the stiffness of heel tissues.

<table>
<thead>
<tr>
<th>Authors</th>
<th>tools</th>
<th>loading rate</th>
<th>ULHPT (±sd) mm</th>
<th>LHPT (±sd) mm</th>
<th>HPCI (±sd)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Wearing et al. (2009)</td>
<td>lateral radiographs</td>
<td>walking</td>
<td>19.1 (±1.9)</td>
<td>8.8 (±1.5)</td>
<td>normal</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>19.3 (±1.7)</td>
<td>10.0 (±2.1)</td>
<td>heel pain</td>
</tr>
<tr>
<td>Kanatli et al. (2001)</td>
<td>lateral radiographs</td>
<td>weight bearing</td>
<td>19.55 (±2.52)</td>
<td>11.81 (±2.84)</td>
<td>normal</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>20.45 (±2.89)</td>
<td>14.02 (±3.38)</td>
<td>heel pain</td>
</tr>
<tr>
<td>Tong et al. (2003)</td>
<td>test rig</td>
<td>quasi static</td>
<td>15.5 (±2.4)</td>
<td>10.1 (±1.8)</td>
<td>normal</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>16.8 (±2.4)</td>
<td>12.2 (±2.5)</td>
<td>heel pain</td>
</tr>
<tr>
<td>Prichasuk (1994)</td>
<td>lateral radiographs</td>
<td>weight bearing</td>
<td>18.77 (±2.33)</td>
<td>9.85 (±2.22)</td>
<td>normal</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>20.66 (±2.59)</td>
<td>12.13 (±2.64)</td>
<td>heel pain</td>
</tr>
</tbody>
</table>

Table 1.2. ULHPT, LHPT e HPCI in patients with plantar heel pain
These results are associated with changes in the histology and morphometry of the system caused by pathology or trauma (Tsai et al., 1999). Tsai et al. (1999) evaluated the mechanical properties of the heel pad in unilateral plantar heel pain syndrome. The load-displacement curve during a loading-unloading cycle was considered and the compressibility index and the energy dissipation ratio were calculated. The heel pad thickness in the painful heel was 1.82 (±0.24) cm and 1.76 (±0.29) cm in the painless heel. After the loading weight of 1 Kg, the displacement measured on the painful heel pad was 0.59 (±0.12) cm while it was 0.70 (±0.15) cm on painless heel pad. The results showed that the heel pad in plantar heel pain syndrome was stiffer under light pressure than the heel pad on the painless side.

Pathology

Some studies report the mechanical properties of the heel pad in patients with pathology. Atrophy of the fat pads is observed with diseases such as diabetes, peripheral vascular disease, senescence and collagen disorders. Hsu et al. (2000) studied the mechanical properties in the heel pad of patients with type II diabetes, but found no difference in unloaded heel pad thickness, compressibility index or elastic modulus. A similar study was subsequently carried out by Hsu et al. (2009). Using a loading device consisting of an ultrasound transducer and a load cell, they measured the unloaded thickness, strain and elastic modulus in microchambers, macrochambers and heel pads. Microchamber strain in diabetic patients (0.291 (±0.14)) was significantly greater than that in healthy subjects (0.104 (±0.057)). Macrochamber strain in diabetic patients was significantly less (0.355 (±0.098)) than that in healthy subjects (0.450 (±0.092)). This alteration, determined by the fact that collagen fibrils in diabetic heel samples were ruptured with uneven distribution (Hsu et al., 2002) can cause diminished cushioning capacities in diabetic patients. This consideration is reported also by Tong et al (2003). This study revealed that the heel pad properties of diabetic patients were less compressible compared with those of healthy individuals.

The compressive mechanical properties of diabetic plantar fat pad tissue were also determined by Pai and Ledoux (2010) which consider cylindrical specimens from six relevant locations beneath the foot. Specimen were subjected to biomechanically realist strain of 50% in compression using triangle wave tests at five frequencies to determine tissue modulus, energy loss and strain rate dependence. This study
demonstrated altered mechanical properties with a significantly increase modulus (1146 kPa vs 593 kPa), but no change in energy loss (68.5% vs 67.9%) for diabetics plantar soft tissue.

Some studies have already shown that the collagen septa in diabetic heel fat pads are thicker and the adipose cells smaller than in normal tissues. In the case of diabetes, there is an increase in the stiffness of the plantar tissue and, having compromised the ability to dissipate the stress, the risk of developing ulcers has been established (Kao et al., 1999).

1.4. THE ROLE OF THE FOOT

The foot has to support body weight during standing and to act as a lever during locomotion. These functions require a high degree of stability. The foot can adapt to standing and walking on uneven surfaces, because of its flexibility. It is able to form a rigid platform which does not collapse under body weight and to make a good contact with almost any supporting surface. This is made possible by bones and joints, which form the structure of the foot, and by ligaments, muscles and tendons, that provide support and balance the body.

1.4.1. The load of the foot during standing

The foot plays an extremely important role in the biomechanical function of the lower extremity, providing support and balance during standing and stabilizing the body during gait. For this purpose notes about the load acting on the foot during standing and walking are reported.

In quiet standing the ground reaction force is vertical and constant, being equal and opposite to the body weight. The load on the foot is transmitted by the tibiotalar articulation, but many authors have demonstrated that the fibula carries a part of the body weight (Wang et al. 1996; Takebe 1984; Lambert 1971).
The body weight transmitted to the foot through the ankle joint is distributed over three very specific points of the foot (Figure 1.10). In detail, if 6 Kg are applied on the foot, a sixth of the force acts on the fifth metatarsal head, a third on the first metatarsal head and half of the weight force acts on the calcaneus (Kapandji I.A., 1970).

### 1.4.2. The movements of the foot and the ankle

To understand the load on the foot during walking it is necessary to describe the movement that the foot can undergo during the gait cycle.

The movement of the body is defined by reference to a plane (Figure 1.11). The sagittal plane is a vertical plane which passes from front to rear dividing the body into right and left sections, and the terms medial and lateral is related to this plane. The frontal or coronal plane passes from side to side at right angles to the sagittal plane which divides the body into an anterior and posterior section. The transverse plane is parallel to the flat surface of the ground. Planes in this direction divide the body into an upper and lower part. The same configuration is used to describe the movement of the foot.

Looking closely, a discussion of mechanics of the motion of the ankle joint complex requires consistent terminology. The major motions about an anatomical joint coordinate system are rotations: plantarflexion/dorsiflexion, inversion/eversion, and internal/external rotation (or abduction/adduction).
To define the movements of the ankle complex joint it is useful to define the coordinate system of the joint. The most widely adopted in the literature is represented in Figure 1.12 (Siegler et al. 2005). The points A1, A2, A3 illustrated in the figure are the lateral malleolus, medial malleolus, and the centroid of the tibial cross-section respectively. These three points define an anatomical frame for the tibia as follows: The line A2A1 is the Z-axis of the tibia. The perpendicular to the plane containing A1, A2, A3 is the X-axis of the tibia, and the common perpendicular is the Y-axis. The origin is located at the mid-point between A1 and A2.

Figure 1.12. (a) Definition of the axis of the joint ankle complex (b) the movements of the foot
Flexion and extension are movements in the sagittal plane. Flexion movements bend the body part away from the anatomical position. Extension is the movement in the opposite direction back to the anatomical position and beyond into a reversed position. In the case of the foot the flexion and extension movements are called plantarflexion and dorsiflexion respectively and represent the rotations around the X-axis. Plantarflexion and dorsiflexion are the major components of the motion at the ankle joint during gait. The range of motion of the ankle joint varies across the literature but generally is reported to be around 70 degrees of full motion, although this number is dependent on loading conditions and measurement technique. The typical breakdown of total talocrural motion is usually around 30-40 degrees of plantarflexion, and about 20-30 degrees of dorsiflexion. Many authors cite variable ranges of motion for the ankle joint in the plantarflexion / dorsiflexion rotations (Table 1.3).

<table>
<thead>
<tr>
<th>Authors</th>
<th>Range of motion (deg)</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Plantarflexion</td>
<td>Dorsiflexion</td>
</tr>
<tr>
<td>Kapandji, 1970</td>
<td>30 -50</td>
<td>20 -30</td>
</tr>
<tr>
<td>Nordin and Frankel, 1989</td>
<td>25-35</td>
<td>10 - 20</td>
</tr>
<tr>
<td>Siegler et al., 1988</td>
<td>37.6 - 45.8</td>
<td>20.3 – 29.8</td>
</tr>
<tr>
<td>Allinger and Engberg, 1993</td>
<td>20 - 50</td>
<td>13 - 33</td>
</tr>
</tbody>
</table>

Table 1.3. Range of motion with regard to plantarflexion and dorsiflexion movements

The inversion and eversion movement of the foot are movements that raise the medial and the lateral border of the foot respectively. This kind of motion is developed with rotations along the long axis of the foot, Z-axis. Motion in this direction is thought to be primarily contributed to by the subtalar joint. The movement of inversion is produced by any muscle that is attached to the medial side of the foot. The tibialis anterior and tibialis posterior are responsible, assisted by the extensor and flexor hallucislongus on both occasions. The tibialis anterior dorsiflexes and the tibialis posterior plantarflexes the foot at the ankle joint and these opposite effects cancel each other out when the two muscles combine to produce an inversion of the foot. The movement of eversion is produced by muscles that are attached to the lateral side of the foot. The peroneus longus, brevis and tertius are responsible for this. The former two, whose tendons pass behind the lateral malleolus, are plantarflexors, the latter is a
dorsiflexor of the ankle joint. These opposite effects cancel each other out when the three muscles combine to produce a simple eversion of the foot.

In the literature there are many works that reported the range of motion of the inversion and eversion movement (Table 1.4). The abduction and adduction are movements in the frontal plane. Adduction movement carries a body part away from the midline. Abduction is movement in the opposite direction towards the midline. Rotations about the long axis of the tibia, the Y-axis, are internal and external rotation or adduction and abduction respectively. These types of rotations usually do not occur by themselves but in combination with plantarflexion, dorsiflexion, inversion, and eversion. The range of motion of the abduction and adduction movement of the foot are reported in the following table (Table 1.5).

<table>
<thead>
<tr>
<th>Authors</th>
<th>Range of motion (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Inversion</td>
</tr>
<tr>
<td>Kapandji, 1970</td>
<td>20</td>
</tr>
<tr>
<td>Siegler et al., 1988</td>
<td>14.5 - 22</td>
</tr>
<tr>
<td>Siegler et al., 2005</td>
<td>12.5 ± 5.8, in vivo</td>
</tr>
<tr>
<td>Cass et al., 1984</td>
<td>15 - 20</td>
</tr>
<tr>
<td>Sarrafian, 1993</td>
<td>30</td>
</tr>
</tbody>
</table>

Table 1.4. Range of motion with regard to inversion and eversion movements

Supination and pronation are a combination of the above motions. It is common to use both supination and inversion, and pronation and eversion interchangeably. But, supination is actually a combination of inversion, plantarflexion and adduction. Pronation is a combination of eversion, dorsiflexion and abduction.
Supination is a triplanar motion involving the foot moving down and towards the center of the body, while pronation is a triplanar motion of the subtalar joint involving the foot moving up and away from the center of the body.

1.4.3. The gait cycle

The gait cycle is defined as the time between the first contact with the ground by the heel of one foot and the next heel-to-ground contact with the same foot. One single limb cycle is usually composed of two phases: stance and swing. The stance phase begins when the foot first contacts the ground while the swing phase begins as the foot leaves the ground. On average, the gait cycle is about one second in duration with 60 percent in stance and 40 percent in swing (Figure 1.13). The stance and the swing phases of the gait can be divided into different sub-phases. With regard to the stance phase, there are five sub-phases: initial contact (heel-contact), loading response (or foot-flat), mid stance, terminal stance and pre-swing (or toe-off).

The first phase is the heel contact phase, which is when heel hits the ground. This phase continues until the foot is flat on the ground. This constitutes 20 percent of the total gait cycle. The next phase is the midstance phase. In this part of the gait cycle, which represents 30 percent of the cycle, your body weight passes over your foot as the body comes forward. This is where your foot supports your body weight.

![Figure 1.13. The two phases of the gait cycle: stance and swing](image-url)
This is the part of the gait cycle where an abnormally functioning foot, such as an over pronated foot (flat foot) or an over supinated foot (high arch foot), will manifest its problems. This phase ends as your body weight passes forward, eventually forcing your heel to rise during the terminal stance. During the last phase the foot pushes off the ground to propel the body forward and to prepare the body for the swing phase of the gait (Figure 1.14). Most forefoot pain occurs in this phase of the cycle.

As shown in Figure 1.15 the swing phase of the gait can be divided into three sub-phases: initial swing, mid-swing and terminal swing. The initial swing begins at toe off and continues until maximum knee flexion (60 degrees) occurs while the mid swing phase is the period from maximum knee flexion until the tibia is vertical or perpendicular to the ground. The swing phase ends with the terminal swing that begins when the tibia is vertical and ends at the point of initial contact of the other foot. Different measuring devices are available to evaluate the components of the ground reaction force during walking. The force platforms are a commonly used system for ground reaction force measurement, to investigate the three components of the reaction.
force in different conditions, for example, with regard to stroke patients (Chou et al. 2003), to women with a total arthroplasty (Perron et al., 2000), to diabetic neuropathic patients or to evaluate the influence of the walking speed during the gait cycle (Liikavainio et al., 2007). The vertical ground reaction force is represented by the characteristic curve reported in Figure 1.16, with two peaks and a local minimum. The value of the peaks is about 110% of BW (body weight), while the force in the minimum is about 80% of BW. The first peak occurs in the midstance when the center of mass moves rapidly towards the ground and this gives acceleration to the body mass. The minimum is due to the fact that the center of the mass rises as the body rotates around in support of the foot, and is accentuated by the oscillation of the contralateral limb, which tends to decrease the load on the sensor. The second peak occurs in the terminal phase, when the center of mass tends again to fall.

Figure 1.16. Experimental results: the vertical component of the ground reaction force
1.5. REFERENCES


Biomechanical analysis of heel pad tissues

CMBM – Centre of Mechanics of Biological Materials – University of Padova, Italy
Chapter 1


Matteoli S., Wilhjelm J.E., Torp-Pedersen S.T., 2009. Some of the factors influencing the heel pad compressibility index (HPCI).


CHAPTER TWO

HISTOLOGICAL CONFIGURATION OF SOFT TISSUES WITH RELATION TO THE BIOMECHANICAL BEHAVIOUR OF THE HEEL PAD TISSUES

2.1. INTRODUCTION

Detailed information about the structural and micro-structural configuration of the heel pad tissues, together with data from mechanical tests, is necessary to define the proper constitutive formulation.

Indeed, the microstructural configuration determines the isotropic or anisotropic behaviour, the linearity or non-linearity of the mechanical response and the dissipative phenomena that may develop when loads are applied. Soft tissues are characterised by strongly hierarchical configurations and the overall mechanical behaviour is determined by the mechanical properties of subcomponents and the biological and mechanical interactions developing between them. Attention will be focused on soft plantar tissues, including the plantar adipose tissue and the skin. Soft connective tissue is composed of few cells embedded within an abundant extracellular matrix (ECM). The ECM is a fibre reinforced composite material, composed of collagen fibres in a ground substance matrix. With regards to collagen, details describing the configuration of collagen reinforcing elements and the biological interactions and mechanical properties of its subcomponents will be discussed.

Experimental procedures on biological tissues can be classified as in vivo, in situ or in vitro. The mechanical properties of biological tissues are usually evaluated by in vitro techniques. Moreover, specific in situ and in vivo methodologies are designed to evaluate the overall mechanical response in the rearfoot. In detail, indentation tests on cadaveric foot with and without the skin are considered to evaluate the influence of the different tissues on the overall mechanical response. The mechanical behaviour of the living tissues are considered through the development of in vivo indentation tests.
2.2. SOFT CONNECTIVE TISSUES

Soft tissues, including calcaneal fat pad and skin, are composed of cells and ECM, and cell type is dependent on the particular tissue. Its main functions include: the development of the specific function of the tissue, production, regeneration and remodelling of the ECM and defence against external organisms. The composition, structure and functions of the ECM change greatly in different tissues. Soft tissue can be initially classified based on the relative abundance of cells and ECM. The connective tissue is characterised by few cells embedded within an abundant ECM, responsible for the particular mechanical response of this tissue. Examples of soft connective tissues are tendons, ligaments and cartilage. Cellular tissue is mainly composed of cells and the minimal ECM serves to separate cells or groups of cells. Muscular, nervous and epithelial tissues are typical examples. Soft connective tissue performs many functions, including load-bearing (tendons and ligaments), and the protection of organs from injury, as well as providing protective capsules around organs (a fibrous sheath around bones, the dermis, etc.), separation of cells or group of cells within organs and cellular tissues, and immune defence against infection (epimysium, perimysium and endomysium within muscles, epinerium, perinerium and endonerium within nerves, etc.), or insulation and a reservoir of energy (adipose tissues).

![Figure 2.1. Soft connective tissue components: cells and ECM which consists of ground substance and protein fibers](image)

*Figure 2.1. Soft connective tissue components: cells and ECM which consists of ground substance and protein fibers*
Generally, the cellular material occupies about 20% of the total tissue volume, while the ECM accounts for the remaining 80% (Frank and Shrive, 1999) (Figure 2.1). The ECM consists of two main components: protein fibres and the amorphous matrix. Protein fibres include elastic fibres, which are extremely flexible and behave much in the same way as rubber, and collagen fibres, which are stiff and form the main tensile load bearing components in the tissue. The amorphous matrix or ground substance, in which fibres are embedded, is a viscous gel composed of water, proteoglycans and other glycoproteins (Minns et al., 1973) (Figure 2.2).

![Figure 2.2. Photomicrograph of the principal fibers that limit movement and stabilize the body’s shape. The blue diagonal line is a single tropocollagen (pre-collagen unit). They combine into the triple helix (three stranded rope) of a collagen molecule (lower middle). The yellow snaky one is elastin. The thin cobweb-like fibers are reticulin - immature collagen found mostly in embryos](image)

As a consequence, soft connective tissue must, therefore, be considered and analysed as a fibre-reinforced composite material (Limbert and Taylor, 2002; Minns et al., 1973).

### 2.2.1. Configuration and mechanical behaviour of soft connective tissues

In the context of a fibre-reinforced composite material, fibrous proteins, with particular regard to collagen, represent the reinforcing fibres, while the ground substance is the isotropic matrix (Minns et al., 1973). Consequently, the mechanical behaviour is determined by the properties and abundance of the components and their interactions. Furthermore, the specific orientation of the collagen fibres determines the characteristic anisotropic response. Because collagen elements are characterised by
significantly higher stiffness than ground substance, they are largely responsible for the tensile behaviour of soft connective tissue. Consequently, the micro-structural phenomena which develop when tensile load increases are mainly those associated with collagen elements, with the tensile stress-strain response also qualitatively similar to that of collagen fibres. In contrast, because of the high length-thickness ratio characteristic of collagen elements when compressive loads are applied, micro-buckling of the fibres occurs and the compression behaviour of the tissue is mainly determined by the ground substance and its interaction with the fibrous network. When compressive loads are applied to the tissue, two main phenomena occur: the ground substance undergoes fluid fluxes within the solid skeleton; and the space between proteoglycans is reduced. Tissue compressive stiffness is consequently determined by the incompressible behaviour of the ground substance, its capacity to move within the fibrous network and the electrostatic interaction between glycosaminoglycans. The ground substance is a non-Newtonian viscous gel and its movement capacity within the solid skeleton strongly depends on the applied strain rate. As reported in literature (Nishihira et al., 2003) the influence of strain rate on the mechanical response is more evident in compression conditions because the mechanical behaviour is determined by the ground substance. Very high strain rate conditions prevent ground substance fluxes, resulting in high compression stiffness due to the incompressible nature of the gel. By reducing the strain rate, fluid flux occurs with less resistance and compression stiffness decreases. The microstructural phenomena which develop within soft tissue during loading require time to reach a condition of thermodynamic equilibrium, both for compressive and tensile situations. With regard to the latter case, time-dependence is less evident because micro-structural rearrangements within collagen fibres develop more quickly. As a consequence, the analysis of the mechanical behaviour of the tissue requires consideration of viscosity (Natali et al., 2004a).

2.2.2. Collagen components: configuration and mechanical behaviour

Collagen is the most abundant protein in the human organism. It is primarily responsible for the tensile behaviour of soft connective tissues, and in particular, tendons and ligaments. Collagen is characterised by a strongly hierarchical
organization: small tropocollagen molecules link together to form fibrils, which in turn give rise to fibres and fibre bundles. The primary structure of the tropocollagen molecule consists of an uninterrupted sequence of about 300 Glycine-X-Y triplets, where X and Y are frequently represented by proline. During post-translational modification, hydroxylation, oxidation and glycosylation processes of amino acid residues usually occur (Ottani et al., 2002). The triplets organize themselves into a right-handed $\alpha$-helix secondary structure and the conformational arrangement is defined by a left-handed helix tertiary structure. The quaternary super-structure of the tropocollagen molecule is obtained by linking three helixes by hydrogen bonds (Figure 2.3).

![Figure 2.3. Polypeptide chain (a). Possible sites of cleavage by chemomechanical caries removal reagents by degradation of glycine or hydroxyproline are indicated by red arrows. Triple helix (b). Sites of cleavage by degradation of intra-molecular cross links are shown by red arrows (b). With regard to collagen types I, II, III, V and XI, tropocollagen molecules link together to form fibrils. Tropocollagen units assembled to form a collagen fibril. Sites of cleavage by degradation of intermolecular cross links are indicated by red arrows (c)](image)
The molecule is approximately 300 nm in length and 1.5 nm in diameter (Ottani et al., 2002). Tropocollagen molecules are highly reactive and undergo spontaneous fibrillogenesis. For collagen types I, II, III, V and XI, tropocollagen molecules link together by inter-molecular covalent bonds to form fibrils (Figure 2.4), that are characterised by diameter and length, ranging between 20 to over 280 nm (Silver et al. 2003) and 5 µm to over 1 mm respectively (Kadler et al., 1996). For collagen types I, II, III, V and XI, tropocollagen molecules link together by inter-molecular covalent bonds to form fibrils (Figure 2.3), that are characterised by diameter and length ranging between 20 to over 280 nm (Silver et al., 2003) and 5 µm to over 1 mm respectively (Kadler et al., 1996).

Visualization of collagen fibrils by Transmission Electron Microscopy (TEM) or Atomic Force Microscopy (AFM) shows that tropocollagen molecules organize themselves in a cross striated structure (Figure 2.4) with a characteristic 67 nm repeat (Silver et al., 2003), known as the D-period.

Adjacent repeat elements are separated by a “gap” zone (Ottani et al., 2002). Each repeat element is composed of parallel tropocollagen molecules linked together by covalent bonds. Repeat elements are joined by further covalent bonds passing through the “gap” zone (Figure 2.4). This organization is known as the “D-period structure” or the “Hodge-Petruska” model (Petruska and Hodge, 1963).
Proteoglycans, most often decorin (Redaelli et al., 2003; Raspanti et al., 1997) and FACIT (fibril associated collagen) filaments (Eyre et al., 2004), serve to form inter-fibrillar bonds. Inter-fibrillar bonds tie adjoining fibrils together to form fibres (Figure 2.5) and seem to have a definite role in guaranteeing the mechanical coupling of fibrils.

Figure 2.5. Proteoglycans and collagen filaments enable bonds between collagen fibrils. (a) SEM (the horizontal field of view spans 20 µm), from Price et al., 2009, and (b) schematic representation

As mentioned, collagen reinforcing elements can assume different arrangements, such as fibrils, fibres or fibre bundles (Figure 2.6).

