BIOMECHANICAL ANALYSIS OF THE DIABETIC FOOT: AN INTEGRATED APPROACH USING MOVEMENT ANALYSIS AND FINITE ELEMENT SIMULATION

Relatore: Ch.mo Prof. Ing. CLAUDIO COBELLi
Correlatori: Ing. ZIMI SAWACHA, PhD
Ing. ANNAMARIA GUIOTTO

Laureando: SCARTON ALESSANDRA

ANNO ACCADEMICO 2011-2012
# Contents

**Introduction** 8

1 The Foot 9
   1.1 Introduction .............................................. 9
   1.2 Anatomy of the foot ........................................ 9
   1.3 Function of the foot ....................................... 11
   1.4 The gait cycle ............................................. 12

2 The diabetic foot 15
   2.1 Introduction ................................................ 15
   2.2 Epidemiology ............................................... 16
   2.3 Complications ............................................... 17
   2.4 The diabetic foot .......................................... 19
      2.4.1 Neuropathic foot ...................................... 19
      2.4.2 Ischemic foot .......................................... 20
   2.5 Plantar ulcers and plantar pressures ..................... 20
      2.5.1 Plantar pressure measurement .......................... 21

3 Motion analysis 23
   3.1 Introduction ................................................ 23
      3.1.1 Stereophotogrammetric systems ......................... 24
      3.1.2 Force platforms ....................................... 28
      3.1.3 Calibration of force platforms ......................... 30
      3.1.4 Pressure Platforms .................................... 31
      3.1.5 Pressure Insoles ....................................... 34
   3.2 Protocols .................................................. 35
      3.2.1 Protocol used in this work ............................. 37

4 Foot Finite Element Model 43
   4.1 Introduction ................................................ 43
   4.2 Finite Element Models ..................................... 43
      4.2.1 Discretized geometry .................................. 44
4.3 FEM models of the foot ........................................... 46
4.4 Workflow to obtain a model of the foot ....................... 50
  4.4.1 Computed tomography and Magnetic Resonance Imaging 51
  4.4.2 Segmentation ............................................. 52
  4.4.3 Cad modelling ........................................... 52
  4.4.4 Fem Modelling ........................................... 53

5 Materials and methods ........................................... 57
  5.1 Introduction ............................................... 57
  5.2 Subjects .................................................. 57
  5.3 MRI acquisition ........................................... 58
  5.4 Segmentation ............................................... 58
  5.5 Construction of the geometry ............................... 61
  5.6 Finite Element Models .................................... 62
    5.6.1 COMSOL Multiphysics ................................ 62
    5.6.2 Abaqus/ Cad Environment Complete .................. 64
  5.7 Kinematic and kinetic data ................................ 67

6 Results .......................................................... 71
  6.1 Introduction ............................................... 71
  6.2 Healthy subject ............................................ 72
    6.2.1 Experimental vs simulated data in barefoot condition 73
    6.2.2 Experimental vs simulated data of all the phases of the
        stance phase in barefoot condition ....................... 75
    6.2.3 Experimental vs simulated data of the initial contact
        and the loading response phases in barefoot condition 80
    6.2.4 Experimental vs simulated data in the frame of maximum
        medio-lateral force in barefoot condition ............... 82
    6.2.5 Experimental vs Simulated healthy subject’s data in barefoot condition ....................... 83
    6.2.6 Comparison among the possible materials for the design of the insoles ....................... 84
  6.3 Diabetic subject ............................................. 86
    6.3.1 Experimental vs simulated data in barefoot condition 87
    6.3.2 Experimental vs simulated data of all the phases of the
        stance phase in barefoot condition ....................... 89
    6.3.3 Experimental vs simulated data of the initial contact and
        the loading response phases in barefoot condition ........ 94
    6.3.4 Experimental vs simulated data in the frame of maximum
        medio-lateral force in barefoot condition ............... 96
CONTENTS

6.3.5 Experimental vs Simulated diabetic subject’s data in bare-foot condition ........................................ 97
6.3.6 Comparison among the possible materials for the design of the insoles ........................................ 98

6.4 Healthy vs diabetic subject ......................................................... 100
6.4.1 Experimental vs simulated data in barefoot condition ........................................ 100
6.4.2 Experimental vs simulated data of all the phases of the stance phase in barefoot condition ................ 101
6.4.3 Other tests ............................................................................. 101

Conclusions .................................................................................... 104

Bibliography .................................................................................... 105
Introduction

The work presented in this thesis has been done in the Laboratory of Bioengineering of the Movement of the University of Padua. The aim was to obtain two 2D Finite Element Models of the hindfoot for both a diabetic and an healthy subject in order to assess the consequences in terms of peak plantar pressures, of the application of forces during gait. The idea has been taken from the work of Goske et al. [33].