Figure 2.6. The organization of collagen in fibres bundles

Within soft tissue, collagen fibrils show a typical wavy configuration in the unstrained state, referred to as crimped, that is characterised by a helical nature with a periodicity
of between 10 and 100 µm (Figure 2.7), depending on the particular tissue type (Freed and Doehring, 2005).

![Image](image.png)

*Figure 2.7. Typical wavy configuration of collagen fibrils in the unstrained state. The photographs are acquired by Scanning Electron Microscopy (SEM)*

Because of its hierarchical organization, collagen reinforcing elements usually show a complex mechanical response that can be evaluated by analysing the mechanical behaviour of the components and the interactions which occur when tensile loads are applied. Experimental studies performed by Sasaki and Odajima (1996) on specimens from bovine Achilles Tendon have made it possible to evaluate the stress-strain behaviour of tropocollagen molecules. Linear elastic behaviour was assessed using a 3 GPa elastic modulus. The elastic modulus of the tropocollagen molecule has been investigated by other researchers (Hofmann et al., 1983; Nestler et al., 1983; Cusack and Miller, 1979), in samples from different animal species and anatomical sites, demonstrating similar results (ranging between 3 and 5 GPa).

The stress-strain behaviour of an uncrimped collagen fibril was investigated by Sasaki and Odajima (1996). A linear relationship in accordance with a 400 MPa elastic modulus was obtained, (Freed and Doehring, 2005; Redaelli et al., 2003).

The difference between the stiffness of tropocollagen molecules and collagen fibrils is determined by the larger number of deformation mechanisms acting within fibrils (Gautieriet al., 2011). In contrast to tropocollagen molecules and collagen fibrils, the tensile behaviour of collagen fibers and collagenous tissue is non-linear. At this purpose a schematic representation of the mechanical response of a collagen fiber is reported in Figure 2.8.
The situation can be explained by considering the crimped configuration of collagen fibrils and the orientation of inter-fibrillar bonds in the unstrained state. The stress-strain curve can be subdivided into three main regions. In the first region (the toe-region), collagen fibres extend, carrying load as an uncoiling spring. The material undergoes a progressive increase in stiffness due to the uncrimping of fibrils and the re-alignment of inter-fibrillar bonds in the direction of loading. In region II (the quasi-linear region), collagen fibrils are completely uncrimped and inter-fibrillar bonds are predominantly aligned along the direction of loading (Redaelli et al. 2003; Reese et al., 2010). The material stiffness value is almost constant and reaches its peak. When strain exceeds a specific limit, damage phenomena develop (region III or damage region). Collagen fibrils and inter-fibrillar bonds progressively breakdown and the material stiffness value decreases until the tissue fails (Natali et al. 2004b, 2005). Because of the many deformation mechanisms taking place in collagen fibres, their stiffness is not constant and the value is lower than that of collagen fibrils. In the toe-region and the quasi-linear region, the elastic modulus of collagenous tissue falls in the range of 10 and 100 MPa respectively (Freed and Doehring 2005).

### 2.2.3. Elastic components: configuration and mechanical behaviour

To limit the deformation and to prevent tearing of the tissues elastic fibers are intermingled with the collagen fibers. Elastic fibers (Figure 2.9) are fibrous proteins found in large amounts in tissues, such as the elastic walls of the aorta, skin and ligaments.
Figure 2.9. SEM image of elastic fibers with diameters in the range of 3 to 5 µm (a). Fluorescence microscope image of immunostained fibrillin-microfibrils in elastic fibers (b). From Yang, 2008

The elastic components are thinner than the collagen components. They are branched and unite with one another, forming an irregular network. Elastic fibers consist of an amorphous central region containing elastin surrounded by a sheath of 14 nm tubular microfibrils (Yang, 2008). The amino acid composition of elastin resembles that of collagen in so far as elastin is rich in glycine (Gly) and proline (Pro). Differences include greater quantities of valine (Val) and alanine (Ala), along with small amounts of hydroxyproline and no hydroxylysine.

Figure 2.10. Elastin molecules are joined together by covalent bonds to generate an extensive cross-linked network. Because each elastin molecule in the network can expand and contracts as a random coil, the entire network can stretch and recoil like a rubber.
Histological configuration of soft tissues with relation to the biomechanical behavior of the heel pad tissues

Through this structure (Figure 2.10) and its numerous links, elastic fibers are capable of stretching to one and one-half times their length, yielding easily to very small traction forces, but returning to their original shape when these forces are relaxed. Fung et al. (1993) reported the elastin tensile stress-strain response. Loading and unloading form two different curves, showing the existence of an energy dissipation mechanism in the material, even if the difference is small.

2.2.4. Ground substance: configuration and mechanical behaviour

The amorphous intercellular ground substance is colorless, transparent and homogeneous. It fills the space between cells and fibers of the connective tissue. The ground substance is a viscous gel mainly composed of an electrolytic water solution and highly negatively charged proteoglycans (PGs). The water solution behaves as a pore fluid within the solid skeleton of the extracellular matrix, and dissolved ionic species are mainly sodium Na\(^+\) cations and chloride Cl\(^-\) anions. Proteoglycans are long molecular structures developing along an axis made of hyaluronic acid (Figure 2.11).

Figure 2.11. SEM micrograph and the schematic representation of the structure of proteoglycans

On that axis other proteins (aggregans) are attached laterally and structured around their own axis. Along this axis, threads of amino-acids, called glycosaminoglycans (GAGs) are attached. The basic structure of GAGs is made by disaccharide units containing an uronic acid and an amino-glycan. The uronic acid displays a negatively
charged carboxyl COO\(^{-}\) and the amino-glycan displays at least one sulphate SO\(_3\)\(^{-}\). The two main GAGs that compose proteoglycans are chondroitin-sulphate with valence -2 and keratin-sulphate with valence -1. Because of the high water content of the ground substance, proteoglycans are hydrated and electroneutrality is ensured by sodium cations (Loret and Simoes, 2004).

The compressive properties of soft connective tissues are partly provided by the proteoglycans that resist compression because GAGs repulse each other due to their negative charges (Figure 2.12(a)). The presence of cations Na\(^{+}\) shields the negative charges of the PG, and the mutual repulsive forces decrease with increasing sodium concentration (Loret and Simoes, 2004). Shielding results in decreasing macroscopic compressive moduli when the salt content increases (Figure 2.12(b)) (Dean et al., 2006; Eisenberg and Grodzinsky, 1985).

![Figure 2.12](image)

**Figure 2.12.** Compressive mechanical behaviour of glycosaminoglycans in NaCl solutions. (a) Schematic representation of the experimental setup and (b) experimental results for different concentrations of NaCl. From Dean et al., 2006

Time-dependent mechanical properties of soft biological tissue are strongly influenced by the fluid fluxes that the ground substance undergoes when external loads are applied. Fluid flux phenomena depend on the rheological behaviour of the ground substance itself. It has been suggested (Szwajczak, 2004) that solutions of biopolymers, such as proteoglycans, are able to organize themselves as liquid crystal polymers (LCPs). LCPs behave like non-Newtonian fluids and their viscosity (Figure 2.13) depends on the strain rate (Szwajczak, 2004). In the case of the ground substance of the soft tissue, this behaviour is determined by the combined action of GAGs and hyaluronic acid (Nishimura et al., 1998).
2.3. HISTOLOGICAL CONFIGURATION OF HEEL PAD TISSUES

The mechanical behaviour of heel pad tissues is determined by the mechanical properties of calcaneal fat pad and skin and by the mechanical interactions developing between them. Indeed, the analysis of the histological and morphometric configuration enables the isotropic or anisotropic behaviour and the non-linearity of the mechanical response to be identified.

2.3.1. Calcaneal fat pad tissue

The structure of the calcaneal fat pad is complex and hierarchically organized. The fat pad mainly consists of adipose tissue chambers, separated by connective tissue septa. With a honeycomb configuration (Jahss et al., 1992; Snow and Bohne, 2006) and the dense strands of fibrous tissue are circular or cone-shaped septa. The adipose chambers contain closely packed fat cells with diameters ranging between 100 and 200 μm (Tong et al., 2003) and arranged in lobules within a framework of connective bundles. Larger groups of lobules are enclosed by additional transverse and diagonal bundles of collagen and elastin fibres, there by forming a supporting network (Figure 2.14).
The fibrous tissue strands are firmly attached to the underside of the calcaneus extending to the subcutaneous tissues in the form of a letter U, with the open end of the U pointing toward the calcaneus (Kuhns, 1949). Vertically oriented chambers are located in the central portion of the heel pad. The chambers are small in the region close to dermis, progressively increasing towards the calcaneal perichondrium. In the lateral and posterior regions, chambers are small and more transversally oriented (Rome, 1998). The connective septa configuration varies according to the adjacent adipose chambers geometries. Collagen fibres within septa are spirally arranged and superiorly fixed to bone or other septa, inferiorly to other septa or the dermis (Cichowitx et al., 2009; Weissengruber et al., 2006).

This structural configuration of a network of fibrous elements binding fat cells allows the heel pad to withstand impacts and prolonged pressure loads. Because of the high liquid content characterizing calcaneal fat pad and the complex arrangement of solid components influencing fluid movement, this tissue is considered to have an almost incompressible behaviour, which is a typical feature of soft biological tissues (Weiss et al., 1996). Indeed, the typical features of calcaneal fat pad mechanical response is characterized by large displacements and strains, almost incompressible behaviour, a non-linear stress-strain relationship, and time-dependent effects (Natali et al., 2010).

### 2.3.2. Plantar skin

The skin of the sole, such as the skin throughout the body, is a complex and multilayered structure. The skin is composed of two layers, the epidermis and the dermis, which are made of different tissues and have different functions (Figure 2.15). The
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epidermis is about 0.14-0.8 mm, while the dermis is about 0.9-1.3 mm (Thoolen et al., 2000).

Figure 2.15. Histology of the skin of the sole of the foot, highlighting the several layers of the epidermis.
From Thoolen et al., 2000

The epidermis is hairless and composed of avascular layers of keratinocytic and non-keratinocytic cells. It is made of stratified squamous epithelium and can be further subdivided in five layers. Beginning with the outermost layer, the names are stratum corneum, stratum lucidum, stratum granulosum, stratum spinosum, stratum basale (Figure 2.16). The skin of the sole has a thicker stratum corneum, amore marked stratum granulosum and a thick stratum spinosum than the other skin.

Figure 2.16. A schematization of the several layers of the epidermis

The dermis lying below the epidermis contains a number of structures including blood vessels, nerves, hair follicles, smooth muscle, glands and lymphatic tissue. The dermis is composed of fibre protein (collagen and elastin) embedded in the ground substance and different cell types. The collagen accounts for 70-77% of skin’s dry weight and is
composed mainly by the type I, while the elastin accounts for 4%. The dermis is divided into two regions, papillary dermis and reticular dermis. The papillary layer is the superficial layer of the dermis, made up of loose areolar connective tissue with elastic fibres. These fibers are typically 0.3-3 µm in width and form a relatively open network in which no regular arrangement is apparent. The reticular layer is more dense and thick, and is continuous with the hypodermis. The reticular layer is composed of irregularly arranged fibres, 10-40 µm in width and adipose tissue, and contains sweat glands, sebaceous glands, and blood vessels. This part of the skin is dominated by a dense and three dimensional meshwork of fibres and fibre bundles. The skin is firmly bound to the underlying adipose tissue by fibrous bands of collagen and elastin, which interlace to form a dense network in the dermis (Figure 2.17). The fibre bundles extend in various directions and are arranged irregularly.

Figure 2.17. Thick skin (palms or soles) with epidermis and dermis layers, and the underlying fat pad tissue

The dermis is supported by pressure-absorbing adipose tissue and projects it from lateral shearing stresses.

The arrangement and organization of collagen fibres depends on the location within the body and is described by cleavage lines. This distribution, together with the mechanical response of ground substance, determines the typical response of skin tissues, such as large displacements and strains, almost incompressible behaviour and anisotropic fiber-reinforced configuration (Natali et al., 2012).
2.4. EXPERIMENTAL TESTS ON TISSUES

In vitro experimental tests are a powerful tool to investigate the mechanical properties of biological tissues, specifically with regard to elastic, damaging and time-dependent behaviour. Such experiments are the starting point for the evaluation of constitutive parameters, determined by comparing computational models results and experimental data by means of optimization procedures. The experimental tests to be performed must take into account the specimen geometry, the boundary conditions and the particular mechanical properties under investigation. The material properties of soft tissues published in the literature vary considerably (DeFrate et al., 2006). Many different factors affect the mechanical properties, including age (Woo et al., 1991), alignment and geometry of the specimen (Atkinson et al., 1999), the temperature and hydration of the testing environment (Haut and Powlison, 1990) and time-dependent effects (Johnson et al., 1994; Natali et al., 2004a). Variations in the stress and strain distributions applied to the tissue during mechanical testing might also affect the measurement of material properties. Strains are, in general, measured using the displacement of the clamps used to secure the tissue or by using cameras to record the motion of markers placed on the surface of the tissue (Johnson et al., 1994). Mechanical parameters are generally calculated from load-displacement data with the assumption that the stress and strain distributions are uniform throughout the specimen. However, the testing of soft tissues is technically challenging due to their non-uniform geometry and microstructure (such as variations in fibrous element length, quantity and orientation), to difficulties in securing the tissue so that loads are uniformly applied, and to accurately quantifying the strain and stress fields that are applied. All these factors may result in non-uniform stress and strain distributions and therefore alter the mechanical parameters that are measured (DeFrate et al., 2006). Specimen geometry should be designed in order to avoid local stress-strain intensification that may corrupt the overall load-displacement behaviour of the sample. Finite element analyses can be developed before hand for this purpose. The specimen dimensions should be defined according to the investigation of local or overall tissue properties. Boundary conditions must be chosen by taking into account the particular stress-strain fields to be imposed and the presence or absence of fluid exchange with the neighbouring environment.
In order to correctly interpret the generic stress-strain behaviour of the biological tissue, many deformation fields and loading conditions should be experimentally analysed. Frequently, it is necessary to evaluate the instantaneous mechanical behaviour of the tissue, that is the material response in the absence of time-dependent effects (Natali et al., 2004a). In this situation, mechanical tests should be performed according to a strain rate high enough to ensure that effects of time-dependent processes can be neglected. The investigation of time-dependent mechanical behaviour requires the development of typical tests usually performed on viscoelastic materials, such as creep tests under different stress values, stress-relaxation tests under different strain values and hysteresis tests in different strain rate situations.

Samples are usually cut out of spread tissue by cutting stamps according to specific requirements of geometry and orientation. The geometry of the specimen strictly depends on the specific mechanical test to be performed and must be defined to achieve as far as possible uniform stress-strain fields within the measurement region during testing. The orientation of the sample is of great importance, and accurate sample cutting is imperative in order to ensure the required distribution of fibrous components according to the symmetry of the material to be tested and the anisotropic mechanical properties to be investigated. During testing, the samples must be continuously hydrated with physiological solutions to avoid dehydration, possibly influencing mechanical behaviour. The initial geometry of the specimen is usually evaluated by optical devices. Applied forces are continuously measured during testing, making it possible to calculate nominal stresses (such as the ratio between force and undeformed cross section area). Strains are evaluated by measuring the relative displacements of clamps or by processing video records of the test. In the latter situation markers are placed on the specimen surface in order to achieve a better quantification of the strain field, as markers can be positioned in the most relevant region of the sample, avoiding possible areas of stress-strain intensification occurring near clamps.

### 2.4.1. In vitro experimental tests on calcaneal fat pad tissue

The mechanical properties of calcaneal fat pad tissue are evaluated by taking into account the experimental results of Miller-Young et al. (2002) because this data is both
Histological configuration of soft tissues with relation to the biomechanical behavior of the heel pad tissues

accurate and complete. Mechanical tests were performed on tissue specimens from human cadavers according to unconfined compression loading conditions. Twenty cadaveric feet were obtained from 6 males and 4 females. The calcaneal fat pad was removed from surrounding tissues. Cylindrically shaped samples were cut from the fat pad with a constant diameter and height of about 8 and 10 mm. Five samples were removed from 10 feet in the anterior, central, posterior, medial and lateral regions of the heel pad in directions perpendicular to the skin surface (Figure 2.18).

Figure 2.18. Location from which five cylindrical fat pad samples were cut perpendicular to the skin surface, corresponding to anterior, central, posterior, medial and lateral regions. From Miller-Young (2003)

A sample was cut from the central region of the heel pad of the other 10 feet with the long axis parallel to the skin surface in the medial-lateral direction. Tests were developed up to 50% strain, according to different strain rates, of 0.1, 0.01, 2100 and 4200 %/s. Stress-relaxation tests were performed by loading the specimen up to 40% strain and holding it constant for 60 s (Figure 2.19).

Figure 2.19. Approximately uniform expansion of a fat pad specimen undergoing unconfined compression. From Miller-Young (2003)
Further experimental data from relaxation tests were performed by Ledoux and Blevins (2007). Specifically, experimental tests were performed on specimens from human cadaveric feet. All feet were taken from donors without any influencing pathology. Specimens of plantar soft tissues were acquired from the subcalcaneal, subhallucal, submetatarsal and lateral submidfoot locations (Figure 2.20).

Figure 2.20. The six tissue specimen locations that were tested, including: the subhallucal (A), the 1st, 3rd and 5th submetatarsal (B, C and D), the lateral midfoot (E) and the subcalcaneal (F). From Ledoux and Blevins (2007)

The plantar soft tissue was dissected free from bone, cut into 2 × 2 cm specimens and a scalpel was used to remove the skin. The specimen was placed between two smooth stainless platens in an environment chamber. The specimen was kept at 35°C and close to 100% humidity (Figure 2.21).

Figure 2.21. (a) The compression apparatus: top (A) and bottom (B) platens, specimen (C), heater (D), environmental chamber (E). (b) The apparatus was covered with plastic wrap (A) to prevent leaks, as well as the thermometer (B), water heater (C) and load cell (D). From Ledoux and Blevins (2007)

The target load was 20% body weight, based on the normative ground reaction force and contact area data. The tissue was compressed (unconfined compression) to the
target displacement over a period of 0.1 s and held at a constant strain for 300 seconds. To test the tissue frequency dependence, a series of three triangle waves to the target displacement was conducted at frequency of 10, 1, 0.1, 0.01 and 0.005 Hz.

2.4.2. In vitro experimental tests on skin

The evaluation of mechanical behaviour of heel skin takes into account experimental tests on pig skin specimens, according to data from literature (Ankersen et al. 1999; Liu and Yeung, 2008; Xu et al. 2009; Zhou et al. 2010). Pig skin belly is used because its structural and mechanical response is close to that of human skin in the rear foot region, including its histology and morphology. The thickness of human and pig dermis is similar. For human skin the dermis thickness ranges from 1 to 2.5 mm, while the dermis of the pig varies from 2 to 4 mm (Shergold et al. 2006). In literature, the mechanical properties of the pig skin are investigated using in vitro tests, such as tensile tests and unconfined/confined compression tests. To identify the constitutive parameters of skin tissues, particular attention is performed with compression tests developed by Shergold et al. (2006) and tensile tests developed by Ankersen et al. (1999).

Compression tests were developed on rump skin specimens, typically 7.0 mm and 2.3 mm in diameter and thickness, respectively (Shergold et al. 2006). The tensile tests were developed on specimens from belly skin, measuring 2 mm in thickness, stretched parallel and perpendicular to the principal axis of the skin (Ankersen et al. 1999). The mechanical response of pig skin underlined the anisotropy of skin tissues, due to a preferential orientation of fibrous elements embedded within the isotropic ground matrix of the dermis.

2.5. EXPERIMENTAL TESTS ON THE HEEL PAD STRUCTURE

In vivo experimental tests are developed within a living organism. They can be performed to analyse the mechanical behaviour of tissues by means of non-invasive methods, including ultrasound techniques, or minimally-invasive small probes. Furthermore, in vivo tests are adopted to provide information about the mechanical behaviour of biological structures.
2.5.1. In situ experimental tests

In situ experimental tests were performed on a male cadaveric foot by Miller-Young (Miller-Young 2003). Various indentation tests were performed on the plantar surface of the heel using a testing machine (MTS Systems Corporation, Eden Prairie, Minnesota, USA). The cylindrical indenter was composed by stainless steel (2 cm diameter), polished smooth to minimize friction during testing and attached to a 10 N load cell and actuator (Figure 2.22).

The foot was then fixed in a stainless steel clamp and a sagittal plane X-ray was obtained in order to measure the fat pad and skin thickness. The foot clamp was bolted to the MTS table with the plantar surface of the foot facing up. Indentation tests were performed to approximately 50% of the initial thickness of the heel pad, adopting 175 and 350 mm/s strain rates.

Subsequently, the calcaneal fat pad was exposed by removing a region of the skin (9 x 6 cm) from the plantar surface of the heel (Figure 2.23). Indentation tests were performed on the plantar surface of the fat pad at 0.01 and 350 mm/s strain rates.
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In situ experimental tests were also performed on a right foot from a male donor (age 58 years) by Sirimamilla (Sirimamilla, 2009; Erdermir et al., 2009). Before experimental testing, computer tomography scans in the anterior-posterior direction were obtained, with a scan width of 1 mm. The talocalcaneal joint and the talonavicular joint of the foot were fixed to each other by passing screws and the upper part of the rear foot was attached to an aluminium fixture. In turn, this fixture was jointed to a steel load cell interface component through an aluminium support rod. The spatial load cell recorded the reaction forces and moments generated on the specimen during experiments.

Different tests were conducted on the rearfoot, forefoot and whole foot, although only the rearfoot protocols are considered here. The heel pad was tested initially with a spherical indenter (25.4 mm diameter) (Figure 2.24), then an elevated platform
(86×51×151 mm in width, height and length) was used for the overall rear foot compression test (Figure 2.25).

![Image](image-url)

*Figure 2.25. The cadaveric foot and the platform. From Sirimamilla (2009)*

The indenter and the elevated platform were attached to a moving platform of a 6 degree of freedom robot. All tests were conducted with a mean rate of 40 mm/s.

### 2.5.2. *In vivo* experimental tests

Further experimental activities by Tong et al. (2003) investigated the biomechanical properties of living tissues of the rear foot, using a specific test rig (Figure 2.26). A total of 55 specimens were measured from healthy patients and patients with diabetes and plantar heel pain.

Using an ultrasound probe, the thickness of the heel region was measured with an incremental load from 5 to 30 N in 5N steps, obtaining the relationship between force and compressibility index. The compressibility index was calculated considering the unloaded heel pad thickness and the loaded heel pad thickness measured at 30 N.
Matteoli S. (Fontanella et al., 2012) developed an easy-to-build compression device which was used to apply loading/unloading cycles and record the hysteresis of the heel pad. Such a device is not meant to reproduce the physiological condition of walking or running, but to be a possible clinical device capable to characterize the biomechanics of injured heel pads.

A subject (W=54kg, H=165cm) was enrolled for the compression test applied on her left foot. She declared herself as being healthy and enjoying a sporty lifestyle, with no previous injuries/trauma to any of her feet. She was informed about the conditions of the test that involved no harmful procedures or physical pain.

The measurement part of the device consisted of a load cell (model 31, RDP Electronics Ltd, UK) and a linear transducer (LVDT, RDP Electronics Ltd, UK), both connected to an amplifier (E725, RDP Electronics Ltd, UK). The load cell and the linear transducer were both assembled in a cylindrical aluminium body. One end was fixed to a Plexiglas vertical plate, as shown in Figure 2.27(a). The other end of the cylinder consisted of a threaded shaft which was connected to a stepper motor (PK245-03A, Oriental motor, Japan) by a shaft and a flexible joint, also visible in Figure 2.27(b).
Chapter 2

Figure 2.27. The compression device (a). The stepper motor is connected to the threaded shaft with a shaft and flexible joint (b)

The more the threaded shaft was tightened by the stepper motor, the more compression was applied to the heel pad by a circular shaped indenter (diameter of 40 mm) guided by the threaded shaft. The sole of the foot under investigation was in contact with the vertical plate, while the heel pad touched the circular indenter during the compression/decompression.

The compression device was fixed at one of the extremity of a table. The selected foot was positioned in such a way that the anterior part touched the vertical plate, with the heel pad in front of the indenter. Once the foot was well positioned, it was fixed with six Velcro fasteners (two to strap down the anterior part of the foot, one to keep the heel in front of the indenter, one to stabilize the ankle, and other two to stabilize the lower part of leg), as shown in Figure 2.28(a-b). Specifically, the heel pad was placed with the center almost coincident with the center of the indenter, as shown in Figure 2.28(c).
Two strain rates were used at 0.8 and 1.96 mm/s. The compression test was repeated five times with one-minute break between each trial. The superior limit of the displacement was fixed at 9 mm (to avoid arriving at the end of the thread of the shaft that guides the indenter), while the maximum value of the load was 40N. Furthermore, in order to investigate the stress relaxation characteristics of the heel pad compression tests were repeated with adding a pause of 40 s once the displacement had reached its maximum limit. During this pause the piston was still in contact with the heel pad, making it possible to visualize the decrease in load with time at a constant deformation. After the pause the decompression started until the piston reached the initial position. The entire approach was designed for subsequent applications to the clinical evaluation of injured heel pads.

The analysis of hysteresis curves obtained from experimental tests on the human heel pad allows for the quantitative and qualitative investigation of the mechanical response of the tissue. The knowledge of the biomechanical properties of a healthy and a diseased heel pad may be used for screening patients and for prevention of further complications in the foot.
2.6. REFERENCES


Histological configuration of soft tissues with relation to the biomechanical behavior of the heel pad tissues


Chapter 2


Histological configuration of soft tissues with relation to the biomechanical behavior of the heel pad tissues


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CHAPTER THREE

DEFINITION OF THE NUMERICAL MODELS OF THE HEEL REGION: FINITE ELEMENT MODELS AND CONSTITUTIVE FORMULATIONS

3.1 INTRODUCTION

The numerical modeling approach based on the finite element method is a robust tool to improve the knowledge of foot biomechanics for the integration and reciprocal validation of the experimental data, as well as for evaluating the mechanical behaviour of the foot tissues. The complex mechanical behaviour of the foot implies an adequate modeling in terms of 3D morphometric characteristics and constitutive analysis. To this purpose a solid model of the foot is developed. The model consists of 30 bony segments, including the distal segments of the tibia and fibula and the 28 foot bones. The model of the bones and of the soft tissues of the foot is obtained starting from the elaboration of computer tomography and magnetic resonance images, respectively. Constitutive models describe the relationship between stress and strain history of the tissues and are a necessary tool for the development of numerical models. The definition of constitutive laws requires an in depth knowledge of the inner arrangement of the material and data from mechanical tests, as described in the previous chapters. Furthermore, constitutive models must be formulated by considering mathematical and physical requirements. Details of hyperelastic models are presented in the context of a tool for the characterization of non-linear elastic behaviour of soft tissue. These models can predict the typical characteristics of soft tissues, such as anisotropic configuration, material and geometrical non-linearity and almost incompressible behaviour. The introduction of viscous variables entails the definition of visco-hyperelastic models that make it possible to interpret the typical time-dependence in the mechanical response of soft tissues.
3.2 NUMERICAL MODELS FROM LITERATURE

A numerical modeling approach based on the finite element method is a robust tool to deepen our knowledge of foot biomechanics. It makes it possible to integrate and validate the experimental data reciprocally, evaluate the local interaction phenomena between the foot and insole and look in depth at the mechanical behavior of the foot tissues. The purpose of obtaining refined models of the foot is recognized as mandatory and can be found in several works already present in the literature. Computational modeling, such as the finite element method, has been used increasingly in many biomechanical investigations with great success. This is due to its capacity to model structures with irregular geometry and complex material properties, and also to the ease with which it can simulate complicated boundary and loading conditions in both static and dynamic analyses.

The numerical models of the foot present in the literature have been developed under certain simplifications and assumptions concerning morphological configuration, the mechanical properties of tissues, and loading and boundary conditions. Computer modeling of the soft tissues of the foot, particularly finite element modeling, has been developed over the years to analyze the influence of specific variables on heel pad mechanics, to better understand stress-related injuries such as plantar fasciitis and diabetic ulceration, and to improve orthotics and footwear design for various populations (Nakamura et al., 1981; Thompson et al., 1999; Gefen et al., 2000; Chen et al., 2003; Cheung et al., 2005; Goske et al., 2006; Cheung and Nigg, 2007; Spears et al., 2007; Cheung and Zhang, 2008). Numerical models which have analyzed the human foot as a mechanical structure have used various simplifying assumptions concerning morphological configuration, the mechanical properties of tissues, and loading and boundary conditions. Regarding the morphological configuration, the phalanges are often fused together (Gefen et al. 2000), the ligaments and plantar fascia are modelled with a tension-only truss element (Cheung et al. 2005, 2007; Yu et al. 2008; Wang et al. 2009) without considering the geometry of the foot ligaments. Nonetheless, a fundamental problem arises in connection with the definition of the constitutive formulation of the foot tissues. There are many works that have reported numerical models of the foot, even if none of them describe the actual mechanical properties of the foot tissues.
3.2.1. Hind foot numerical models

The heel pad region has been studied using a numerical method to assess the internal biomechanical behaviour of its tissues, which can explain the causes of heel pain. Modern finite element codes allow the representation of nonlinear and viscoelastic material behaviour which are typical of soft biological tissues.

A two-dimensional numerical model was useful to analyze the effect of heel pad thickness and loading protocol on the mechanical response of the tissues and to provide a robust and standardized measure of heel pad stiffness (Spears and Miller-Young, 2006). Erdermir et al. (2003, 2004, 2006) developed a numerical-experimental approach to characterize heel-pad deformation. They combined the methodology of finite element modelling and ultrasound imaging for an in vivo identification of heel pad properties and to determine subject-specific hyperelastic material parameters (Ogden model) of calcaneal soft tissue in healthy individuals. A two-dimensional heel-pad numerical model with an anatomically realistic surface was developed and the peak pressure was estimated for different subjects analyzed.

A three-dimensional model of the heel region, composed of the fat pad and skin, were developed to study the effect of the mechanical properties of heel tissues on the load-bearing function of the heel pad and human locomotion. Spears et al. (2005) created a nonlinear time-dependent and three-dimensional finite element model of the heel pad from CT images, and distinguished between the fat pad and the heel skin. Internal stress and external plantar pressures were then investigated for different forces, loading rates and angle of foot inclination in the sagittal plane. The tissues were modelled with a Mooney-Rivlin reduced polynomial hyperelastic material model with viscoelastic behaviour. The finite element method provided information about the localization into the heel pad tissue of the compression stress. Peak internal compressive stress was influenced by the loading rate and by the inclined position.

More recently, Gu et al. (2010) studied the mechanical behaviour of heel skin layer through a combined experimental and numerical approach. The finite element model was obtained using CT and MR images. The first-order Ogden model were used to characterize heel fat pad and the heel skin. The mechanical behaviour of the hind foot during the heel strike was evaluated with potential variation of the skin stiffness based on a subject-specific finite element model and experimental testing. Sopher et al.
(2011) created an anatomically-realistic three-dimensional finite element model of the heel region to study the risk of heel ulcers in bedridden patients. The geometry of this model was constructed from cryosections of a foot. The behaviours of the fat pad and skin were described using the Ogden model. They simulated a heel resting on supports with different stiffness at the upright and inclined foot posture, thereby evaluating the effects on strains and stresses within the fat pad.

Quian et al. (2010) developed a simplified heel pad finite element model comprising a reticular fiber structure and fat cells based on Magnetic Resonance images to investigate the foot pad behaviours under impact during locomotion. The reticular fiber structure was considered to be homogeneous, isotropic and linear elastic, while the fat cells were defined as a second-order polynomial model. The results showed that the heel pad model with fat cells had greater capacity in impact attenuation and energy storage than the model without fat cells.

Together with experimental techniques, numerical models allow for an accurate analysis of the interaction phenomena occurring between foot and footwear products, and to estimate the results taking reliability and comfort into consideration. Integration of data from experimental tests and numerical analysis, accounting for different footwear materials and configurations, leads to a detailed and accurate evaluation (Shariatmadari et al., 2010; Barani et al., 2005; Li et al., 2009). Some studies report the influence of specific footwear variables, such as the properties of materials and the geometrical configuration on heel pad mechanics. In literature, both two dimensional (Verdejo and Mills, 2004; Goske et al., 2006) and three dimensional (Chen et al., 2003; Even-Tzur et al., 2006; Cheung et al., 2008) finite element models are proposed, but the effects of the interaction between fat pad tissue and skin is not considered.

The adoption of two dimensional models determine an approximation in the evaluation of the overall mechanical problem. For example, Verdejo and Mills (2004) developed a two-dimensional model of the heel pad and of a running shoe midsole. The heel pad tissue and the foam were described with a Ogden strain energy function. The mechanical properties of foams used in footwear products were studied in depth by Sharitmadari et al. (2009; 2010). In this study, numerical analyses were adopted to investigate the effect of varying footwear temperatures on plantar stresses. The results
showed that temperature can affect the mechanical response of the heel pad tissue and consequently the shock attenuation properties.

Using a two-dimensional finite element model of the heel pad and shoe during a simulation of static standing, Spears et al. (2007) quantified the potential effect of confinement on internal heel pad stress. The tissues were characterized by the Hyperelastic model (Ogden, 1884) and the geometry was taken from an elaboration of MR images. The shoe was composed of the insole, the midsole and the heel counter. The effect of the heel counter was to reduce both compressive and shear stress in the fat pad and in the skin. The effect of the heel counter was also evaluated by Goske et al., (2006). Here, a finite element analysis was used to provide an efficient computational framework for investigating the performance of a large number of designs for optimal plantar pressure reduction. Indeed, combinations of three insole conformity levels and three insole materials were simulated during the early support phase of gait. Plantar pressure was also evaluated when the heel was modelled by loading the bare foot on a rigid surface and on foam mats.

More recently, Luo et al. (2011) developed a two-dimensional model of heel pad tissues to analyse the effects of various insole designs and materials on the resulting stress, strain and strain energy density. The skin was modelled with a Jamus-Green-Simpson material model, and the fat pad tissue with a second order Ogden model.

Three dimensional models presented adopt a mechanical characterization of the heel pad tissues by means of viscoelastic formulation. Even-Tzur et al. (2006) studied heel pad stress and strains during heel strike in running and developed a three dimensional model of the heel pad and the sole of the running shoe. They considered the viscoelastic constitutive model of both the heel pad and the EVA midsole. Simulations showed that heel pad stresses and strains were sensitive to viscous damping of the EVA.

### 3.3 DEVELOPMENT OF VIRTUAL SOLID MODELS FOR THE DEFINITION OF NUMERICAL MODELS

The definition of a finite element model of the foot starting with the elaboration of the DICOM images generated by x-ray computed tomography (CT) and magnetic resonance imaging (MRI). The DICOM images are processed with a medical imaging
and editing software (MIMICS 10.01) that is used to obtain the primary 3D models (triangulate model) using a density segmentation technique. The generated primary 3D models were exported as geometrical files for a CAD system (UGS NX3, UGS Corporate, Plano, TX) that made the assembly and some 3D geometrical operations possible. Finally, the virtual solid models described above were exported in parasolid format to be discretized in finite element models by a specific software (Patran, MSC.Software Corporation, Santa Ana, CA). The procedure adopted is a guideline that does not impose standards or processing parameters set for all cases (Figure 3.1). Each model is defined by subjective evaluations and reaches a good compromise between an acceptable level of accuracy and regular surfaces (De Souza, 2007; Korioth and Versluis, 1997; Vannier et al. 1997).

![Figure 3.1. Modeling methodology](image)

### 3.3.1. Medical imaging analysis and selection

Medical imaging is a branch of the medical field which involves the use of technology to take images of the inside of the human body in a way that is as non-invasive as possible. These images are used in diagnostics and in routine healthcare for a variety of conditions. There are a number of different types of technology used in medical
imaging, however some of the most famous types are the x-ray, the CT and the MRI (Ayache 1995; Stytz et al. 1991).

The reconstruction of the virtual solid models of the foot requires the acquisition of images from CT or MRI. Specifically, the reconstruction of the skeletal structure of the foot is obtained by acquiring and processing CT images (Figure 3.2(a)), in which the bone tissues are highly distinct from other tissues. While the reconstruction of the soft tissues requires the elaboration of MRI images (Figure 3.2(b)), that make it possible to distinguish the different soft tissues of the foot with different grey levels (Natali et al., 2012; Matteoli et al. 2010; Reach et al. 2007; Rosenberg et al., 2000).

![Figure 3.2. CT (a) and MRI (b) image of the foot](image)

The sequence of collecting tomography scans and the magnetic resonance imaging of the patient is available in files sorted by type DICOM (Digital Imaging and Communications in Medicine), a standard communication protocol created by the National Electrical Manufacturers Association (NEMA). The DICOM standard has been introduced to define a mode of communication, storage and management of medical information, with particular regard to biomedical images. The reconstruction of the primary 3D anatomical structures is obtained by the use of a medical imaging density segmentation software. The DICOM image files generated in the CT and MRI are constituted by pixels with different grey intensities. The different
intensity fields correspond to different material densities presented in the anatomical structures (Antunes et al., 2010).

The reconstruction of the virtual solid model of the bones of the foot is obtained from the direct analysis of tomographic slices generated by the lower limb. Initially, it is necessary to impose the proper axial orientation to imported images. Then the main screen appears (Figure 3.3), which is divided into two parts. The left side has four panels, representing the frontal, transverse and sagittal plane and one additional space for the subsequent viewing of the 3D solid model of the selected structure in 3D. The right side is the Project Management where a list of all the objects created is reported, which can be viewed or modified (Figure 3.3).

To distinguish hand highlight the bone tissues, the procedure used a threshold method. The program sets the default interval of grey levels in the Hounsfield scale for the selection of different tissues from CT images. In particular, the bone tissue of an adult has grey levels between 226 and 3071 of the Hounsfield scale. In order to obtain a reliable selection of bone components, subsequent operations are necessary. The Project Management enables us to create the virtual solid model of the selected tissue, as an enclosed volume delimited by a 2D triangular mesh (De Souza, 2007). During this operation, it is possible to choose the accuracy and the level of image quality to be implemented (Figure 3.4).
Definition of the numerical models of the heel region: finite element models and constitutive formulations

With regard to the soft tissue the same procedure is adopted on the MRI image. The adipose tissue, the skin and the Achilles tendon is the soft tissue that is possible to obtain by means of MRI images. An example is reported for the Achilles tendon (Figure 3.5).

To elaborate the virtual solid models obtained, they must be saved in .STL format to guarantee the subsequent importation into the processing software.

3.3.2. Elaboration of triangulated models

The files in .STL format are imported by the modelling software, where a shell is registered in the project window. The shell is defined as an enclosed volume delimited by 2D triangular elements, usually related to each other, which form a polygon mesh. Working with a mesh is advantageous because of the speed of processing and the multiple control functions. The disadvantages are related to the approximation with which the surfaces are represented, which are formed by small triangles.

In order to obtain a virtual solid model which describes the real configuration of the biological structures, the elaboration of the primitive model are intended to define the shape of the structure as smoothly and clean-cut as possible. At the same time, the...
purpose is to identify the actual geometry of the structure without omitting details, even if some approximations during the procedure are inevitable.

With the process of segmentation previously described, the virtual solid model obtained is characterized by an irregular surface, with gaps, scattering and in some cases a constant trend in steps due to the stratification of the medical images. The software makes it possible to select, by specific commands, the regions of the entire shell which present the major defects and must be corrected.

For this purpose, the first step involves the regularization of the surface, starting from the operation of smoothing that solves the most minor defects, while maintaining the overall shape. This operation can be divided in global and local smooth and it is possible to set the method of intervention (laplacian, loops or bends), the weight and the number of passes. The defects of the surface often cannot be resolved completely, taking into account only smoothing. The software offers several tools that are used in these cases, usually applied to small regions with the aim of not deforming the overall surface. The surfaces that present error and discontinuity can be selected, deleted and replaced with a regular shape which follows the trend of the surrounding region. For this purpose it is possible to reconstruct and close the cavities created during the segmentation procedure. This operation can be done with one hole or with all holes together and enable closing with a flat, smooth or curved shape. It is also possible to create links between the contours of a large cavity, so as to reduce the extension to two smaller gaps and allow the next stage of closing to follow a precise pattern of the form.

When is necessary deform an area, the software allows to deform and transform the shell obtained. For example it is possible to create a reticule around the selected section of the shell and to execute the deformation by moving the balls placed at the top and along the sides of the reticule.

An example of surface elaboration is reported for the calcaneus bone (Figure 3.6). The software offers also the possibility to re-triangulate the overall surface, transforming all triangles into equilateral triangles.
3.3.3. Definition of virtual solid models

From the description of rough tessellated geometry an exact mathematical model is then obtained (Antunes et al., 2010).

The mathematical representation is called NURBS (Non-Uniform Rational B-Splines). The NURBS curves are polynomial curves that come close to the boundaries of free-form. The NURBS curves have the advantage of representing arbitrary shapes with mathematical precision, while maintaining control over the shape of the curves through their nodes and control points, which can be directly manipulated. The procedure uses a fitting algorithm, which consists of a series of patches that fit the discrete data, such as point cloud formed by the vertices of all triangles. The patches are usually of the same mathematical nature and are linked together by the conditions of tangency or curvature.

The software allows to obtain a mathematical surface directly from the tessellated surface (Figure 3.7). Before starting the automatic surfacing, it is possible to set some parameters such as the total number of surfaces that you want to create and the number of control points of the surface. Finally, it is useful to enforce certain conditions of the process: the generation of a uniform surface, the adhesion of the created patches to the original surface, removal of the roughness and the possibility of changing the boundaries of the patch after the execution. With a higher number of chosen areas, the resulting model is clearly more accurate. This operation is limited only by the computational capacity of computers, which can take a long time to achieve.
To share the geometries with the software Unigraphics (UGS NX3, UGS Corporate, Plano, TX) all the surfaces must be exported as an IGES (Initial Graphics Exchange Specification) file. This operation increases the degree of approximation of mathematical models of the surfaces. In reference to solid models developed, it is necessary to make changes to portions of the surface that the previous method has not been able to achieve correctly. With the modelling package, the freeform surfaces created can be modified and corrected.

Solid models of the different components (bones, skin, soft tissue, ligaments and tendon) are exported in parasolid format to be discretized in finite element models (Figure 8).
3.3.4. Definition of the numerical models from virtual solid models

The definition of the numerical model of the foot is performed by the finite element discretization of the virtual solid models described previously. The discretization of the solid model is done using a specific software (Patran, MSC.Software Corporation, Santa Ana, CA).

In order to obtain a numerical model with a proper and accurate discretization, it is necessary, first of all, to proceed with the discretization of the surfaces of the virtual solid models with triangular elements.

These elements are defined by three nodes, with side length of about 2 mm. With this procedure the 2D mesh of all solid models is obtained. After this operation, the area of interaction between the models is modified to make them congruent. This operation is achieved by removing some triangular finite elements and defining new ones.

On obtaining the different numerical models which are consistent with each others, it is necessary to define the 3D (Figure 3.9) mesh using tetrahedral elements.

![Figure 3.9. Finite element model of the bone structures (a), the soft tissues (b) and the skin (c) of the foot](image)

3.4 CONSTITUTIVE MODELING OF SOFT TISSUE

Soft biological tissues are usually composed of fibrous elements (such as collagen or muscle fibers) embedded in an isotropic and almost incompressible ground matrix and, from a mechanical point of view, can be considered as composite materials. The local distribution of fibers determines the material symmetry characteristics (Spencer, 1992), such as in tendon, ligaments and skin. The structural conformation
rearrangements developing within the material when external loads are applied demonstrate a strongly non-linear behaviour (Natali et al., 2005). In a first approach to the problem, non-linear behaviour can be analysed by hyperelastic models. Structural rearrangements, with particular regard to liquid fluxes, require time to reach equilibrium. It follows that the typical time-dependence of mechanical behaviour can be phenomenologically interpreted by visco-hyperelastic models (Limbert and Middleton, 2004; Natali et al., 2004; Fung, 1981). The visco-hyperelastic model is used to represent the calcaneal fat pad with regard to its material and geometric non-linearity, to its almost incompressible behaviour and time dependent response. The anisotropic characteristics of skin tissues induced by collagen fibres are considered using a fiber-reinforced hyperelastic model.

3.4.1. **Formulation of constitutive model**

The balance equations of classical continuum mechanics (such as the balance of mass, linear momentum and rotational momentum) are common to most bodies in nature. These equations are insufficient to describe the behaviour of a body because they do not distinguish between different types of materials. Therefore, it is necessary to introduce hypotheses, called constitutive assumptions, that characterise the mechanical behaviour of the material. Three types of constitutive assumptions are generally considered: constraints on the possible motions the body may undergo (such as rigidity or incompressibility constraints); assumptions on the form of the stress tensor (such as the imposition that the stress tensor be a pressure in the case of ideal fluids); and constitutive models that relate the stress to the strain history. Constitutive assumptions are a necessary tool for the mathematical description of the mechanical behaviour of a material. They are usually developed by accounting for general mathematical and physical requirements, the structural conformation of the material and data from mechanical tests.

In the field of continuum mechanics, the second principle of thermodynamics states that, during a generic mechanical process, the work of internal stresses must be higher or at least equal to the energy reversibly stored within the material. In other words, mechanical work developed on the material must be higher or at least equal to the mechanical energy that the material can give back. In the latter situation, the material
is conservative or hyperelastic, while in the former case the material has dissipative behaviour and a portion of the mechanical work is either used to irreversibly change the structural conformation of the material itself or it is transformed to heat. Typical examples of dissipative phenomena are damage, viscous and plastic effects.

The mathematical formulation of the second principle of thermodynamics is stated by the Clausius-Duhem dissipative inequality (Holzapfel, 2000):

$$D_{\text{int}} = \frac{1}{2} S : \dot{C} - \dot{\psi} \geq 0$$

(3.1)

where $S$ is the second Piola-Kirchhoff stress tensor, $C$ is the right Cauchy-Green strain tensor (Gurtin, 1981), $\psi$ is the Helmholtz free energy function and $D_{\text{int}}$ is the rate of internal dissipation. The Helmholtz free energy defines the portion of the work of internal stresses that is reversibly stored within the material during the generic stress-strain path and specifies the current mechanical state of the material itself.

According to the principles of determinism the mechanical state of the material is determined by its strain history (Wang and Trusdell, 1973). It follows that the Helmholtz free energy depends on the current strain state and dissipation phenomena that eventually develop during the strain history:

$$\psi = \psi(C, x^i)$$

(3.2)

where $x^i$ are internal variables that are associated with the development of dissipation phenomena. By equations (3.1), (3.2) and the chain rule, the following formulation of the rate of internal dissipation can be achieved:

$$D_{\text{int}} = \left( \frac{1}{2} S - \frac{\partial \psi}{\partial C} \right) : \dot{C} - \frac{\partial \psi}{\partial x} : \dot{x}^i \geq 0$$

(3.3)

According to the principle of universal dissipation (Wang and Trusdell, 1973), the previous inequality must be satisfied for any process the material can undergo (such as
∀ \(C, \dot{C}, \dot{x}, \dot{x}'\), entailing the following expressions for the stress and the rate of internal dissipation:

\[
S(C, \dot{x}') = 2 \frac{\partial \psi(C, \dot{x}')}{\partial C} \quad (3.4)
\]

\[
D_{int} = - \frac{\partial \psi}{\partial \dot{x}'} \cdot \dot{x}' \geq 0 \quad (3.5)
\]

The definition of the constitutive model consequently requires to specify the dependence of the Helmholtz free energy on the current strain state and the internal variables. Furthermore evolution laws of the internal variables during the generic stress-strain history must be defined in accordance with the condition (3.5).

### 3.4.2. Material symmetry

Any symmetry in the structural conformation of the material is reflected by symmetry in its mechanical properties.

Material symmetry is mathematically characterised by a symmetry group \(\mathcal{G}_Q\) that includes orthogonal transformations that leave the Helmholtz free energy function unchanged when applied to the material before any deformation process (Limbert and Taylor, 2002). If the sub-components of the material are equally distributed along all the directions, the material is said to have isotropic behaviour. In this event, mechanical properties do not depend on the specific direction and the symmetry group is composed by all the orthogonal transformations.

The Helmholtz free energy function is said to have isotropic behaviour (Gurtin, 1981) and its dependence on the current strain state can be specified by the three principal invariants of the right Cauchy-Green strain tensor (Holzapfel, 2000):

\[
I_1 = tr(C), \quad I_2 = 1/2 \left[ I_1^2 - tr\left(C^2\right) \right], \quad I_3 = J^2 = det(C) \quad (3.6)
\]

where \(J = det(F)\) is the deformation Jacobian and \(F\) the deformation gradient.
When sub-components of the material are distributed according to preferential directions, the material is characterised by anisotropic behaviour and mechanical properties change with the direction considered.

Soft tissue is usually composed of fibrous elements embedded within an isotropic ground matrix. The distribution of fibers along preferential directions is responsible for its typical anisotropic behaviour (Spencer, 1992).

With regard to biological elements, such as tendons and ligaments, it is often possible to assume fibrous elements locally aligned along only one direction (Figure 3.10(a)). The distribution of fibers in the undeformed configuration is usually described by a unit vector field $\mathbf{a}_0$ that is locally tangent to fibrous components (Figure 3.10(b)).

![Figure 3.10. Schematic representation of a transversally isotropic fiber-reinforced composite material (a). Mathematical description of the distribution of collagen fibers by the unit vector field $\mathbf{a}_0$ into a talo-calcaneal ligament (b) (histology from Keller et al., 2010)](image)

The local vector $\mathbf{a}_0$ defines the preferential direction of the material, while in the plane normal to $\mathbf{a}_0$ (the isotropic plane) mechanical properties are the same along all directions. The group of symmetry is composed by the orthogonal transformations around the axis $\mathbf{a}_0$ and the material is said to have a transversally isotropic behaviour:

$$ G_0 = \{ \mathbf{Q} \in \text{Orth}^+ | \mathbf{Qa}_0 = \mathbf{a}_0 \} $$

(3.7)

The dependence of the Helmholtz free energy function on the current strain state can be specified by the three principal invariants of the right Cauchy-Green strain tensor plus two further invariants (Spencer, 1992):
The fourth invariant equals the square of the stretch $\lambda$ along the preferential direction $a_0$. The fifth invariant specifies the influence of shear conditions on the behaviour of the fibers. The concept is explained in Figure 3.11, whereby a simple shear deformation is considered for two different spatial orientations of the fibers: either aligned in the shear plane or aligned perpendicularly to the shear plane. It is clear that the fourth invariant is constant in both the cases, while the fifth invariant assumes different values.

\[
I_4 = a_0^\cdot \mathbf{C} a_0 = \lambda^2, \quad I_5 = a_0^\cdot \mathbf{C}^2 a_0
\] (3.8)

In the case of materials with isotropic matrix reinforced by two or more groups of fibers with different orientation, the use of further invariants is required (Limbert and Taylor, 2002; Spencer, 1992).

### 3.4.3. Hyperelastic constitutive models

A material is said to have hyperelastic behaviour if the internal dissipation equals zero for every stress-strain path (Truesdell and Noll, 1992). This means that the rate of internal dissipation ($D_{int}$) equals zero for any process the material undergoes:
It follows that the Helmholtz free energy function depends only on the current strain state and the work of internal stresses is independent on the stress-strain path (Belytschko et al., 2001). The Helmholtz free energy function is consequently a potential of the strain state and is usually called strain energy function \( \psi = W(C) \). The stress-strain relationship can be calculated according to the equation:

\[
S(C) = 2 \frac{\partial W(C)}{\partial C}
\]

The formulation of a hyperelastic model only needs to specify the dependence of the strain energy function on the strain state, carefully accounting for inequalities which define the material stability requirements (Schroder et al., 2005; Marsden and Hughes, 1983). The strain energy function is usually formulated by analyzing the structural conformation of the material and experimental data that characterise its mechanical behaviour.

### 3.4.3.1. Isotropic hyperelastic model

The strain energy function of an isotropic material, as calcaneal fat pad, is an isotropic function of the current strain state (Gurtin, 1981). It can be expressed in terms of the three principal invariants of the right Cauchy-Green strain tensor:

\[
W(C) = W(I_1, I_2, I_3)
\]

According to equations (3.10) and (3.11), the stress-strain relationship can be reformulated by applying the chain rule:

\[
S = 2 \frac{\partial W}{\partial C} = 2 \sum_{i=1}^{3} \frac{\partial W}{\partial I_i} \frac{\partial I_i}{\partial C}
\]
the derivatives of the principal invariants can be computed from (3.6):

$$\frac{\partial I_1}{\partial \mathbf{C}} = 1, \quad \frac{\partial I_2}{\partial \mathbf{C}} = I_1 \mathbf{1} - \mathbf{C}, \quad \frac{\partial I_3}{\partial \mathbf{C}} = I_3 \mathbf{C}^{-1}$$

(3.13)

where $\mathbf{1}$ is the rank two unit tensor. Soft biological tissues usually contain a large liquid content, composed predominantly of water. Part of this water is chemically bound to the solid matrix and, therefore, cannot move through the tissue. In addition, the quantity of water that is free to move within the tissue can be considered to be bound if the tissue is strained with high rates of loading. Indeed, in such cases, low permeability values represent an effective obstacle to fluid motion. A direct consequence of this structural conformation is the fact that the tissue can behave like an almost-incompressible material. Therefore, a suitable numerical framework for the analysis of the mechanical behaviour of soft tissues requires to split the strain energy function and the stress response of the isotropic matrix in volumetric and volume-preserving (or iso-volumetric) parts (Flory, 1961). The strain energy function is then defined by the following form:

$$W_m(I_1, I_2, I_3) = W_{mv}(I_3) + W_{mi}(\tilde{I}_1, \tilde{I}_2)$$

(3.14)

where $\tilde{I}_1, \tilde{I}_2$ are the two principal invariants of the volume-preserving part of the right Cauchy-Green tensor $\tilde{\mathbf{C}} = I_3^{-1/3} \mathbf{C}$. $W_{mv}$ is related to the volumetric part of strain and $W_{mi}$ to the volume-preserving part.

### 3.4.3.2. Fiber-reinforced hyperelastic model

In the case of transversally isotropic materials, such as skin, it can be demonstrated (Spencer, 1992) that the strain energy function depends on five invariants, including the three principal invariants of the right Cauchy-Green tensor plus two further invariants:

$$W(\mathbf{C}) = W(I_1, I_2, I_3, I_4, I_5)$$

(3.15)
The invariants $I_4$ and $I_5$ arise directly from the anisotropy introduced by the fibers and specify the contributions to the strain energy function from the properties of the fibers and their interaction with the other material constituents (Weiss et al., 1996). The stress response is then given by:

$$
S = 2 \frac{\partial W}{\partial C} = 2 \sum_{i=1}^{5} \frac{\partial W}{\partial I_i} \frac{\partial I_i}{\partial C}
$$

(3.16)

where the derivatives of the fourth and fifth invariants are defined as follows:

$$
\frac{\partial I_4}{\partial C} = a_0 \otimes a_0, \quad \frac{\partial I_5}{\partial C} = a_0 \otimes a_0 + \text{Ca}_0 \otimes \text{a}_0
$$

(3.17)

The hyperelastic behaviour of soft biological tissues arises from the properties of the ground matrix, the fibers and their interaction. Consequently, the strain energy function can be additively decomposed:

$$
W(I_1, I_2, I_3, I_4, I_5) = W_m(\tilde{I}_1, \tilde{I}_2, \tilde{I}_3) + W_f(I_4, I_5) + W_{mf}(I_1, I_2, I_3, I_4, I_5)
$$

(3.18)

where $W_m$ represents the material response of the isotropic ground matrix, $W_f$ represents the contribution from fibers and $W_{mf}$ is the contribution from interactions between fibers and the matrix.

In the general framework of fiber-reinforced composite materials, the function $W_{mf}$ specifies the effects that are determined by the varying stiffness that characterises fibers and the matrix. These effects are particularly relevant when strong bonds effectively take place between the two phases. With regard to soft biological tissues the bonds between fibers and ground substance are usually very weak. It follows then, that the term $W_{mf}$ can be neglected (Limbert and Middleton, 2006; Natali et al., 2003). As previously reported, the almost incompressible behaviour of the ground matrix suggests that its contribution can be split into volumetric ($W_{mv}$) and iso-volumetric ($W_{mi}$) parts, leading to the following formulation of the strain energy function:
3.4.4. Visco-hyperelastic model

The calcaneal fat pad tissue exhibits a viscoelastic behaviour that must be described with visco-hyperelastic models. The time-dependent behaviour of soft biological tissues is due to the development of structural conformation rearrangements (such as fluid fluxes through the solid skeleton, sliding of macromolecules, etc.) that develop during loading of the material. From a phenomenological point of view, typical examples of viscous effects are creep, stress-relaxation and hysteresis. Rearrangement phenomena are usually defined as viscoelastic processes and can be associated with internal variables $q^i$, which express material evolution during the stress-strain history from a phenomenological point of view. The mechanical state of the material is described by a specific configuration of the Helmholtz free energy function:

$$\psi = \psi(C, q^i)$$  \hspace{1cm} (3.20)

The specific formulation of the Helmholtz free energy can be developed accounting for mechanical models that are capable to phenomenologically describe the behaviour of the material. Within visco-elastic theories the Zener model (Figure 3.12) is frequently adopted. This model is made up of an equilibrium spring and viscoelastic branches connected in parallel (McCrum et al., 1997; Natali et al., 2004).

$$W(I_1, I_2, I_3, I_4, I_5) = W_{Aw}(I_1) + W_{Aw}(I_2) + W_f(I_4, I_5)$$  \hspace{1cm} (3.19)
Every viscoelastic branch phenomenologically represents a viscoelastic process, which is characterised by a relative elastic stiffness $\gamma^i$ and a relaxation time $\tau^i$. The relative stiffness describes the contributions of the viscous processes to the whole instantaneous stiffness of the material, which is the stiffness during a straining process characterised by an infinite strain rate. The relative stiffnesses have to satisfy the following relationship (Simo and Hughes, 1998):

$$\gamma^\infty + \sum_{i=1}^{n} \gamma^i = 1$$ (3.21)

where $\gamma^\infty$ is the relative stiffness of the equilibrium spring, which defines the material behaviour during an equilibrium strain process (i.e. a process that is characterised by a strain rate approaching zero). The analysis of the Zener model suggests the formulation of a Helmholtz free energy capable of interpreting the visco-hyperelastic behaviour:

$$\psi(C, q^i) = W^\infty(C) + \sum_{i=1}^{n} \psi^i(C, q^i)$$ (3.22)

where $W^\infty$ is an hyperelastic potential that defines the behaviour of the equilibrium spring, while $\psi^i$ is the Helmholtz free energy related to the $i^{th}$ viscous branch. In the present approach the variables $q^i$ describe the non-equilibrium stresses associated with the viscous processes and the following formulation of $\psi^i$ is assumed (Natali et al., 2004):

$$\psi^i(C, q^i) = W^i(C) - \frac{1}{2} q^i : \dot{C}$$ (3.23)

where $W^i$ is an hyperelastic potential associated with the $i^{th}$ spring, while the second term is the energy dissipated because of the viscous process. The potentials $W^\infty$ and $W^i$ can be related to an instantaneous hyperelastic strain energy $W^0$, as $W^\infty = \gamma^\infty W^0$
and $W^i = \gamma W^0$. The stress-strain relationship can be evaluated according to equation (3.4):

$$S(C,q^i) = 2\frac{\partial W^\infty}{\partial C} + \sum_{i=1}^{n} \left[ 2\frac{\partial W^i}{\partial C} - q^i \right] = S^\infty (C) + \sum_{i=1}^{n} S^i (C,q^i)$$

(3.24)

The evolution law for viscous variables $q^i$ can be obtained by means of the Zener model. The imposition of a strain compatibility condition on the whole Zener model and a stress equilibrium condition on every single viscoelastic branch makes it possible to achieve the following equations:

$$\dot{q}^i + \frac{1}{\tau^i} q^i = 2\frac{\gamma^i}{\tau^i \gamma^\infty} \frac{\partial W^\infty}{\partial C}$$

(3.25)

The above expression is a system of first-order ordinary differential equations that, with the introduction of the condition of null initial non-equilibrium stresses, as $\lim_{t \to 0} q_i = 0$, can be solved in the following convolution integrals:

$$q^i(t) = \frac{\gamma^i}{\gamma^\infty \tau^i} \int_{0}^{t} \exp \left( -\frac{t-s}{\tau^i} \right) S^\infty(s) ds$$

(3.26)

### 3.4.5. Definition of strain energy function

The formulation of a hyperelastic constitutive model requires to specify the strain energy function. The action is performed starting from the analysis of the structural conformation of the material and experimental data from mechanical tests (Natali et al., 2010). As previously reported, soft biological tissues are composed of fibres embedded within an isotropic ground matrix. This configuration might determine anisotropic behaviour when the fibres are locally aligned along preferential directions. Furthermore, the high liquid content characterising soft tissue requires the use of formula applicable to almost-incompressible materials. The various terms of equation (3.19) will be now specified. The volumetric contribution to strain energy is mainly
due to the mechanical response of liquid components and the electrostatic interaction phenomena taking place inside the ground substance. In fact, proteoglycans are characterised by a distribution of negative charges that interact with water positive sodium ions. Compressive loads determine a tendency towards squeezing water out of the network and bringing the proteoglycans closer. These phenomena are counteracted respectively by electrostatic interactions of proteoglycans and sodium ions (Loret and Simoes, 2004). The application of tensile load determinates a lower water squeezing tendency and proteoglycans tend to get further, leading to a different tensile response from the loading condition. The following formulation of the volumetric term is able to account for previous characteristics:

\[
W_{\text{v}}(I_3) = \frac{K_v}{2 + r(r+1)} \left[ \left( I_3^{1/2} - 1 \right)^2 + I_3^{-1/2} + r I_3^{1/2} - (r+1) \right] \quad (3.27)
\]

where \( K_v \) and \( r \) characterise the material compressibility and can be related to the tangent bulk modulus:

\[
K^T_v = \frac{K_v}{2 + r(r+1)} \left[ 2 + r(r+1) I_3^{-(r+2)/2} \right] \quad (3.28)
\]

It is possible to assume that \( K_r \) is the tangent bulk modulus in the unstrained configuration. The characteristic non-linear response of the tissue, as outlined by experimental data (Nishihira et al. 2003), suggests the assumption of an exponential formulation for the volume-preserving term (Fung 1981):

\[
W_{\text{m}}(\tilde{I}_1, \tilde{I}_2) = \frac{C_1}{\alpha_1} \left\{ \exp \left[ \alpha_1 (\tilde{I}_1 - 3) \right] - 1 \right\} + \frac{C_2}{\alpha_2} \left\{ \exp \left[ \alpha_2 (\tilde{I}_2 - 3) \right] - 1 \right\} \quad (3.29)
\]

where \( C_1, C_2 \) characterise the shear stiffness of the tissue, as \( G = 2(C_1 + C_2) \), while \( \alpha_1, \alpha_2 \) are parameters that regulate the non linearity of the material response, with reference to experimental results. The formulation of the iso-volumetric term is proposed in a general form. However, the correct evaluation of both contributions...
from first and second invariants \( I_1, I_2 \) requires a large set of experimental data from tests developed according to different strain states, such as uniaxial, biaxial and shear. The \( I_1, I_2 \) response terms are highly co-linear for uniaxial deformations and consequently the contribution of the second invariant is usually neglected when data from uniaxial tests only are at disposal.

With regard to skin it is possible to assume a preferential fibers direction. The mechanical contribution of fibers can be described by considering their microstructural organization (Redaelli et al. 2003; Ottani et al. 2001). In the unstrained configuration, fibers are usually characterised by a typical wavy conformation. When tensile load is applied along the direction of the fibers, they first uncrimp and then get stretched. This mechanism determines a strongly non-linear mechanical response that can be described by an exponential formulation of the fibers strain energy contribution (Natali et al. 2003; Weiss et al. 1996):

\[
W_f (I_4) = \frac{C_4}{(\alpha_4)^2} \left\{ \exp[\alpha_4 (I_4 - 1)] - \alpha_4 (I_4 - 1) - 1 \right\}
\]  

(3.30)

where \( C_4 \) is a constant that defines the initial stiffness of the fibers, as \( E_f = 4C_4 \), while \( \alpha_4 \) depends on the initial wavy conformation of fibers (Natali et al. 2004, 2005). The influence of the fifth invariant is omitted in this formulation, because the contribution of the fibers to the tissue shear behaviour is often really poor (Weiss et al. 1996). When compressive loads are applied to soft biological tissues, fibers undergo micro-buckling phenomena. Consequently, contribution of fibers to overall mechanical response of the tissue is really lower for a compressive situation than a tensile one. Exponential formulations as (3.30) are able to account for this peculiar aspect.

The specific stress-strain relationship can be calculated according to equations 3.16, 3.27, 3.29 and 3.30:

\[
S_{mv} = \frac{K_v}{2 + r(r+1)} \left[ 2I_3^{1/2} \left( I_3^{1/2} - 1 \right) - rI_3^{-1/2} + rI_3^{1/2} \right] C^{-1}
\]  

(3.31)
Definition of the numerical models of the heel region: finite element models and constitutive formulations

\[
\mathbf{S}_m = C_1 \exp \left[ \alpha_4 \left( I_1 - 3 \right) \right] \left[ 2 I_3^{-1/3} \mathbf{1} - 2/3 I_1 \mathbf{C}^{-1} \right] \\
\mathbf{S}_f = 2 \frac{C_4}{\alpha_4} \{ \exp \left[ \alpha_4 \left( I_4 - 1 \right) \right] - 1 \} \mathbf{a}_u \otimes \mathbf{a}_u
\] (3.32) (3.33)

As regards the constitutive formulation adopted for the calcaneal fat pad the strain energy function is defined by the volumetric and volume-preserving terms, while the strain energy contribution of fibres is added when considering the skin.
3.5 REFERENCES


Definition of the numerical models of the heel region: finite element models and constitutive formulations


Biomechanical analysis of heel pad tissues

CMBM – Centre of Mechanics of Biological Materials – University of Padova, Italy


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CHAPTER FOUR

EVALUATION OF CONSTITUTIVE PARAMETERS OF THE HEEL PAD TISSUES

4.1 INTRODUCTION

Once a constitutive model is formulated, it is necessary to evaluate the constitutive parameters that characterise the mechanical behaviour of the heel pad tissues. This chapter describes a procedure for the evaluation of constitutive parameters using a combined experimental and computational approach.

The non-linearity of the mechanical behaviour of the soft tissues implies that constitutive parameters are evaluated by directly comparing analytical and numerical model results with experimental data. To define the discrepancy existing between experimental data and model results, a specific cost function is formulated. Constitutive parameters are evaluated by minimizing the cost function through a stochastic-deterministic procedure. With regard to the calcaneal fat pad tissue, the preliminary definition of constitutive parameters is developed using a visco-hyperelastic formulation, considering experimental data from in vitro compression tests on a specimen of calcaneal fat pad and data from in vivo tests to identify the actual trend of tissue stiffness. Further experimental tests are performed on the calcaneal fat pad of a cadaveric foot using in situ indentation tests. A numerical model is developed to interpret the experimental situation. The constitutive formulation is applied using different sets of parameters evaluated by a grid defined around the preliminary set. The comparison of data from in situ tests and numerical results leads to an optimal domain of parameters according to an admissible discrepancy criterion. The evaluation of constitutive parameters adopted for the skin follows the procedure previously described. The agreement between model predictions and experimental results determines the reliability of the constitutive characterisation.
4.2 IDENTIFICATION OF CONSTITUTIVE PARAMETERS

The definition of parameters that govern the constitutive equations of biological materials is a challenge that becomes difficult to face when the number of parameters is significant. Such a case occurs when anisotropic materials and refined nonlinear laws are considered (Grédiac et al., 2002; Araújo et al., 1996). The usual approach consists in the inverse analysis of constitutive models. An inverse analysis assumes that stress-strain history is given by experimental procedures and it attempts to estimate parameter values that would yield the best fit to the given stress-strain history (Lei and Szeri, 2006). Constitutive parameters are consequently evaluated using experimental data, corresponding model results and optimization techniques. Mechanical tests should be performed on geometrically simple specimens and appropriate boundary conditions should be adopted to generate the most homogeneous stress-strain fields possible. Indeed, simple experimental tests can be described using analytical formulations. For more complicated situations, numerical methods must be adopted. Furthermore, experimental data should explore several different deformation modes in order to provide the necessary information for the characterisation of the generic stress-strain behaviour of the tissue and the univocal definition of constitutive parameters (Natali et al., 2006).

4.2.1. Definition of the cost function

The procedure adopted for the definition of the constitutive parameters requires the minimization of the discrepancy between experimental and analytical or numerical model results through a specific cost function. The cost function depends on the accuracy of the input data (such as experimental data), the quantity of data at disposal and the weight that each data is associated with. There are several cost functions proposed in literature (Praagman et al., 2006; Cash et al., 2005; Stokes et al., 2001; Kyriacou and Davatzikos, 1998; Crowninshield and Brand, 1981). With regard to the biomechanical contest, the cost function used (Natali et al., 2009a, 2009b, 2010) is reported in equation 4.1 where the weight of each data in the output is related with the ratio between the experimental data and model results:
Evaluation of constitutive parameters of the heel pad tissues

\[ \Omega(\omega) = \frac{1}{n} \sum_{i=1}^{n} \left[ 2 - \frac{P_{ii}^{\text{mod}}(\omega, \lambda_i^{\exp})}{P_{ii}^{\exp}} - \frac{P_{ii}^{\exp}}{P_{ii}^{\text{mod}}(\omega, \lambda_i^{\exp})} \right]^2 \]  \hspace{1cm} (4.1)

where \( \omega \) is the set of constitutive parameters, \( n \) the number of experimental data, \( \lambda_i^{\exp} \) the \( i \)-th experimental input datum, \( P_{ii}^{\exp} \) the \( i \)-th experimental output value, and \( P_{ii}^{\text{mod}} \) the \( i \)-th analytical or numerical model output result corresponding to the constitutive parameters \( \omega \) and the experimental input \( \lambda_i^{\exp} \). The function \( \Omega \) is a measure of the overall difference between experimental and model results when constitutive parameters \( \omega \) are adopted. The optimization problem involves the evaluation of the set of constitutive parameters \( \omega_{\text{opt}} \) that minimizes \( \Omega \).

Some limitations on constitutive parameters may be necessary, including, for example, the imposition on the tendency of the hyperelastic strain energy function to increase strictly with strain. It may be difficult to define these conditions by boundaries the parameters domain and should be more easily implemented by penalty contributions to the cost function (Boukari and Fiacco, 1995):

\[ \Omega(\omega) = \frac{1}{n} \sum_{i=1}^{n} \left[ 2 - \frac{P_{ii}^{\text{mod}}(\omega, \lambda_i^{\exp})}{P_{ii}^{\exp}} - \frac{P_{ii}^{\exp}}{P_{ii}^{\text{mod}}(\omega, \lambda_i^{\exp})} \right]^2 + \frac{1}{n} \sum_{i=1}^{n} \Theta_i(\omega, \lambda_i^{\exp}, P_{ii}^{\exp}, P_{ii}^{\text{mod}}) \]  \hspace{1cm} (4.2)

where the penalty term \( \Theta_i \) assumes a reasonably high value when the model result \( P_{ii}^{\text{mod}} \) does not satisfy a specified criterion.

4.2.2. Implementation of a stochastic-deterministic procedure

If the adopted constitutive model is highly non-linear, the cost function is often characterised by multimodal behaviour (i.e. the function presents a global minimum and further local minima). Solving the optimization problem by deterministic methods (Stoer and Bulirsch, 1992) may result in the definition of only one of the local minima, without generating the optimal solution. On the other hand, a stochastic algorithm performs well in the presence of a very high number of variables. This is based on
random evaluations of the cost function, in such a way that transitions out of a local minimum are possible. However, this does not guarantee reaching the global/local minimum, but only moving close to the minimum itself. It becomes necessary, therefore, to perform the optimization using a new algorithm formulated by coupling a stochastic and deterministic method.

A specific simulated annealing procedure (Corana et al., 1987; Kirkpatrick et al., 1983) and the Nelder-Mead method (Begambre and Laier, 2009; Lagarias et al., 1998) can be adopted for this purpose. The new procedure explores all minima, evaluates the region where the global minimum is located and then returns to the exact position of the global minimum itself. More specifically, the computation begins from an initial set of constitutive parameters that is recorded in the vector $\omega_0$. According to the simulating annealing technique, new candidate points $\omega'$ are generated around the current point $\omega_i$ by applying random moves. A new point $\omega'$ is accepted or rejected according to the Metropolis criterion (Metropolis et al., 1953) leading to a new current position $\omega_{i+1}$. The best point reached corresponding to the set of constitutive parameters that mainly minimizes the cost function is recorded as $\omega_{opt}^*$. The solution is used as input to the Nelder Mead method that returns a new point $\omega_{opt}^{**}$, which represents a minimum (global or local) that is compared with the previous $\omega_{opt}^*$. The best of them is recorded as $\omega_{opt}$ and the stochastic-determinist procedure restarts until no further useful cost function improvements can be expected. The procedure returns to the set of constitutive parameters associated with the best solution.

4.2.3. Definition of the analytical models for uniaxial tests

The first step of the procedure is to develop analytical models to interpret the experimental tests, accounting for the constitutive formulation and the specific boundary conditions. With regard to the uniaxial tests (Miller-Young et al., 2002; Ledoux and Blevins, 2007), the analytical model has to provide a relationship between nominal stress along the loading direction (i.e. the ratio between the load applied and the initial cross sectional area of the specimen) and the imposed strain configuration. For a hyperelastic material, the general stress-strain relationship is evaluated as
Evaluation of constitutive parameters of the heel pad tissues

\[ \mathbf{P} = 2\mathbf{F} \frac{\partial W}{\partial \mathbf{C}} = \mathbf{FS}, \] where \( \mathbf{P} \) is the first Piola-Kirchhoff stress tensor, a measure of nominal stress, and \( \mathbf{F} \) is the deformation gradient.

The definition of the analytical models, taking into consideration the fiber-reinforced hyperelastic model for skin, has to provide a relationship between nominal stress along the loading direction and the imposed strain history. Considering the proposed formulation of the strain energy function, reported in Chapter 3, the general stress-strain relationship is evaluated as:

\[ \mathbf{P} = \mathbf{P}_{mv} + \mathbf{P}_{mi} + \mathbf{P}_f \] (4.3)
\[ \mathbf{P}_{mv} = 2\mathbf{F} \frac{\partial U}{\partial \mathbf{C}} = \left[ K_s / 2 + r (r + 1) \left[ 2J (J - 1) - rJ^{-r} + rJ \right] \mathbf{F}^{-T} \right] \] (4.4)
\[ \mathbf{P}_{mi} = 2\mathbf{F} \frac{\partial \tilde{W}}{\partial \mathbf{C}} = C_i \exp\left[ \alpha_i (\tilde{I}_i - 3) \right] \left[ 2J^{-2/3} \mathbf{F} - 2/3 \tilde{I}_i \mathbf{F}^{-T} \right] \] (4.5)
\[ \mathbf{P}_f = 2\mathbf{F} \frac{\partial W}{\partial \mathbf{C}} = 2 \left( C_f / \alpha_f \right) \left[ \exp\left[ \alpha_f (I_4 - 1) \right] - 1 \right] \mathbf{F} \left( \mathbf{a}_0 \otimes \mathbf{a}_0 \right) \] (4.6)

For the uniaxial loading condition the deformation gradient can be assumed to be a diagonal tensor with principal stretches \( \lambda_i, \lambda_j, \lambda_k \). The stretch along \( i \) is imposed, while the other two stretches can be measured by specific optical or strain-gauge systems.

\[ \mathbf{F} = \begin{bmatrix} \lambda_i & 0 & 0 \\ 0 & \lambda_j & 0 \\ 0 & 0 & \lambda_k \end{bmatrix} \] (4.7)

Accounting for the deformation gradient assumed and the orientation of fibers, the specific formulations of nominal stress components can be evaluated. With regard to the data from uniaxial tests, only the stretch along the loading direction is usually experimentally evaluated. On the other hand, the first Piola-Kirchhoff stress tensor defined by equation 4.3 depends on principal invariants \( \tilde{I}_i, I_3, I_4 \) that, in turn, are functions of all the principal stretches \( \lambda_i, \lambda_j, \lambda_k \). The stretch components that are not experimentally evaluated can be calculated using analytical methods.
With regard to the specific experimental situation the axial loading direction is indicated as $i$. The stress components along the remaining directions $j$ and $k$ are null because of the uniaxial configuration of experimental tests, as:

$$
P_{ij}(\lambda_i, \lambda_j, \lambda_k) = 0, \quad P_{ik}(\lambda_i, \lambda_j, \lambda_k) = 0$$

(4.8)

Accounting for the experimental value of stretch component $\lambda_i^\text{exp}$, the solution of the algebraic non-linear system 4.8 leads to stretch components $\lambda_j$, $\lambda_k$, making it possible to evaluate the nominal stress component $P_{ij}(\lambda_i^\text{exp}, \lambda_j, \lambda_k)$ along the loading direction.

In order to obtain a set of constitutive parameters capable of correctly interpreting the lateral behaviour of the tissue, it is necessary to introduce specific conditions on the Poisson ratios (Natali et al., 2007).

According to the large strain field assumed, tangent Poisson ratios are properly defined by the following formulation (Natali et al., 2009a):

$$
V_{ij} = -\frac{d \ln \lambda_j}{d \ln \lambda_i} = \frac{\lambda_i}{\lambda_j} \frac{d \lambda_i}{d \lambda_j} \quad l = j, k
$$

(4.9)

The evaluation of the tangent Poisson moduli $V_{ij}, V_{ik}$ for the experimental value of stretch component $\lambda_i^\text{exp}$ requires the knowledge of the corresponding stretches $\lambda_j$, $\lambda_k$ and their first derivatives. Given the non-linear algebraic system 4.8 and an experimental value of stretch component $\lambda_i^\text{exp}$, accounting for regularity of functions $P_{ij}(\lambda_i, \lambda_j, \lambda_k)$ and $P_{ik}(\lambda_i, \lambda_j, \lambda_k)$, the implicit function theorem (Krantz et al., 2002) states the existence of implicit functions $\lambda_j(\lambda_i)$ and $\lambda_k(\lambda_i)$ satisfying system 4.8 for values of $\lambda_i$ close to $\lambda_i^\text{exp}$. Furthermore, from the implicit function theorem it is possible to evaluate the derivatives of functions $\lambda_j(\lambda_i)$ and $\lambda_k(\lambda_i)$ as:

$$
\begin{bmatrix}
\frac{d \lambda_j}{d \lambda_i} \\
\frac{d \lambda_k}{d \lambda_i}
\end{bmatrix} = -\begin{bmatrix}
\frac{\partial P_{ij}}{\partial \lambda_j} & \frac{\partial P_{ij}}{\partial \lambda_k} \\
\frac{\partial P_{ik}}{\partial \lambda_j} & \frac{\partial P_{ik}}{\partial \lambda_k}
\end{bmatrix}^{-1} \begin{bmatrix}
\frac{\partial P_{ij}}{\partial \lambda_i} \\
\frac{\partial P_{ik}}{\partial \lambda_i}
\end{bmatrix}
$$

(4.10)
The values of stretch components \( \lambda_j, \lambda_k \) corresponding to the experimentally imposed stretch \( \lambda_i^{\text{exp}} \) are evaluated by solving the system 4.8, while equation 4.10 provides the derivatives. The tangent Poisson ratios \( \nu_{ij}, \nu_{ik} \) can, consequently, be evaluated by equation 4.9.

The analytical model for visco-hyperelastic formulation used for calcaneal fat pad has to provide a relationship between nominal stress along the loading direction and the imposed strain condition. The general stress-strain relationship is evaluated as:

\[
P_m = P_{mv} + P_{mi} \quad (4.11)
\]

\[
P_m \left( C, q' \right) = 2F \partial \psi_m / \partial C = P_m^\infty \left( C \right) + \sum_{i=1}^{n} \left( P'_m \left( C, q' \right) \right) \quad (4.12)
\]

\[
P_m^\infty = 2F \partial W_m^\infty / \partial C \quad (4.13)
\]

\[
P'_m = 2F \gamma \partial W_m^\infty / \partial C - F q' \quad (4.14)
\]

The analytical solution of differential equations of viscous variables \( q'_i \) is quite difficult and numerical integration is more suitable (Simo and Hughes, 1998). With regard to the uni-axial compression test along direction 1, the relationship between nominal stress component \( P_m \), strain and time can be formulated as:

\[
P_m \left( C, t \right) = P_m^\infty \left( C \right) + \sum_{i=1}^{n} \gamma_m h'_m \left( t \right) \quad (4.15)
\]

The computation of equilibrium stress is performed according to the procedure previously described for transversally isotropic materials. The evolution of viscous variables \( h'_m \) is computed by an incremental algorithm (Simo and Hughes, 1998):

\[
h'_m \left( t^{p+1} \right) = \exp \left\{ -\frac{t^{p+1} - t^p}{\tau'} \right\} h'_m \left( t^p \right) + \frac{1}{\gamma'} \exp \left\{ -\frac{t^{p+1} - t^p}{2\tau} \right\} \left[ P_m^\infty \left( t^{p+1} \right) - P_m^\infty \left( t^p \right) \right] \quad (4.16)
\]

where \( t^p \) and \( t^{p+1} \) are subsequent time steps.
4.3 EVALUATION OF CONSTITUTIVE PARAMETERS FOR CALCANEAL FAT PAD TISSUES

With the goal of ultimately developing a finite element model with realistic representations of the heel pad, the aim of the study presented is to first isolate the calcaneal fat pad tissue and then determine a constitutive model that represents its material behaviour. This involves experimental activities investigating the structure and micro-structure of the tissue and gathering data from in vitro, in situ and in vivo mechanical tests.

The visco-hyperelastic constitutive model is considered by taking into account the typical features of calcaneal fat pad tissue mechanical response, including large displacements and strains, almost incompressible behaviour, a non-linear stress-strain relationship, and time-dependent effects. A preliminary definition of constitutive parameters is developed based on experimental data from in vitro and in vivo tests reported in Chapter 2. These preliminary parameters are evaluated using an inverse analysis (Natali et al., 2010a) performed using an optimization algorithm that minimizes the discrepancy between experimental data and results from analytical models of the tests. Since it is easy to manage these analytical models, the analysis of these tests made it possible to offer a prompt evaluation of a preliminary set of constitutive parameters. Furthermore, analytical models can be used to rapidly evaluate the influence of each parameter on the constitutive model results. The preliminary set is able to interpret the general trend of the mechanical behaviour of the tissue and is adopted as a basis for the subsequent analysis of in situ tests, also involving a reduction in the range of the parameters to be investigated. In order to interpret the mechanical response of living heel pad tissues, which is very much influenced by the interaction and connection with surrounding tissues and structures (Aerts et al., 1996), constitutive parameters have to be updated by considering data from experimental investigations that account for the actual conditions of the tissue (Natali et al., 2011; Natali et al., 2012). Consequently, detailed data from indentation tests are investigated with regards to in situ conditions. Because of the complex configuration of experimental tests, the analysis is performed using numerical methods. Different sets of constitutive parameters are defined based on the preliminary set obtained, and adopted for the numerical analysis. By evaluating the discrepancy
between the numerical results and experimental data, it is possible to accurately compute a domain of parameters where discrepancy assumes minimal values, as a region of parameters that are capable of interpreting the mechanical response of the heel pad tissues.

4.3.1. Procedure for the evaluation of constitutive parameters using analytical models

The evaluation of constitutive parameters of calcaneal fat pad tissue is obtained considering the in vitro tests (Miller Young et al., 2002) reported in Chapter 2. Accounting for the visco-hyperelastic constitutive formulation proposed for calcaneal fat pad and the experimental boundary conditions under uni-axial compression, a specific analytical model is developed. The cost function is adopted to evaluate the discrepancy between model and experimental results, accounting for compression tests developed at different strain rates, 0.01, 0.1, 2100 and 4200 %/s, and a stress relaxation test developed at 40% strain. The different strain rates are assumed to characterize the time-dependent response over a wide spectrum of loading conditions, leading to an efficient definition of parameters. Within the cost function, strain and time are the input data while nominal stress values are the output data.

Differences in the mechanical properties of heel pad tissues with regard to in vitro and in vivo conditions have been found by many authors (Hsu et al., 2007; Miller-Young et al., 2002; Ledoux and Blevins, 2007). Zheng et al. (Zheng et al., 2000) analyzed the mechanical properties of the heel pad by performing indentation tests on healthy young people (21-24 years). A similar technique was adopted by Erdemir et al. (Erdemir et al.,2003) and Erdemir et al. (Erdemir et al., 2006)on healthy people with average ages of 47 and 52 years, respectively. A different approach was adopted by Gefen et al. (Gefen et al., 2001). In this approach the mechanical properties of the heel pad were investigated by measuring the heel pad tissue deformation and the heel-ground contact pressure during the stance phase of the gait. The analysis of available data makes it possible to obtain information about the in vivo relationship between tissue shear stiffness and strain rate, which can be evaluated by the cost function. With regard to a pure shear stress configuration, the visco-hyperelastic relationship between
stress and strain history can be re-written according to the following formulation (Simo and Hughes, 1998):

\[ \tau(\gamma,t) = \int_0^t g(t-s) \frac{d\tau^0}{ds} ds \]  

(4.17)

where \( \tau \) and \( \gamma \) are shear stress and strain, respectively, \( \tau^0 \) specifies the instantaneous shear response of the material, as the material response when strain rate approaches infinitum, while \( g \) is a standard relaxation function, calculated as:

\[ g(t) = \gamma^\omega + \sum_{i=1}^n \gamma^i \exp\left[-t/\tau^i\right] \]  

(4.18)

According to the proposed hyperelastic formulation, the instantaneous stress can be computed as:

\[ \tau^0(\gamma) = 2C^0_1 \gamma \exp[\alpha_I \gamma] \]  

(4.19)

where \( C^0_1 \) is an instantaneous hyperelastic parameter, evaluated as \( C^0_1 = C_1/\gamma^\omega \).

The aim pertains to the evaluation of the initial shear modulus, as the tissue shear stiffness corresponding to small strain values. The previous relationship can be linearized by a Taylor series for small shear strain values:

\[ \tau^0_{\gamma \to 0} = 2C^0_1 \gamma \]  

(4.20)

During a constant strain rate test, shear strain linearly increases with time according to strain rate \( k \):

\[ \gamma(t) = kt \]  

(4.21)

Accounting for equations (4.18), (4.20) and (4.21), the convolution integral (4.17) leads to the following formulation (Natali et al., 2008a):
\[
\tau_{\gamma \to 0} = \gamma^0 2C_1 \gamma + \sum_{i=1}^{n} \gamma' 2C_1^0 k \tau' \left\{ 1 - \exp\left[ -\gamma / k \tau' \right] \right\}
\]

(4.22)

According to standard solid mechanics definitions, the initial shear modulus can be evaluated as:

\[
G = \frac{d\tau}{d\gamma}_{\gamma \to 0} = \gamma^0 2C_1 + \sum_{i=1}^{n} \gamma' 2C_1^0 \exp\left[ -\gamma / k \tau' \right]
\]

(4.23)

The average value of shear stiffness for strain rate \( k \) can be defined by applying the integral mean value theorem over the shear strain range \([0, \gamma_L]\), where \( \gamma_L \) is the strain limit of almost-linear shear response, leading to the following relationship:

\[
\bar{G}(k) = \gamma^0 2C_1 + \frac{1}{\gamma_L} \sum_{i=1}^{n} \gamma' 2C_1^0 k \tau' \left\{ 1 - \exp\left[ -\gamma_L / k \tau' \right] \right\}
\]

(4.24)

4.3.1.1. Results from analytical model

The optimization procedure uses in vitro experimental results (Miller-Young et al., 2002) which lead to the constitutive parameters reported in Tables 4.1 and 4.2, with regard to hyperelastic and viscous response, respectively (Natali et al., 2010a).

The hyperelastic parameters entail the equilibrium initial shear stiffness \( G^- \) to be 9.30 \( 10^{-3} \)MPa. The shear stiffness parameter is here assumed to be a reference for evaluating the properties of tissue stiffness in the light of their relationship with the hyperelastic parameters, as \( G^- = 2C_1 \). The relative stiffness at equilibrium, defined as \( \gamma^- = 1 - \gamma' - \gamma'' - \gamma' \), is about 9.96 \( 10^2 \). It is associated with the mechanical response of the material when the strain rate approaches zero (McCrum et al., 1997). The instantaneous initial shear stiffness, i.e., the shear stiffness when strain rate approaches infinitum, can be evaluated as \( G^0 = G^- / \gamma^- \). Considering a time-dependent response, the initial shear stiffness ranges between 9.30 \( 10^{-3} \) and 9.33 \( 10^{-2} \) MPa for strain rates included between zero and infinitum, respectively. The achieved values are in agreement with the data reported in literature (Spears et al., 2005).
Chapter 4

Biomechanical analysis of heel pad tissues

<table>
<thead>
<tr>
<th>$K_v$ (MPa)</th>
<th>$r$</th>
<th>$C_1$ (MPa)</th>
<th>$a_1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$4.73 \cdot 10^{-2}$</td>
<td>$1.06 \cdot 10^1$</td>
<td>$4.65 \cdot 10^{-3}$</td>
<td>$1.19 \cdot 10^0$</td>
</tr>
</tbody>
</table>

*Table 4.1. Hyperelastic parameters for calcaneal fat pad tissue*

<table>
<thead>
<tr>
<th>$\gamma_1$</th>
<th>$\tau_1$ (s)</th>
<th>$\gamma_2$</th>
<th>$\tau_2$ (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$8.95 \cdot 10^{-1}$</td>
<td>$6.23 \cdot 10^{-4}$</td>
<td>$4.29 \cdot 10^{-5}$</td>
<td>$1.55 \cdot 10^{-2}$</td>
</tr>
<tr>
<td>$\gamma_3$</td>
<td>$\tau_3$ (s)</td>
<td>$\gamma_4$</td>
<td>$\tau_4$ (s)</td>
</tr>
<tr>
<td>$2.76 \cdot 10^{-3}$</td>
<td>$9.88 \cdot 10^4$</td>
<td>$3.12 \cdot 10^{-3}$</td>
<td>$9.82 \cdot 10^5$</td>
</tr>
</tbody>
</table>

*Table 4.2. Viscous parameters for calcaneal fat pad tissue*

Both experimental (empty circles) and model (continuous lines) results are reported in Figures 4.1 and 4.2, corresponding to constant strain rate tests developed for different strain rates, as 0.1 (Figure 4.1(a)), 0.01 (Figure 4.1(b)), 2100 (Figure 4.2(a)) and 4200 %/s (Figure 4.2(b)).

![Figure 4.1](image)

*Figure 4.1. Comparison of experimental data (empty dots) and model results (continuous lines) for unconfined compression tests on tissue samples at different strain rates: 0.01%/sec (a) and 0.1%/sec (b). The value reported for $\Omega$ represents the discrepancy between experimental data and model results reported as the value assumed by the cost function.*
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Figure 4.2. Comparison of experimental data (empty dots) and model results (continuous lines) for unconfined compression tests on tissue samples at different strain rates: 2100%/sec (a) and 4200%/sec (b). The value reported for $\Omega$ represents the discrepancy between experimental data and model results reported as the value assumed by the cost function.

Figure 4.3. Comparison of experimental data (empty dots) and model results (continuous lines) for stress relaxation tests (40% compression strain): nominal stress vs. time (a) and normalized stress vs. time (b). The value reported for $\Omega$ represents the discrepancy between experimental data and model results reported as the value assumed by the cost function.

The quality of the fit is reported as the value assumed by the cost function. The results from stress relaxation test are reported in Figure 4.3(a), while in Figure 4.3(b) the same results are proposed according to normalized stress values. The agreement between experimental data and model results is good for both constant strain rate and stress relaxation tests.

The proposed constitutive formulation together with the constitutive parameters is adopted to analyze further experimental data from relaxation tests performed by Ledoux and Blevins (Ledoux and Blevins, 2007). In particular, data on specimens from the subcalcaneal region, like the heel pad, are compared with results from the proposed constitutive formulation, as reported in Figure 4.4. The experimental
magnitude of relaxation phenomena is quite lower than model predictions (indicated by term \( b \)) while the experimental instantaneous stress is quite greater than model values (indicated by term \( a \)). The different response is interpreted as a consequence of the age of the donors. The data from Ledoux and Blevins (Ledoux and Blevins, 2007) is significantly lower than the data adopted for the constitutive parameters evaluation (Miller-Young et al., 2002). In fact, different authors reported the influence of aging phenomena on tissue stiffness properties (Kinoshita et al., 1996; Hsu et al., 1998; Natali et al., 2008b), showing results in agreement with the interpretation of the analysed experimental data.

![Figure 4.4. Experimental data (empty dots) from a stress relaxation test developed by Ledoux et al. (2007), together with a model prediction (continuous line). Terms \( a \) and \( b \) evaluate the discrepancy between experimental and model results with regard to instantaneous and equilibrium response, respectively. The value reported for \( \Omega \) represents the discrepancy between experimental data and model results reported as the value assumed by the cost function](image)

To interpret the \textit{in vivo} response, considering the relationship between tissue stiffness and strain rate previously reported together with specific experimental data, it is necessary to properly rearrange the constitutive parameters. Specifically, the hyperelastic parameters \( K_v \), \( C_1 \) and viscous relative stiffness are properly modified, as reported in Table 4.3.

<table>
<thead>
<tr>
<th>( K_v ) (MPa)</th>
<th>( C_1 ) (MPa)</th>
<th>( \gamma_1 )</th>
<th>( \gamma_2 )</th>
<th>( \gamma_3 )</th>
<th>( \gamma_4 )</th>
</tr>
</thead>
<tbody>
<tr>
<td>( 1.25 \cdot 10^1 )</td>
<td>( 1.23 \cdot 10^2 )</td>
<td>( 7.17 \cdot 10^1 )</td>
<td>( 1.55 \cdot 10^4 )</td>
<td>( 6.52 \cdot 10^2 )</td>
<td>( 6.26 \cdot 10^2 )</td>
</tr>
</tbody>
</table>

Table 4.3. Constitutive parameters adopted for the analysis of data from \textit{in vivo} tests
The set of constitutive parameters obtained is defined as the preliminary set $\omega^p$ (Table 4.4).

<table>
<thead>
<tr>
<th>$K_v$(MPa)</th>
<th>$r$</th>
<th>$C_1$(MPa)</th>
<th>$\alpha_1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$1.25 \cdot 10^{-1}$</td>
<td>$1.06 \cdot 10^1$</td>
<td>$1.23 \cdot 10^{-2}$</td>
<td>$1.19 \cdot 10^0$</td>
</tr>
</tbody>
</table>

Table 4.4. Preliminary set of constitutive parameters $\omega^p$

The modification of hyperelastic and viscous parameters makes it possible to account for the influence of in vivo tissue configuration on equilibrium and the time-dependent response of the material, respectively. The results show the ability of the model to interpret the mechanical response of the tissue with regard to the different experimental situations considered. Specifically, in Figure 4.5 the experimental relationship between initial shear stiffness and strain rate is represented by empty dots, while model results are reported by a continuous line.

![Figure 4.5](image)

Figure 4.5. Influence of strain rate on initial shear modulus for heel pad tissue: model results (continuous line) and data from in vivo experimental tests (empty dots). Experimental data are taken from different authors: (1) Zheng et al. (2000), (2) Erdemir, et al. (2006), (3) Erdemir et al. (2003), (4) Gefen et al. (2001). The value reported for $\Omega$ represents the discrepancy between experimental data and model results reported as the value assumed by the cost function.

4.3.2. Procedure for evaluating constitutive parameters through an analysis of in situ tests

The preliminary set of constitutive parameters $\omega^p$ is adopted as a reference point for the analysis of in situ tests reported in Chapter 2 (Miller-Young, 2003), performed using numerical methods (Nataliet al., 2011). The numerical models of the heel region...
are obtained by the discretization of solid models. The solid model of the heel pad structure is developed by starting from CT images and histological and morphometric measurements of the specific foot. The virtual solid model of the skeletal components and plantar tissues is reported in Figure 4.6(a). The numerical model is developed for the region around the indentation area of the foot where mechanical tests are performed (Fig. 4.6(b)).

The proposed visco-hyperelastic constitutive formulation is adopted to define the mechanical behaviour of the heel pad while a linear elastic model is used for the bone. To analyze indentation tests, the superior surface of the calcaneus is fully fixed, while a displacement field is imposed over a region of the heel pad corresponding to the real interaction zone of the indenter, which considers the different strain rates investigated. The applied boundary and loading conditions correspond well to the actual experimental protocol and results. With regard to the numerical strategy adopted for the analysis of indentation, the assumption of an equivalent displacement field is acceptable because of the high stiffness of the indenter. Furthermore, the assumption makes it possible to avoid heavy computational efforts of numerical simulation of interaction processes by contact algorithms.

The visco-hyperelastic constitutive model is implemented in the general purpose finite element software ABAQUS 6.8 (Simulia, Dassault Systèmes, Providence, RI, USA) by developing an ad hoc routine able to interpret the behaviour of heel pad tissues in consideration of the specific formulation described.

Numerical analyses are performed when accounting for different sets of parameters that are evaluated by defining a grid around the preliminary set $\omega^p$. The grid is defined...
according to a variational process, by multipliers $m_1$ and $m_2$ of the groups of parameters, where $K_i = m_1 K_i^p$, $C_i = m_1 C_i^p$ refers to initial stiffness parameters and $r = m_2 r^p$, $\alpha = m_2 \alpha^p$ to non linearity parameters. The values of the multipliers are determined by accounting for the conformation of the function in terms of its variation within the associated interval relating to the optimization region (Figure 4.7).

Figure 4.7. Different sets of constitutive parameters are obtained from the preliminary set

For the low and high strain rate tests, the numerical results and experimental data (Miller-Young, 2003) are evaluated using the previously reported cost function. The imposed displacement and time are the input data and indentation load values are the output data. The procedure provides an evaluation of the discrepancy between experimental data and model results for the different sets of parameters investigated (Figure 4.8).

Figure 4.8. Evaluation of the discrepancy between experimental data and numerical results for a set of constitutive parameters

With regard to the different tests investigated, the relationship between the discrepancy and constitutive parameters is interpolated using a two-dimensional spline function.
Figure 4.9. Representation of the relationship of experimental-numerical results discrepancy vs. constitutive parameters. Conformation of the cost function, as interpolated by spline functions vs. groups of constitutive parameters, adopting multipliers of initial stiffness $m_1$ and non-linearity $m_2$ parameters: representation for low strain rate test (a) and high strain rate test (b). Subdivision of the domain of constitutive parameters, represented by multipliers of initial stiffness $m_1$ and non-linearity $m_2$ parameters, by iso-discrepancy curves: representation for low strain rate (c) and high strain rate (d) tests.

A graphical representation is reported in Figure 4.9 as an example of the cost function depending on initial stiffness parameters $K_v$ and $C_1$, and non-linearity parameters $r$ and $\alpha$. Figures 4.9(a) and 4.9(b) report the shape of the cost function for low and high strain rate tests, respectively. A continuous definition of relationship between discrepancy and parameters makes it possible to accurately compute the region where discrepancy assumes minimal values. Figures 4.9(c) and 4.9(d) report subdivisions of the parameter domain dependent on the values the discrepancy assumes, as regards low and high strain rate tests, respectively. The graphs highlight the definition of regions where discrepancy assumes the lowest values. By intersecting such regions, it is possible to evaluate the optimal domain of constitutive parameters values that can be used to interpret the mechanical response of living heel pad tissues.
4.3.3. Numerical results

The results from numerical tests are reported subsequently. Figure 4.10(a) reports the superposition of regions of the domain of constitutive parameters, where discrepancy, evaluated for the different tests assumes the lowest values. Three sets of parameters are assumed and their numerical results are compared with experimental data in Figures 4.10(b) and 4.10(c) for low and high strain rate tests, respectively.

Figure 4.10. Comparison of experimental and numerical results from indentation tests: (a) superposition of minimum discrepancy regions for low and high strain rate tests and identification of a set of parameters, results from (b) low and (c) high strain rate indentation tests. Numerical results are computed for the three different sets of constitutive parameters within the optimal region.

The results reported relate to sets within the optimal domain that proves it can contain values capable of interpreting the problem, in spite of the variation within the domain.
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itself, showing different levels of acceptable approximation. As an example, the optimal set of hyperelastic parameters, indicated as \( \bm{\omega}^1 \), is reported in Table 4.5.

<table>
<thead>
<tr>
<th>( K_v ) (MPa)</th>
<th>( r )</th>
<th>( C_1 ) (MPa)</th>
<th>( \alpha )</th>
</tr>
</thead>
<tbody>
<tr>
<td>( 4.07 \times 10^{-2} )</td>
<td>( 2.32 \times 10^1 )</td>
<td>( 4.00 \times 10^{-3} )</td>
<td>( 2.63 \times 10^0 )</td>
</tr>
</tbody>
</table>

Table 4.5. Hyperelastic parameters from set \( \bm{\omega}^1 \), resulting from the optimization procedure performed on the basis of data from in situ tests

The results from the numerical analyses that use the set of constitutive parameters chosen are reported in Figures 4.11 and 4.12. In Figure 4.11 the deformed configuration of heel pad structure after indentation (Fig. 4.11(a)) is reported together with contours of vertical displacement (Fig. 4.11(b)), as displacement along the indentation direction, and minimum principal stretch fields (Fig. 4.11(c)).

Figure 4.11. Results from the numerical analysis of indentation tests: (a) deformed configuration of heel pad structure because of indentation contours of (b) vertical displacement, as displacement along indentation direction and (c) minimum principal stretch fields. Contours are reported over a transversal section of heel pad structure, as indicated in (a) by section plane A-A

The contours are reported over a transverse section of heel pad, corresponding to section plane A-A and relate to the configuration achieved when a 5 mm depth indentation is imposed. In Figure 4.12 minimum principal stress fields are reported,
which refer to analyses developed at low (Fig. 4.12(a)) and high (Fig. 4.12(b)) strain rates.

![Figure 4.12. Contours of minimum principal stress for indentation tests developed at (a) low and (b) high strain rates](image)

**4.3 EVALUATION OF CONSTITUTIVE PARAMETERS FOR HEEL SKIN**

Results from mechanical tests reported in Chapter 2 show the typical features of skin tissue mechanics, such as anisotropic response, almost incompressible behaviour, material and geometrical non linearity. Anisotropic behaviour is mainly determined by the distribution of collagen fibres, i.e. the cleavage lines (Langer 1978a, 1978b; Andermahr et al. 2007, Jor et al. 2007, Lokshin and Lanir 2009). According to experimental and histological evidence, a fiber-reinforced hyperelastic model is adopted to interpret the mechanical behaviour of heel skin tissues, which accounts for different contributions from isotropic ground matrix and fibre family. This assumes a transversally isotropic material scheme (Natali et al., 2012).

Experimental data are essential for the subsequent step where the definition of constitutive parameters of heel skin is addressed, especially with regard to the actual mechanical response of the heel region; this considers data from *in situ* indentation tests performed by Miller-Young (Miller-Young 2003) on the intact heel pad. According to the geometrical complexity of the test specimen a numerical model is provided by introducing skin tissues within the previously described model of the
calcaneal fat pad region. The procedure adopted evaluates the discrepancy between experimental and numerical results for different sets of constitutive parameters in order to arrive at a determination of the optimal domain as a region of minimal discrepancy between experimental and model results. Within the optimal domains obtained through the specific procedure adopted, it is possible to access the sets of constitutive parameters, as reported later, for calcaneal fat pad and heel skin, respectively.

4.4.1. Procedure for the evaluation of constitutive parameters

The steps required to evaluate constitutive parameters follow the procedure adopted to identify the constitutive parameters of calcaneal fat pad. A preliminary evaluation of constitutive parameters takes into account experimental tests on pig skin specimens, according to data from literature (Ankersen et al. 1999; Liu and Yeung, 2008; Xu et al. 2009; Zhou et al. 2010).

The mechanical response of pig skin underlines the anisotropy of skin tissues due to a preferential orientation of fibrous elements embedded within the isotropic ground matrix of the dermis. The fibro-reinforced hyperelastic model is considered and the constitutive parameters are evaluated by a procedure that follows the one adopted in the previous computation. Due to the simple geometry of specimens and the uni-axial mechanical tests, analytical models of the constitutive formulation are adopted in order to interpret the experimental tests (Natali et al., 2012).

Figure 4.13. Comparison of experimental and model results: (a) from tensile tests on pig skin specimens longitudinal and transversal to spine directions and (b) from compression tests on cylindrical pig skin specimens
In Figure 4.13, model results are reported with experimental data from tests performed on pig specimens. Model results (continuous lines) are compared with experimental data from tensile tests along (black empty circles) and across (grey empty circles) skin fibres and from compression tests (grey empty circles) showing the capability of the model to interpret the mechanical properties of skin tissues, thereby accounting for non-linearity and anisotropic behaviour.

The preliminary set of constitutive parameters obtained from \textit{in vitro} tests on pig skin tissues are updated using a grid which multiplies the group of parameters by different factors, such as initial stiffness parameters and non-linear parameters. Numerical results and experimental data from \textit{in situ} tests are compared with the previously reported cost function. The procedure provides the determination of an optimal domain as a region of minimal discrepancy between experimental and model results. Within such an optimal domain, it is possible to access the sets of constitutive parameters for heel skin.

\subsection*{4.4.2. Distribution of skin fibres on the foot}

A specific procedure is developed in order to assign the orientation of the fibres of the foot skin on the numerical model. Different splines are created on the solid model of the foot corresponding to the Langer’s lines of the foot, as reported in the literature (Andermahr et al., 2007). Each spline has been interpolated with a poly-line. To determine the orientation of the fibres, the tangent vector of each poly-line is evaluated at different points on the poly-line, with a distance of 5.0 mm from each other.

A \texttt{.txt} file is created with six columns and rows, the same amount of rows as the number of points analyzed. The first three columns report the coordinates of each point analyzed, while the latter three columns report the coordinates of the correspondent tangent vector.

An ad hoc routine has been developed to assign to each Gauss point of the numerical model the correct orientation of the fiber direction, considering the Langer’s lines of the foot. To obtain this, the orientation of the fibre in each Gauss point is calculated with a mean weight between the orientation of the nearest three points known reported in the \texttt{.txt} file. The procedure developed is able to assign the specific orientation of
each point of the numerical model from the `.txt` file and to interpret the behaviour of skin in view of the specific fibro-hyperelastic formulation described.

Figure 4.14 reports the fiber orientation in the numerical model of the skin obtained with the procedure described.

![Figure 4.14. Langer’s lines reported in the numerical model of the foot with regard to lateral (a) and medial (b) view. Definition of the distribution of Langer’s lines in the heel skin in the solid (c) and numerical (d) model.](image)

4.4.3. **Numerical results**

Parameters are updated by analysing data from *in situ* tests. Three sets of constitutive parameters (set a, set b and set c) of the heel skin are chosen within the optimal domain of skin tissues achieved (Figure 4.15(a)). Numerical results obtained by these sets of parameters are reported together with experimental data from Miller-Young (2003) on an intact cadaveric foot heel pad, in Figure 4.15(b) and Figure 4.15(c) for low and high strain rate tests, respectively.
Figure 4.15. Comparison of experimental and numerical results from indentation tests: (a) numerical model of the heel region, results from low (b) and high (c) strain rate indentation tests. Numerical results are computed for the three different sets of constitutive parameters of the skin

4.4 VALIDATION OF CONSTITUTIVE MODELS

With the aim of evaluating the real response of the heel pad region, further experimental tests (Chapter 2) need considering to validate the overall procedure performed by the constitutive model and the procedure for the constitutive parameter identification. Experimental tests on the rearfoot reported by Sirimamilla (2009) are considered by performing a solid model of the rearfoot region starting from the CT images of a cadaveric foot. The numerical model of the rearfoot is meshed with more
than seven hundred thousand four-noded tetrahedral elements and is reported in Figure 4.16(b) and Figure 4.16(d).

![Figure 4.16. Comparison of experimental and numerical results: results from (a) indentation tests and (c) elevated platform tests from Sirimamilla (2009). Numerical results are computed for the three different sets of constitutive parameters of heel pad tissues. Models of the calcaneal region of a foot pointing out the bony components: (b) numerical model of the heel region and the spherical indenter, (d) numerical model of the heel region and the elevated platform.](image)

Models of the indenter and elevated platform are developed by taking in account experimental conditions. Bones are described using an orthotropic linear elastic model, while cartilages, muscle, ligaments and plantar fascia are described using hyperelastic formulations (Chen et al., 1998; Goske et al., 2006; Gu et al., 2010; Natali et al., 2010b). The superior surfaces of the astragal and the calcaneus are fully fixed and the indenter or the platform is moved upward to the plantar foot to correspond to the configuration of the experimental tests.

The skin-indenter interface was modelled using a contact surface with a friction coefficient of 0.42 according to reference in the literature (Zhang and Mak, 1999; Elkhyat et al., 2004).
Numerical results are compared with experimental data for indentation tests (Figure 4.16(a)) and platform tests (Figure 4.16(c)) using different sets of constitutive parameters within the optimal domains. Results show the capability of the procedure to interpret the mechanical response of heel pad tissues, while considering different feet and boundary conditions.

Figure 4.17. Results from the numerical analysis: contours of vertical displacement as displacement along indentation direction of indentation tests (a) and elevated platform tests (b). Contours are reported over a transversal section of the rear foot structure.

In Figure 4.17, deformed configurations of the rear foot structure after indentation tests (Figure 4.17(a)) and elevated platform tests (Figure 4.17(b)) proposed by Sirimamilla (2009) are reported together with contours of vertical displacement as displacement along the experimental loading direction.

The minimum principal stretch fields are also represented for indentation tests (Figure 4.18(a)) and elevated platform tests (Figure 4.18(b)). All contours are reported over a transverse section of the rearfoot and refer to the configuration achieved when a 4 mm depth indentation is imposed.
In order to evaluate the mechanical response of rear foot *in vivo* conditions, further numerical analyses are performed, considering the *in vivo* experimental setup proposed by Tong et al. (2003). Carrying out the same procedure adopted for the analysis of data from Sirimamilla (2009), different constitutive parameters for heel pad tissues are considered. In Figure 4.19, the experimental data (empty circles), such as those reported in literature, and numerical results for three sets of constitutive parameters (continues lines) are reported.

**Figure 4.18.** Results from the numerical analysis: contours of minimum principal stretch fields of indentation tests (a) and elevated platform tests (b). Contours are reported over a transversal section of the rear foot structure.

**Figure 4.19.** Comparison of numerical results and experimental data from Tong et al. (2003), using three different sets of constitutive parameters for heel pad tissues.
4.5 REFERENCES


Evaluation of constitutive parameters of the heel pad tissues


Sirimamilla AS. 2009. Elaborate experimentation for mechanical characterization of human foot using inverse finite element analysis. Doctoral Dissertation, Department of Mechanical Engineering, Case Western reserve University, Cleveland.


Evaluation of constitutive parameters of the heel pad tissues


CHAPTER FIVE

NUMERICAL ANALYSES OF HEEL PAD TISSUES

5.1 INTRODUCTION

The experimental approach can be investigated and integrated with a computational model of the heel pad. Such a model would allow a better understanding of the stress-strain relationship of the tissues in order to evaluate phenomena that are not accessible with sufficient accuracy and extension by means of experimental tests. Finite element modeling of the soft tissues of the foot would pave the way for understanding stress-related injuries (e.g. plantar fasciitis and diabetic ulceration), as well as improve orthotics and footwear design, considering the stress induced in the plantar region (e.g. distribution of the plantar pressure).

Numerical analysis are performed considering in vivo tests on a subject-specific heel pad. A numerical model of the heel region was developed adopting the solid model from MR images. The calcaneal fat pad tissue was described with the previously provided visco-hyperelastic model, while the previously described fiber-reinforced hyperelastic model was adopted for the skin. Different numerical analyses were performed to evaluate the mechanical response of heel pad tissues, considering the specific conditions of experimental tests.

Subsequently, a combined experimental and numerical approach is used to investigate the interaction phenomena occurring between the foot and footwear during the heel strike phase of the gait. Two force platforms are used to monitor the ground reaction force of a subject during bare and shod walking. The reaction forces obtained from experimental analyses define the loading conditions for different numerical analyses. Three dimensional numerical models of the heel region and of a running shoe are developed to evaluate the influence of different material characteristics and shoe conformations on the mechanical response of the heel pad region.
5.2 INVESTIGATION OF LOAD-DEFORMATION CURVES OF A HEALTHY HUMAN HEEL PAD

The heel pad exhibits non-linear visco-elastic characteristics, as do the majority of soft biological tissues. Due to the visco-elastic nature of the heel pad and its ability to deform under loading, a load-deformation curve is obtained showing a characteristic hysteresis curve when a loading/unloading cycle is applied. The mechanical behaviour of the heel pad is determined by the mechanical response of the adipose tissues and skin and by the interaction phenomena between the two tissues. Under compression loading, the heel pad has initially low stiffness, then the collagen fibre of the fat pad and skin come under tension, limiting the deformation of the tissue and resulting in an increase in stiffness. A way to perform a loading/unloading cycle and determine the mechanical behaviour of heel pad is to use a mechanical compression device (Fontanella et al., 2012a).

A three-dimensional quantitative subject-specific heel pad model, composed of fat pad and skin, is built on the basis of MRI data and other techniques for geometrical identification. The mechanical properties of the tissues are deduced by experimental data previously developed. The load-displacement curves obtained from the numerical analysis representing the mechanical behaviour of the model when subjected to an external compression are compared with those obtained by experimental activity using a compression device on an in vivo healthy heel pad.

5.2.1. Numerical modeling

A 3D model of the left heel pad is built on the basis of MRI scan data [3T Siemens Magnetom Trio, Fat-suppressed 3D dual echo steady state (DESS) sequence with (0.7 mm) isotropic resolution, matrix 320x576x104. TE/TR=5.5/13 ms, flip angle 25 degrees, TA=5 min]. The DICOM images are processed with a medical imaging and editing software in order to obtain the solid model by using density segmentation techniques. The solid model of the heel region, composed of calcaneus, soft tissues, fat pad and skin, is meshed by adopting almost four hundred thousand tetrahedral elements.
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A linear elastic model is adopted to describe the mechanical properties of the bone (Natali et al., 2010a), while soft tissues are described using a hyperelastic formulation as reported in the literature (Goske et al., 2006). A specific visco-hyperelastic constitutive model is adopted to describe the typical stress-strain behaviour of the fat pad tissues (Natali et al., 2010b, 2011). A fiber-reinforced hyperelastic model is considered to describe the heel skin tissue (Natali et al., 2012). The set of constitutive parameters adopted for calcaneal fat pad are reported in Tables 5.1 and 5.2 for hyperelastic and viscous parameters, respectively. Table 5.3 shows the hyperelastic constitutive parameters adopted for describing the mechanical behaviour of heel skin.

<table>
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<th>$C_i$(MPa)</th>
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Table 5.1. Hyperelastic constitutive parameters adopted for calcaneal fat pad tissue

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Table 5.2. Viscous constitutive parameters adopted for calcaneal fat pad tissue

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<th>$C_i$(MPa)</th>
<th>$\alpha_i$</th>
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<td>6.47 · 10^{0}</td>
<td>5.48 · 10^{0}</td>
</tr>
</tbody>
</table>

Table 5.3. Hyperelastic constitutive parameters adopted for the skin

The numerical analyses are developed in order to interpret the experimental tests previously described. A numerical model of the piston (40mm in diameter) is positioned under the heel skin corresponding to the area under investigation (Figure 5.1).
The superior surface of the calcaneus is fully fixed and the piston is moved upward to the plantar skin in order to mimic the configuration of the experimental tests. The contact surface between heel skin and indenter is modelled with a friction coefficient of 0.42. Five loading-unloading cycles with a one-minute break in between each are computed at the strain rates of 0.80 mm/s and 1.96 mm/s, in order to match the conditions of the experimental tests. Subsequently, a numerical analysis is developed in order to investigate stress relaxation phenomena within the heel tissues, according to the performed experimental conditions. In order to match the experimental conditions, the numerical analysis is computed using the same strain rate of 1.73 mm/s. Once the piston reaches the displacement of 9 mm, it remains in contact with the heel pad for 40 s, and then the decompression starts.

5.2.2. Numerical results

Figure 3 shows the experimental load-deformation curves in the hysteretic process of the heel pad when subject to loading, compared with the curves obtained from the numerical analysis. Specifically, hysteresis curves obtained by applying five loading-unloading cycles on the heel pad with a velocity of 0.8 mm/s (Figure 5.2(a), grey lines) and 1.96 mm/s (Figure 5.2(b), grey lines), and one-minute-break per cycle, are compared to those of the computational model (Figure 5.2(a) and 5.2(b), black lines). The constitutive properties adopted in the computational model are able to interpret the mechanical response of the tissues as regards the non-linear response and the time dependent behaviour of the heel pad. The formulations adopted enable the model to
interpret the typical hysteretic curve of the heel pad under loading/unloading conditions. Experimental compression tests and a computational simulation follow the same trend, highlighting the reliability of the computational model developed.

![Graph](image)

**Figure 5.2.** Compression tests with recovery of 1 minute-break between each cycle. Comparison of experimental (grey lines) and model (black lines) results for five cycles of loading and unloading compression tests with cylindrical piston: 0.80 mm/s strain rate tests (a) and 1.96 mm/s strain rate tests (b)

Deformed configurations of the heel region after the compression are reported with regard to the minimum principal stretch (Figure 5.3(a)) and stress (Figure 5.3(b)) fields for the strain rates of 1.96 mm/s. All contours are reported over a transverse section of the fat pad, reported in Figure (5.1(a)), and refer to the configuration achieved when a compression of 8.2 mm depth is imposed.

The numerical model developed in this study can be used for evaluating loading conditions which are not possible with experimental tests. For example, a small recovery time between each trial or different displacements and forces imposed can be considered by allowing for a more detailed evaluation of the mechanical behaviour of heel pad tissues under loading. Moreover, the numerical results make it possible to
investigate the mechanical response of the calcaneal fat pad after a cycle of loading/unloading.

![Figure 5.3. Results from the numerical analysis of compression tests on section AA after the first loading: contours of the minimum principal stretch field at 1.96 mm/s strain rate (a) and of minimum principal stress fields at 1.96 mm/s strain rate (b)](image)

Figure 5.4 shows the deformed configuration of fat pad after the first loading-unloading cycle. The minimum principal stretch fields for the analysis with velocity of 0.8 mm/s (Figure 5.4(a)) and 1.96 mm/s (Figure 5.4(b)) are compared. These figures allow the evaluation of residual deformation after the loading condition. The numerical results are different depending on the velocity adopted, leading to the evaluation of the effect of the strain rates on the mechanical response of the heel pad.

![Figure 5.4. Results from the numerical analysis of compression tests at a (a) 0.80 and (b) 1.96 mm/s strain rate. Contours of minimum principal stretch in the deformed configuration of heel pad structure after the first cycle of loading-unloading. Contours are reported over a transversal section AA of heel pad structure](image)
Figure 5.5 shows the deformed configurations of the heel region considering the loading-unloading tests at the strain rate of 1.73 mm/s. These contours illustrate the viscous recovery phenomena developed in the next 60 s to the loading-unloading curve, with regard to the minimum principal stretch. The calcaneal fat pad tissues are able to recover the residual deformation in 60 s, with a sharp decrease of the recovery stretch in the initial phase. Indeed, in the present study the results show the capability of the heel region to recover the residual deformation completely in one minute. This confirms that the one-minute-break applied between each trial of the experimental tests is a sound choice. The results show a distribution of residual stretches and stresses, indicating the visco-elastic response.

Figure 5.5. Results from the numerical analysis of compression test at 1.73 mm/s strain rate with a piston of 40 mm diameter. Contours of minimum principal stretch in the deformed configuration of the heel pad structure after a loading-unloading cycle: viscous recovery time $t_r = 0$ s (a), $t_r = 5$ s (b) and $t_r = 60$ s. Contours are reported over a transversal section AA of heel pad structure.

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Figure 5.6 shows the results obtained from experimental stress relaxation tests and numerical analyses. Specifically, the experimental loading/unloading curves (Figure 5.6(a)) and normalized load-time curves (Figure 5.6(b)), obtained by applying five loading-stress relaxation-unloading cycles on the heel pad with a velocity of 1.73 mm/s (grey lines) are compared with the curves obtained from the numerical analyses (black lines).

![Figure 5.6. Stress relaxation tests performed with a piston of 40mm diameter: (a) force vs. displacement and (b) force vs. time.](image)

Deformed configurations of the heel region when interacting with the piston of 40 mm in diameter are reported with regard to minimum principal stress (Figure 5.7(b)) after compression and after 40 s of the relaxation period (Figure 5.7(c)). All contours are reported over a transverse section of the fat pad, as reported in Figure 5.2(a).

![Figure 5.7. Contours of minimum principal stress in the deformed configuration of heel pad structure after a loading: stress relaxation time t=t_l+0 s (b) and t=t_l+40 s (c), where t_l is the time of loading. Contours are reported over a transversal section AA of heel pad structure](image)

Some observations must be made with regard to the experimental procedure. The developed compression device uses low force at a low speed compared with the
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condition of walking. This limitation is due to the fact that this device is designed for use in a clinical setting involving diseased heel pads which cannot necessarily tolerate high loads. The five curves recorded do not overlap completely, which might be due to various reasons: the tissue fixing the heel pad (skin, calcaneus, connective tissue, etc) cannot be completely fixed relative to the measurement device, thus both the rotation and translation of the heel pad tissue can occur. Moreover, muscles in the leg might not be completely relaxed during all the measurements. An unconscious pressure against, for example, the approaching piston cannot be completely ruled out. On a more speculative basis, it is not known if the individual components of the heel pad tissue return to exactly the same position after every trial, or what influence there is on possible variations in liquid content known (e.g. blood perfusion). On the contrary, the finite element analysis is not influenced by these external conditions and the five curves overlap.

The numerical model developed can be used for evaluating loading conditions which are not possible with experimental tests. For example, a small recovery time occurs between each trial or different displacements and forces imposed, allowing for a more detailed evaluation of the mechanical behavior of heel pad tissues under loading. Moreover, the numerical results allow for the investigation of the mechanical response of the calcaneal fat pad after a cycle of loading/unloading. The results show a distribution of residual stretches and stresses and the stress relaxation phenomena, pointing to the visco-elastic response.

These studies represent preliminary analyses directed at healthy subjects, which can be extended to pathological patients and, in their present form, can be interpreted as a validation of the overall procedure. These analyses may be useful for diseased patients, firstly evaluating the mechanical properties of the heel pad tissues using experimental tests and then implementing a numerical model to interpret the mechanical response of the tissues by means of damage model capable of accounting for degenerative processes already in progress.
5.3 HEEL PAD MECHANICAL BEHAVIOUR AT THE HEEL STRIKE IN BARE AND SHOD CONDITIONS

The body is exposed daily to forces resulting from the impact of the foot with the ground when walking and running. The heel strike represents the beginning of the stance phase of gait in the interaction between the heel and the ground, during which the heel pad tissues are subjected to high load values. The heel region has the ability to dampen and redistribute impact forces and represents a weight-bearing and shock-absorbing structure. Shoes are designed to make the effects of the heel strike less injurious by using different materials that, due to their damping properties, reduce the intensity of the force gradient by spreading the impulse over time. For this reason, it is important to study the mechanical behaviour of the heel pad tissues during the heel-strike and how it is influenced by footwear products (Fontanella et al., 2012b). The ground reaction force during the heel-strike phase in both bare and shod conditions must be evaluated. Different measuring devices are available to evaluate the components of the ground reaction force. The AMTI force platforms (Advanced Mechanical Technology Inc, Watertown, MA, USA) are commonly used for this purpose because of their accuracy of measurement (Chou et al., 2003; Liikavainio et al., 2007; Perron et al., 2000).

5.3.1. Experimental testing

AMTI BP400600 force plates (Advanced Mechanical Technology Inc, Watertown, MA, USA) are used to evaluate the ground reaction force. Each platform is 400 mm wide, 600 mm long and 82.5 mm high. Force plates are located halfway along a 15 m walkway. These kinds of platforms calculate the longitudinal, anterior-posterior and medial-lateral ground reaction force components.

To calculate the three dimensional kinematics data the platforms are associated to a Vicon MX (Oxford, UK) system with six high-resolution cameras.

The subject performing the experimental tests is a healthy 23–year-old female, 1.67 m tall and weighing 53 kg. The volunteer has normal feet and does not present any disease that might affect her gait.

The experimental tests aim to record the ground reaction force during bare and shod conditions.
In bare conditions lightweight retro reflective markers are placed on the skin in line with the tuber calcaneus posterior surface of the first and fifth metatarsal-phalangeal joint and of the lateral and medial malleolus (Figure 5.8).

In the shod condition, markers are placed on the shoe at the height of the previously described landmarks.

During experimental testing, the subject walks along the walkway at normal speed (1.6 m/s), first in the bare condition and then wearing running shoes. Trajectories of the markers are captured by the Vicon MX system. The ground reaction force is measured corresponding to the fifth and the sixth step. Seven trials are collected and the data obtained are averaged.

The vertical (Fz), the anterior-posterior (Fx) and the lateral-medial (Fy) ground reaction force components are reported in Figure 5.9(a) while the vertical forces measured in the bare and shod experimental conditions are illustrated in Figure 5.9(b).

The intensity and trend of the magnitude of the vertical component of the ground reaction force is reported in Figure 5.9. During the heel strike the mean value of the normal force is equal to 0.8 BW (SD: 0.13 BW) in bare conditions and 0.6 BW (SD: 0.14 BW) when walking with running shoes. The evaluation of the loading rate is important because of the possible of visco-elastic effects. In bare conditions the mean value of the loading rate during impact assumed is 23 BW/s, while in shod conditions
walking the loading rate decreases to 16 BW/s. This trend is confirmed by other authors (De Wit et al., 2000; Liddle et al., 2000; Spears et al., 2005).

![Graph](image1.png)

Figure 5.9. Results from the experimental tests: vertical and longitudinal components of the ground reaction force (GRF) in bare conditions (a) and comparison between vertical reaction force in bare and shod conditions (b)

The angle between the foot and the floor during the bare condition is calculated through the coordinates of the marker located on the skin corresponding to the posterior surface of the calcaneus and to the head of the first metatarsal bone. On closer investigation during the heel strike phase the angle between the plantar of the foot and the top surface of the platform is about 14 degrees. With regard to the shod condition the angle between the sole of the shoe and the top surface of the platform is 24 degrees. This value is in agreement with other authors who have evaluated the angle between the sole of a running shoe and the floor (Heidenfelder et al., 2008).
5.3.2. Numerical models

A solid model of the heel region is developed from MRI data of the subject’s foot, accounting for the heel bone, muscle, calcaneal fat pad and skin (Figure 5.10(a)). The numerical model is developed by a finite element discretization procedure, adopting more than four hundred thousand tetrahedral elements (Figure 5.10(b)). This discretization is capable of representing the geometry of the biological structures with sufficient accuracy and of correctly interpreting the structural behaviour of the system. A visco-hyperelastic model is used for representing the calcaneal fat pad, which can account for the typical features underlying the mechanical response of soft tissues, namely material and geometric non-linearity, almost incompressible behaviour and time-dependent phenomena (Natali et al., 2010b, 2012). The non-linearity and anisotropic characteristics of skin tissues induced by collagen fibres are also considered by providing a fiber-reinforced hyperelastic formulation (Natali et al., 2012). The evaluation of constitutive parameters for calcaneal fat pad and skin is performed by the analysis of in vitro tests and in situ tests previously developed. The mechanical behaviour of other tissues is specified by constitutive formulations from the literature. The mechanical response of bone tissue is evaluated by an orthotropic linear elastic model (Natali et al., 2010a), while a hyperelastic formulation is assumed for soft tissue according to data from the literature (Goske et al., 2006).

Figure 5.10. Solid models of the heel region and of the running shoe: longitudinal section to show calcaneus, soft tissues, fat pad tissue and skin of the heel region and insole, midsole, outsole and heel counter of the running shoe (a). Numerical model of the floor and of the heel region, with indication of transversal section AA (b)
A solid model of the running shoe used during the experimental tests is developed in order to evaluate the interaction phenomena between the heel pad region and the shoe during the heel strike. The virtual solid model is composed of the insole (3.0 mm thickness), the midsole (20.0 mm thickness) and the outsole (6.0 mm thickness) layers. The heel counter is defined as a 3.0 mm thick semi-rigid cup designed to fit around the heel (Spears et al., 2007) (Figure 5.10(a)).

Different commonly used elastomeric foams are considered to represent the possible materials of the soles of the running shoes. The structural configuration of the elastomeric foams reduces shock phenomena and enables a redistribution of loads on the overall heel structure. The materials for midsole are classified as ethylene vinyle-acetate foams, Medium Density EVA (indicated as 1 in Figures 6 and 7) and Medium Density Lunasoft (indicated as 2 in Figures 6 and 7), while the foams for insole are polyurethane foams, Diabetic PORON® (indicated as A in Figures 6 and 7), and a cross-linked closed-cell polyethylene, Plastazote Medium (indicated as B in Figures 6 and 7). The outsole is composed by a layer with properties of high density EVA foam (Cheung and Zhang, 2008; Shariatmadari et al., 2009). Mechanical properties of the foams are evaluated by analysing experimental data from mechanical tests. Because of the accuracy of experimental procedure and data, results from (Shariatmadari et al., 2009) are assumed. Shariatmadari et al. (2009) developed compressive tests on cylindrical samples of the selected foams. These samples measure 50.0 mm in diameter and 20.0 mm in thickness, apart from Plastazote and PORON® which are 10.0 mm and 6.0 mm respectively. The tests are performed by up to 40% deformation, loading at the rate of 10 mm/s. A more refined characterization of a foam response should require data from more experimental situations, such as those that calculate biaxial and volumetric mechanical properties (Petre et al., 2006), but no other data have been found in the literature. Results from the experimental tests show the non linearity of the mechanical behaviour that is interpreted by a hyperelastic formulation. The associated strain energy function is represented by the following equation (Fontanella et al., 2012b):

\[
W(\lambda_1, \lambda_2, \lambda_3) = \sum_{i=1}^{2} \frac{2\mu_i}{\alpha_i} \left[ \lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3 + \frac{1}{\beta} \left( \left( J^{el} \right)^{-\alpha_i\beta} - 1 \right) \right]
\]  

(5.1)
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where $\lambda_1, \lambda_2, \lambda_3$, are the principal stretches and $\mu, \alpha_i, \beta$ are the material properties representing the compressible foam behaviour. The parameters $\mu, \beta$ are related to the initial shear modulus, as $G_0$, the initial bulk modulus, as $B_0$, and the Poisson ratio, as $\nu$, by the following equations:

$$G_0 = \sum_{i=1}^{\nu} \mu_i$$  \hspace{1cm} (5.2)

$$B_0 = \sum_{i=1}^{\nu} 2\mu_i \left( \frac{1}{3} + \beta \right)$$  \hspace{1cm} (5.3)

$$\nu = \frac{\beta}{1 + 2\beta}$$  \hspace{1cm} (5.4)

Mechanical tests show that often small values of transversal deformations occur. This consideration leads to the assumption of a simplified formulation. Considering the constitutive formulation, the simple geometry of the specimens and the boundary conditions, analytical models are provided to interpret the experimental tests. The identification of constitutive parameters is performed using the cost function which evaluates the discrepancy between analytical results and experimental data. The minimisation of the cost function, which is performed by a stochastic-deterministic optimization algorithm, leads to the definition of the optimal set of parameters (Table 5.4).

<table>
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<th>Material Type</th>
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<th>$\mu_1$</th>
<th>$\mu_2$</th>
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</tr>
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</tr>
<tr>
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<td>3.24 · 10^4</td>
<td>0 · 10^9</td>
</tr>
</tbody>
</table>

Table 5.4. Constitutive parameters adopted for the mechanical characterisation of different foams
The material of the heel counter, which represents the confining element of the heel region, is described by a linear elastic formulation according to data in the literature (Spears et al., 2007).

5.3.3. Numerical analyses

Numerical analyses are developed to evaluate the heel strike phase during bare and shod walking. According to experimental conditions, a numerical model of the floor is developed, composed of the AMTI force platform (10.0 mm) and covered by a rubber layer (3.0 mm), (Figure 5.10(b)). The AMTI force platform and the rubber layer are defined using a linear elastic model.

A preliminary analysis is performed to evaluate the mechanical response of the heel pad tissues in the heel strike in bare-foot conditions. The magnitude of the heel strike transient force peak occurs at 0.033 s of the gait cycle. The numerical model of the heel region is positioned according to experimental tests.

The superior region of the calcaneus is fixed and the inferior surface of the plane is loaded with a force and a loading rate defined on the basis of the experimental data. The skin-platform interface is modelled using contact surfaces with a friction coefficient of 0.61 according to references in the literature, which report a friction coefficient between the skin and the rubber layer in normal conditions (Zhang and Mak, 1999).

A further analysis is performed to estimate the mechanical response of the heel pad tissues in the heel strike in shod conditions. From the experimental tests, the magnitude of the heel strike transient force peak occurs at 0.043 s of the gait cycle. The numerical model of the heel region and of the running shoe is positioned to imitate the heel strike phase observed during the experimental trials.

As in the previous analysis, the superior surface of the calcaneus is fixed and the inferior surface of the plate is loaded. The skin-insole and the outsole-platform interface are modelled using a contact surface with friction coefficients of 0.51 and 0.35, respectively, according to references in the literature (Li et al., 2004; Natali et al., 2010c; Zhang and Mak, 1999). On closer investigation, these data interpret the friction coefficient between skin and cotton and between outsole and rubber layer, according to experimental conditions.
Different numerical analyses are made in order to assess the influence of the different foams on the mechanical behaviour of the heel pad tissues.

5.3.4. Numerical results

The deformed configuration of the heel pad structure during the heel strike is evaluated with regard to bare and shod conditions. In Figure 4 results referring to the displacement along the vertical direction are reported for bare (Figure 4a) and shod (Figure 4b) conditions, where the insole is composed of Diabetic PORON® and the midsole of Medium Density Lunasoft (group A2). The displacement field of the heel pad tissues which occurs during the barefoot heel strike is comparable with the results reported by other authors (Spears et al., 2005).

Figure 5.11. Results from the numerical analysis: contours of displacement fields in bare (a) and shod (b) conditions, where the insole is composed of Diabetic PORON® (insole) and the midsole of Medium Density Lunasoft (midsole). Contours of minimum principal stretch fields in bare (c) and shod (d) conditions, where the insole is composed of Diabetic PORON® (insole) and the midsole of Medium Density Lunasoft (midsole). Contours are reported over a transversal section AA of the heel pad region.
Figure 5.11 reports contours of minimum principal stretch. It is interesting to point out the differences in the deformation fields between bare (Figure 5.11(a)) and shod (Figure 5.11(b)) conditions, in particular the effects of bounding determined by the footwear and particularly by the heel counter (Fontanella et al., 2012b). In Figure 5.12, contours of minimum principal stress are reported. In general, the results confirm that peak stress values are found in the heel pad tissues in the bare condition, while this phenomenon does not occur during the shod condition because of the mechanical role of the shoe.

Figure 5.12. Results from the numerical analysis: contours of minimum principal stress fields in bare (a) and shod (b) conditions, where the insole is composed of Diabetic PORON® (insole) and the midsole of Medium Density Lunasoft (midsole). Contours are reported over a transversal section of the heel pad region.

Figure 5.13. Diagram of the minimum principal stress along the path during heel strike in bare and shod conditions, depicted by black circles in the heel pad region. The results are reported as a combination of different materials: insoles are assumed as A (Diabetic PORON®) and B (Plastazote Medium), while midsoles are assumed as 1 (Medium Density EVA) and 2 (Medium Density Lunasoft), leading to groups A1 and B2.
Figure 5.13 shows the minimum principal stress that occurs in the heel pad during the heel strike, reported by the black circles and quoted in sequence from left to right. The diagram shows the differences between bare and shod conditions, where the shoe limits the bulging of the heel pad. With regard to the same path located in the heel pad, Figure 5.14 shows the difference in minimum principal stress determined by the different materials for the insole, indicated as A (Diabetic PORON®) and B (Plastazote Medium), and for the midsole, indicated as 1 (Medium Density EVA) and 2 (Medium Density Lunasoft). The minimum principal stress in the bare condition is about 34% higher than the minimum principal stress in the heel pad during shod conditions, in agreement with other authors (Even-Tzur et al., 2006; Verdejo and Nills, 2004).

Figure 5.14. Minimum principal stress along the path during the heel strike for different materials, depicted by black circles in the heel pad region (see Figure 4). The results are reported as a combination of different materials: insoles are assumed as A (Diabetic PORON®) and B (Plastazote Medium), while midsoles are assumed as 1 (Medium Density EVA) and 2 (Medium Density Lunasoft), leading to groups A1, A2 and B1, B2.

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Similar studies which assess the influence of mechanical properties of different materials used for the shoe show differences in the mechanical response of heel pad tissues (Aerts and De Clercq, 1993; Cheung et al., 2008; Goske et al., 2006; Shariatmadari et al., 2010; Fontanella et al., 2012b) because of the different materials assumed within the shod components.

This preliminary analysis is affected by some approximations due to several basic assumptions and to the specific operational procedure. The limit that derives from the consideration of a single subject is overcome by a comparison of the specific data with the references in literature. A minor discrepancy has been found between data revealed and average data reported in literature, thereby involving a valid agreement. Shoe material properties deeply influence stress and strain distribution within the calcaneal fat pad structure. This is not a relevant limit as it is possible to represent almost any shoe configuration in the numerical model. It must be pointed out that the loading assumed can be interpreted as the one occurring when the subject simply places the foot on the force plate in a quasi-static condition.

The present model represents the first act of a more extended research activity involving an accurate consideration of different materials and conformation of running shoes and, moreover, evaluates in greater detail the structural conformation of the overall ankle joint with regard to different configurations, also in terms of loading. The complex configuration of loading in terms of a combination of vertical, transversal and longitudinal components must be investigated, taking into account the strain rate effects, both with regard to the shoe and foot response. In this sense, the valid results obtained represent a positive basis for the development of an analysis that can consider the most general configuration in terms of model configuration and loading conditions, but, most of all extend the investigation to the fore region of the foot.
5.4 REFERENCES


CONCLUSIONS

The human heel pad is able to provide a valid mechanical function under different loading conditions and to reduce potential injury to the body during gait cycle. The ability of the heel pad to perform these functions has a fundamental influence on overall foot mechanics.

The investigation of the foot biomechanical response is beset with relevant problems related to the constitutive identification of tissue mechanical behaviour and to their complex structural configuration. A combined experimental-numerical approach seems to offer the possibility of a profitable integration of data, aimed at both characterising materials and validating final results. Constitutive modeling is rather complicated, but this complexity is essential for a proper interpretation of the mechanical behaviour of the tissues. Hyperelastic and visco-hyperelastic models are used for representing soft tissues with regard to the material and geometric non-linearity, the almost incompressible and the time dependent response. The anisotropic characteristics of skin tissues induced by collagen fibres are also considered.

Constitutive formulations are provided according to a phenomenological approach and accounted for different experimental data. The achieved formulations are capable of a comprehensive interpretation of the tissue mechanical behaviour.

Within the constitutive formulation assumed for the different tissues, the constitutive parameters must be defined and this action is provided by means of a stochastic-deterministic procedure. Improvement in result reliability is a consequence of the reduction in the discrepancy between experimental data and model results, which considers the comparison between different experimental situations. This is obtained by the minimisation of a cost function that is a direct representation of the progressive agreement between the data themselves.

It must be noted that the evaluation of constitutive parameters is provided by considering experimental data from different authors and defining an optimal region for the set to be used. In this way, the problem of biological variability of mechanical
Conclusions

properties in heel pad tissues is taken into account. A more exhaustive evaluation of this aspect will require a larger set of experimental data.

Numerical models allow for a possible representation of variable loading conditions and tissue properties with a lower effort with regard to the repetitive action requested by experimental testing. The accuracy of results and the possible extensive definition of parameters, in terms of strain and stress, that are not easily accessible for experimental analysis, confirm the potentiality of the numerical approach. Nonetheless, it requires the basic information from mechanical testing data. In the numerical model the operational relevance of the computational effort must also be considered, due to the presence of a coupled material and geometric non linear problem.

An important task is to evaluate the actual influence of calcaneal fat pad tissue structural configuration, considering the adipose chambers and fibrous septa conformations on the overall mechanical response. The numerical model of calcaneal fat pad developed has proven its ability to interpret the correlation between the histological and morphometric configuration, with regard to the adipose chambers and the connective septae, tissues mechanical properties and the overall mechanical response, comparing it with experimental data. In particular, the results from numerical tests underline the mechanical contribution of adipose tissue and connective tissue on the mechanical response of the calcaneal fat pad. The compressive response of calcaneal fat pad structure is influenced by the adipose tissue, which is almost incompressible, together with the action of fibrous septae that limit the bulging of the chambers themselves. Alterations due to degenerative effects determine variations in the mechanical behaviour of the heel pad region as regards the biomechanical properties of tissues and its structural conformation.

An important task involves the possibility of wide application of the numerical models to evaluate, for example, the interaction phenomena occurring between the foot and orthosis and/or footwear products, to estimate reliability and comfort in relevant tasks for industrial applications. More attention is focused on the comparison between bare and shod conditions and on the evaluation of the influence of different shoe materials. The numerical analyses allow for the definition of the distribution of strain and stress in the biological tissues during the heel strike. These studies show differences in the
mechanical response of heel pad tissues because of the different materials assumed within the shoe structure.

A brief mention must be given to the possibility of representing degenerative phenomena occurring in hard and mostly soft tissues using numerical analyses, resulting in a valid support for healthcare systems diagnosis and therapy. This study reports a preliminary analysis applied to a healthy subject, which can be interpreted as a validation of the overall procedure and can be extended to patients with a pathology. These analyses may be useful for patients, firstly evaluating the mechanical properties of the heel pad tissues using experimental tests and then implementing a numerical model capable of interpreting the mechanical response of the tissues by means of a damage model which accounts for degenerative processes.
ANNEX ONE

A NUMERICAL MODEL FOR INVESTIGATING THE MECHANICS OF THE CALCANEAL FAT PAD

A.1 INTRODUCTION

The mechanical response of calcaneal fat pad tissue is strongly influenced by the structural conformation, i.e., the dimension of adipose chambers, the thickness of connective septae walls and the mechanical properties of the different soft tissues. In order to define the constitutive formulation of adipose tissues, experimental data from pig specimens are considered, according to their functional similarity, while the mechanical response of connective tissue septae is assumed with regard to the mechanical behaviour that characterize ligaments. Different numerical models are provided accounting for the variation of chambers dimensions, septae wall thickness and tissues characteristics. The spiral angles of collagen fibres within the septae influence the capability of the structure to withstand the bulging of chambers. The analysis considers different orientation of the fibres. The response of calcaneal fat pad region is evaluated in comparison with experimental data from unconfined compression tests. The present work provides a preliminary approach to enhance the correlation between the structural conformation and the mechanical properties of tissues towards the biomechanical response of the overall heel pad region.

A.2 CALCANEAL FAT PAD MECHANICS

The calcaneal fat pad is a complex structure of adipose and connective tissues which surrounds the calcaneus in the posterior foot region. The mechanical role of the calcaneal fat pad tissues is to optimize the response to loading, for example, by dumping shocks generated during the gait or running cycle (Whittle, 1999), making it possible to achieve a smooth distribution of the pressure field. In this way, the calcaneal fat pad configuration performs the function of a weight-bearing and shock-
absorbing structure (Miller, 1982). The overall mechanical response of the calcaneal fat pad region is influenced by the configuration of the histologic and morphometric connective and adipose tissues, and by the structural conformation of the septal system (Buschmann et al., 1993; Natali et al., 2010). The mechanical response has been previously investigated by numerical formulations capable of interpreting the overall behaviour of the structure. Phenomenological formulations were provided by different authors to specify the stress-strain relationship of tissues (Fred and Diethelm, 2006; Ledoux and Blevins, 2007; Miller-Young et al., 2002; Natali et al., 2010).

A.3 HISTOLOGIC AND MORPHOMETRIC ANALYSIS

The calcaneal structure is complex and hierarchically organized, consisting of dense strands of elastic fibrous tissue, which defines septae that bind adipose tissues in chambers of different size (Bleschmidt, 1982; Jahss et al., 1992a; Kuhns, 1949). Collagen fibres within septae are spirally arranged and superiorly fixed to bone or other septae, and inferiorly to other septae of smaller size and to dermis. The adipose chambers contain packed fat cells which are arranged in lobules joined by connective tissues. Lobules are grouped by further bundles of collagen and elastin fibres (Jahss et al., 1992b). The dimension and orientation of adipose chambers depend on their specific location within the calcaneal region. Their dimension is small in the region close to dermis and progressively increases towards the calcaneal perichondrium (Hsu et al., 2007). In the central portion, the chambers are oriented vertically, while in the lateral and posterior regions they are smaller and transversally oriented (Rome, 1998). The stiffness of the calcaneal region is determined by the almost incompressible behaviour of adipose tissue bounded by a collagen network (Torp-Pedersen et al., 2008). The spiral distribution of collagen fibres within connective septae is suitable for limiting the bulging of chambers (Rome, 1998).

The calcaneal tissue properties and their structural conformation provide an optimal mechanical response to the applied loads. Alterations caused by aging or degenerative effects influence the mechanical properties of the tissue (Alcantara et al., 2002; Hsu et al., 2009; Kwan et a., 2010; Pai and Ledoux, 2010; Tsai et al., 1999). Degenerative phenomena can determine a gradual loss of collagen, a decrease in the quantity of the elastic fibrous component and a reduction of liquid content. This is associated with a
local loss of fat tissue and damage to the fibrous tissue septae through the distortion and the fragmentation of the fibrous proteins (Prichasuk et al., 1994). Furthermore, increased fat content correlated with obesity leads to increased pressure in the specific closed space and determines a growing local stiffness (Kuhns, 1949; Mirrashed et al., 2004). Pathologies such as diabetic or atrophied phenomena cause alterations in the ratio of saturated and unsaturated fatty acids and determine a variation in the capacity to withstand the stress. Fat globules decreases in size and number and also a reduction of the number of septal fibres takes place. Moreover, the connective septae become thicker and contain a slightly higher percentage of fibrous tissues, thereby assuming a fragmented configuration (Buschmann et al., 1995; Gefen, 2010). Under such conditions, experimental tests confirm the presence of stiffer soft tissues, a less smooth plantar pressure distribution with higher local values and a reduction in the ability to absorb shock phenomena.

### A.4 CONSTITUTIVE MODELS

The biomechanical contribution of adipose and connective tissues can be investigated on the basis of experimental data available (Natali et al., 2012). Experimental tests performed on specimens from pig adipose tissues are considered because of similarities in structural configuration and overall mechanical response to human tissues (Comley and Fleck, 2010; Geerligs et al., 2008). The mechanical properties of connective tissue septae are considered to be similar to the mechanical behaviour of connective tissues, such as ligaments or tendons (Comley and Fleck, 2010).

With regard to adipose chambers, experimental data from mechanical tests performed on adipose tissue show the strong non-linearity of the mechanical response, such as material and geometrical non linearity, almost incompressible and time-dependent behaviour (Comley, Fleck 2009, Comley, Fleck 2010, Geerligs et al., 2008). Different constitutive formulations are reported in the literature for subcutaneous adipose tissues. With the aim of analysing the short time response, an hyperelastic constitutive model is adopted (Natali et al., 2012):

\[
W(C) = U(C) + \tilde{W}(C) \quad (A.1)
\]
where $W$ is the hyperelastic potential. Considering the almost incompressible behaviour, the hyperelastic potential $W$ can be split into volumetric $U$ and iso-volumetric $\tilde{W}$ terms\(^1\).

The evaluation of mechanical response of adipose tissues considers experimental tests on pig adipose tissue specimens (Klein et al., 2007). Experimental data from compression tests on cylindrical specimens developed by Comley and Fleck (2009) are considered. Specifically, circular cylindrical specimens of adipose tissues are cut from jowl of the pigs (diameter 10 mm; height 3 mm) and compression tests are performed at different strain rates.

The identification of constitutive parameters within the hyperelastic constitutive model developed is obtained according to data from high strain rate tests. Due to the simple geometry of specimens and the uni-axial loading configuration, a preliminary analysis is performed by an analytical model that interprets the experimental conditions. The evaluation of constitutive parameters is performed by the minimization of a cost function which specifies the discrepancy between experimental data and model results.

\(^1\)The hyperelastic constitutive model is defined by the strain energy function $W(C)$ (Marsden and Hughes, 1968). The second Piola-Kirchhoff stress tensor $S$ is computed to satisfy thermodynamic requirements (Holzapfel, 2000; Simo and Hughes, 1998), as:

$$S(C) = 2 \frac{\partial W(C)}{\partial C}$$  \hspace{1cm} (A.2)

Almost-incompressible behaviour of the tissue is assumed and the strain energy function can be split into volumetric $U$ and iso-volumetric $\tilde{W}$ contributions (Flory, 1961):

$$W(\tilde{I}, J) = U(J) + \tilde{W}(\tilde{I})$$  \hspace{1cm} (A.3)

where $J$ is the Jacobian deformation, as $J = \sqrt{\text{det}(C)}$, while $\tilde{I}$ is the first iso-volumetric invariant of the right Cauchy-Green strain tensor, as $\tilde{I} = \text{tr}(J^{-2/3}C)$. Because of the characteristic non-linearity of tissue response, specific polynomial and exponential functions are assumed to aptly represent the trend of the energy components, as in (Natali et al., 2004; Natali et al., 2008):

$$U(J) = \frac{K_v}{2 + r(r+1)}\left[ (J - 1)^2 + J^{-r} + rJ - (r+1) \right]$$  \hspace{1cm} (A.4)

$$\tilde{W}(\tilde{I}) = \frac{C_1}{\tilde{I}^r} \left[ \exp\left(\alpha_1(\tilde{I} - 3)\right) - 1 \right]$$  \hspace{1cm} (A.5)

Constitutive parameters $K_v$ and $C_1$ are related to the initial volumetric and shear stiffness, respectively. Parameters $r$ and $\alpha_1$ characterize the evolution of material stiffness with strain because of the non-linearity of the mechanical response highlighted by mechanical tests.
The minimization takes into account a stochastic-deterministic procedure (Natali et al., 2009a).

The mechanical behaviour of connective tissue septae, such as the anisotropic behaviour, material and geometrical non-linearity is determined by the complex configuration of the tissue themselves, being a composite material made of oriented fibres embedded within a ground matrix (Natali et al., 2009b). According to histological evidence, a fibre-reinforced hyperelastic model is adopted to interpret the mechanical behaviour of connective septae, accounting for different contributions from isotropic ground matrix and fibre families (Limbert and Taylor, 2002; Natali et al., 2008; Natali et al., 2009b; Ottani et al., 2001; Spencer, 1984; Weiss et al., 1996).

The following general formulation for strain energy is adopted:

\[
W_{\text{m}}(\mathbf{C}, \mathbf{a}_0^i \otimes \mathbf{a}_0^i) = W_{\text{m}}(\mathbf{C}) + \sum_{i=1}^{n} W_f^i \left( \mathbf{C}, \mathbf{a}_0^i \otimes \mathbf{a}_0^i \right) \tag{A.6}
\]

where \( W_{\text{m}} \) is the term which specifies the isotropic ground matrix contribution, while \( W_f^i \) is the \( i \)-th fibres family term and vector \( \mathbf{a}_0^i \) is a unit vector that defines fibre orientation.\(^2\)

\(^2\) With regard to the constitutive model adopted for connective septae, the hyperelastic formulation has to account for different contributions from fibres and the isotropic ground matrix (Holzapfel et al., 2000; Limbert and Taylor, 2002; Spencer 1984; Weiss et al., 1996):

\[
W \left( J, \tilde{I}_i \right) = W_m \left( J, \tilde{I}_i \right) + \sum_{i=1}^{n} W_f^i \left( I_i \right) \tag{A.7}
\]

where \( W_m \) is the term which specifies the isotropic ground matrix, while \( W_f^i \) specifies the mechanical response of the \( i \)-th fibre family. \( I_i \) is a structural invariant related to the tissue stretch along the \( i \)-th fibre family direction \( \mathbf{a}_0^i \), as \( I_i = \mathbf{a}_0^i \cdot \mathbf{C} \mathbf{a}_0^i = (\lambda^i)^2 \). Because of the predominant role of the fibres, a neo-hookean formulation can be assumed to interpret the mechanical response of the ground matrix, as almost-incompressible materials:

\[
W_m \left( J, \tilde{I}_i \right) = K_v \left( J - 1 \right)^2 + C_1 \left( \tilde{I}_i - 3 \right)^2 \tag{A.8}
\]

where \( K_v \) can be related to the volumetric stiffness, while \( C_1 \) is associated with the shear stiffness of tissues. The mechanical contribution of fibres can be described considering their structural organization (Ottani et al. 2001). In the unstrained configuration, fibres are characterised by a wavy conformation.
Mechanical tests are performed to determine the contribution of the ground matrix and the fibres on the overall mechanical response. In literature, different experimental tests are reported on human tendons and ligaments being considered (Funk et al., 2000; Wren et al., 2003) in order to determine the properties of the connective septae tissues. Tensile tests performed on ligaments of the human foot are analysed (Attarian et al., 1985) assuming high strain rates to interpret the short time response of the tissue. The fibre-reinforced hyperelastic model is considered and the constitutive parameters are evaluated, following the procedure described in Natali et al. (2009a).

### A.5 NUMERICAL ANALYSES

A numerical model of a cylindrical specimen of calcaneal region has been developed to evaluate the influence of the septal size and shape on the overall mechanical response during compression tests. The dimension of the region considered by the numerical model considers the specimen dimension reported by Miller-Young (2002), adopted in consideration of accuracy and completeness. The numerical model is divided into adipose chamber tissue and connective septae tissue (as reported in Figure A.1).

Assuming the dimensions reported in the literature (Bleschmidt, 1982; Buschmann et al., 1993; Cichowitz et al., 2009; Robbins et al., 1989) and considering the measurements from MRI images, for healthy calcaneal fat pad tissue, the dimension of the adipose chambers ranges between 1.0 mm and 5.0 mm, while the thickness of the connective septae ranges between 0.8 mm and 2.0 mm. The configuration of calcaneal fat pad structure is assumed according to 2.4 mm adipose chambers diameter and about 1.0 mm connective septae thickness. The configuration of the adipose chambers

When tensile load is applied, fibres first uncrimp and then become stretched. This mechanism determines a strongly non-linear mechanical response that can be described by an exponential formulation:

\[
W_i(T_i') = \frac{C_4}{\alpha_4} \left\{ \exp\left[\alpha_i(T_i' - 1)\right] - \alpha_i(T_i' - 1) - 1 \right\}
\]

where \( C_4 \) is a constant that defines the initial stiffness of the fibres, described by \( E_f = 4C_4 \), while \( \alpha_4 \) depends on the initial wavy conformation of fibres. The number of fibre families is evaluated according to tissue histology.
is in agreement with the fact that the experimental specimens are taken in the central portion where the chambers are oriented vertically. The model is developed to report the unit reference volume that is representative of the specimen of the calcaneal region as performed in experimental tests.

![Numerical model of fat pad tissue](image)

**Figure A.1.** Numerical model of fat pad tissue (a), including adipose chambers and connective septae, according to the conformation shown in the transversal section AA (b) and the longitudinal section BB (c) reported.

The connective tissues septae are organized in fibres arranged spirally around the adipose chambers and numerical analyses are performed to evaluate the influence of the different orientations and distributions on mechanical response. In literature, a few details are reported about the orientation and direction of collagen fibres in the connective septae. However, fibre orientation can be assumed according to other biological structures which perform a similar mechanical function. For example, the fibre orientation of the human annulus fibrosus is in the range of 25–45 degrees to the transverse plane and located in different subsequent planes (Yin and Elliot, 2004).
Consequently, the most probable condition is described considering two groups of fibres distributed over the entire volume considered, one running in a clockwise and the other in an anticlockwise direction. Different angles values are adopted, namely 15, 30 and 45 degrees, aiming to arrive at a preliminary sensitivity evaluation of the response dependent on this variation.

To evaluate the influence of structural conformations, different configurations of the numerical models have been developed starting from data obtained from the literature and MRI investigations. Small chambers are assumed with the minor diameter of the chamber of 1.6 mm and the septae thickness of 1.8 mm, while larger chambers are assumed with the diameter of 3.0 mm and the septae thickness of 0.4 mm. Numerical analyses are performed in line with experimental conditions described by Miller-Young (2002). The numerical model is laterally unconstrained and squeezed between two steel platens. The bottom platen is fully fixed, while the top platen moves downwards to reach an approximate 50% deformation. The friction coefficient at the contact interface of specimen and platens is 0.1 (Wu et al., 2004). The hyperelastic constitutive models are implemented in the general purpose finite element software ABAQUS (Simulia, DassaultSystèmes, Providence, RI, USA) by developing ad hoc routines able to interpret the behaviour of adipose and connective tissues.

### A.6 RESULTS

The procedure adopted for the identification of the constitutive parameters follows the experience developed in the analysis of soft tissue mechanics (Natali et al., 2009a; Natali et al., 2009b; Natali et al. 2010).

<table>
<thead>
<tr>
<th>$K_v$ (MPa)</th>
<th>$r$</th>
<th>$C_1$ (MPa)</th>
<th>$\alpha_1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$2.31 \cdot 10^{-1}$</td>
<td>$2.74 \cdot 10^{-1}$</td>
<td>$2.19 \cdot 10^{-1}$</td>
<td>$5.35 \cdot 10^{-1}$</td>
</tr>
</tbody>
</table>

*Table A.1. Hyperelastic parameters for adipose tissue*

<table>
<thead>
<tr>
<th>$K_v$ (MPa)</th>
<th>$C_1$ (MPa)</th>
<th>$C_4$ (MPa)</th>
<th>$\alpha_4$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$2.02 \cdot 10^{-2}$</td>
<td>$4.63 \cdot 10^{-3}$</td>
<td>$2.36 \cdot 10^{-1}$</td>
<td>$5.48 \cdot 10^{-1}$</td>
</tr>
</tbody>
</table>

*Table A.2. Fibre-reinforced hyperelastic parameters for connective tissue*
In this context, the use of an optimization procedure leads to the constitutive parameters reported in Tables A.1 and A.2, with regard to adipose tissues on pig specimen and anterior talofibular human ligament, respectively.

In Figure A.2, results are reported on pig adipose tissue and on anterior talofibular human ligament specimens. Model results (continuous lines) are compared to experimental data from compression tests (Figure A.2(a)) showing the capability of the specific hyperelastic model to interpret the mechanical properties of adipose tissues. Model results (continuous lines) from tensile tests on connective tissues (Figure A.2(b)) are compared with experimental data on anterior talofibular ligament, also providing for a fitting of the constitutive parameters within the fibre-reinforced hyperelastic model adopted.

Figure A.2. Comparison of experimental (empty dots) and model results (continuous lines): (a) compression tests on cylindrical pig adipose tissue specimens and (b) tensile tests on talo-fibular human ligament

Figure A.3. Comparison of experimental (empty dots) and model results (continuous lines) for unconfined compression tests on calcaneal fat pad samples at high strain rate (Miller-Young et al., 2002)
In Figure A.3, experimental data from Miller-Young et al. (2002) are compared with model results from the numerical model of a healthy configuration of a calcaneal specimen. The correspondence of model results and the data reported by experimental data confirms the capability of the numerical model to interpret appropriately the problem under investigation. Deformed configurations of the calcaneal region under unconfined compression tests (Figure A.4) are reported with regard to the contours of minimum and maximum principal stress fields. In detail, Figure A.4(a) represents the minimum principal stress fields that characterize the response of the adipose chambers. Figure A.4(b) represents the contribution of connective septae in the overall response under compression. The compressive stiffness of the calcaneal fat pad structure is influenced by the adipose tissue, which is almost incompressible, together with the tension of fibrous septae that limit the bulging of the chambers themselves.

![Figure A.4](image1.png)

*Figure A.4. Results from the numerical model of the calcaneal fat pad region: contours of minimum principal stress fields (a) and maximum principal stress fields for unconfined compression test, reported on a transversal section BB of the calcaneal fat pad numerical model.*

![Figure A.5](image2.png)

*Figure A.5. The stress-strain relationship from the numerical analysis computed on models with different chamber dimensions: small (model a), normal (model b) and large (model c) chambers (a). The stress-strain relationship from the numerical analysis computed on models with different orientation of collagen fibres: 45° (model d), 30° (model e) and 15° (model f) (b).*
The numerical models allow for the possibility to evaluate the variation of the volume ratio between the adipose chamber and fibrous septae and the mechanical properties of the tissues themselves. Different numerical models of the calcaneal region are developed to evaluate the influence of chambers dimension on mechanical behaviour. In Figure A.5(a), model results from different numerical models are reported. In detail, three different configurations of the numerical model are reported: small chambers (model a), medium chambers (model b) and large chambers (model c). The increase in chamber dimensions, in association with septae configuration, provides a variation of the local stiffness. In Figure A.5(b), the stress-strain relationship is reported with regard to an angle of 45 (model d), 30 (model e) and 15 (model f) degrees that the fibres form with the horizontal plane.

![maximum principal stress (MPa)](image)

Figure A.6. Results from the numerical analyses: contours of maximum principal stress fields of unconfined compression test for normal (a - b) and large (c - d) chambers. Contours are reported on a transversal section BB of the calcaneal fat pad numerical model

The mechanical behaviour of the tissues is underlined by the contour of maximum principal stress in Figure A.6. In particularly, the calcaneal fat pad specimen with large adipose chambers (Figure A.6(a)) provides higher values of stress on connective septae than normal adipose chambers (Figure A.6(b)). In Figure A.7, the contours of minimum and maximum principal stress fields in unconfined compression tests for 15 (a-c) and 45 (b-d) degrees collagen fibre orientation are reported, respectively.
The influences of tissue mechanical properties as well as structural conformation are highlighted. The numerical models make it possible to evaluate also the variation of the volume ratio between adipose chamber and fibrous septae and the mechanical properties of the tissues themselves.

The degenerative effects due to age, trauma, or pathologies, such as diabetes or peripheral neuropathies, cause histologic changes in the heel pad structure of the foot. A change occurs in the dimension of the chambers, in septal thickness and in the orientation of fibres. These alterations determine variations in the mechanical behaviour of the heel pad region in terms of tissue biomechanical properties and structural conformation. Both terms must be evaluated to interpret the mechanical response.
A.7 REFERENCES


Annex 1


A numerical model for investigating the mechanics of the calcaneal fat pad


Biomechanical analysis of heel pad tissues

CMBM – Centre of Mechanics of Biological Materials – University of Padova, Italy