Two different Finite Element software have been tried but finally Abaqus/CAE has been used. A slice of the heel has been obtained from subjects’ foot MRI acquisition and the geometry has been constructed using Simpleware and then Rhino 4.0. Kinetic and kinematic data of the subject have been collected by means of a stereophotogrammetric system (BTS, Padova), two pressures platforms (Imagortesi, Piacenza) and two Bertec force plates (FP4060-10), temporally and spatially synchronized.

Compared to the literature a new element has been introduced: the subjects’ specifics forces and plantar pressures of each foot subsegment (hindfoot, midfoot and forefoot) have been used to perform the simulations of the FE models [6,53]. These forces have been applied to the model of the hindfoot exactly where and when they act.

This study is part of a larger project which aims to analyse the ability of different insole materials in decreasing the peak plantar pressure in diabetic subjects. A comparison between them has indeed been done.

The first chapter gives an overview of the foot, of its anatomy and its functions. Furthermore a definition of all the phases of the gait cycle is provided.

In the second chapter the diabetic disease is treated starting from the epidemiology to all its complications. In particular it has been emphasized the diabetic foot and its consequences.
The third chapter explains the motion analysis, focusing on the instruments used as the stereophotogrammetric systems and the force and pressure platforms, and on the protocols proposed in literature.

The fourth chapter is dedicated to the Finite Element Modelling, in particular on how to obtain a model and on the literature of the FE models of the foot.

The fifth chapter explains all the materials and method used to perform this work of thesis.

The sixth chapter finally shows the results obtained with this work.
Chapter 1

The Foot

1.1 Introduction [1]

The foot is an anatomical structure found in many vertebrates. It is the terminal portion of a limb which bears weight and allows locomotion. The human foot and ankle is a strong and complex mechanical structure containing more than 26 bones, 33 joints (20 of which are actively articulated), and more than a hundred muscles, tendons, and ligaments.

1.2 Anatomy of the foot [1]

The foot can be subdivided into three parts: the hindfoot or rearfoot, the midfoot, and the forefoot. The hindfoot is composed of the talus and the calcaneus. The two long bones of the lower leg, the tibia and fibula, are connected to the top of the talus to form the ankle. Connected to the talus at the subtalar joint, the calcaneus, the largest bone of the foot, is cushioned inferiorly by a layer of fat. The five irregular bones of the midfoot, the cuboid, navicular, and three cuneiform bones, form the arches of the foot which serves as a shock absorber. The midfoot is connected to the hind- and fore-foot by muscles and the plantar fascia. The forefoot is composed of five toes and the corresponding five proximal long bones forming the metatarsus. The bones of the toes are called phalanges: the big toe has two phalanges while the other four toes have three phalanges. The joints between the phalanges are called interphalangeal and those between the metatarsus and phalanges are called metatarsophalangeal.

A normal foot morphologically presents a dorsal convexity, while the lower part constituting the support of the foot, has a concavity more or less pronounced. The lower posterior part, consisting of the massive bone of the heel, gives insertion to the Achilles tendon and plantar fascia. The latter is an extensive and
strong tendon structure which integrity ensures protection for delicate bone, vascular and nerve structures of the foot and a correct tension of the plantar arch. The musculotendinous elasticity and innervation make the foot to change shape depending on the ground, activities and footwears.

As already underlined the foot is a complex of bones, ligaments, tendons and muscles.

The skeleton of the foot shown in the figure 1.1 is composed by:

- tibia;
- fibula;
- tarsus: talus, calcaneus, cuneiformes, cuboid, and navicular;
- metatarsus: first, second, third, fourth, and fifth metatarsal bone;
- phalanges.

![Figure 1.1: Structures and anatomical division of the foot. [14]](image)

There can be many sesamoid bones near the metatarsophalangeal joints, although they are only regularly present in the distal portion of the first metatarsal bone.
1.3. FUNCTION OF THE FOOT

The human foot has two longitudinal arches and a transverse arch maintained by the interlocking shapes of the foot bones, strong ligaments, and pulling muscles during activity. The arches of the foot are formed by the tarsal and metatarsal bones and, strengthened by ligaments and tendons, allow the foot to support the weight of the body in the erect posture with the least weight. The slight mobility of these arches when weight is applied to and removed from the foot makes walking and running more economical in terms of energy. The medial longitudinal arch curves above the ground. This arch stretches from the heel bone over the "keystone" ankle bone to the three medial metatarsals. In contrast, the lateral longitudinal arch is very low. With the cuboid serving as its keystone, it redistributes part of the weight to the calcaneus and the distal end of the fifth metatarsal. The two longitudinal arches serve as pillars for the transverse arch which run obliquely across the tarsometatarsal joints.

Finally the muscles acting on the foot can be classified into extrinsic muscles, those originating on the anterior or posterior aspect of the lower leg, and intrinsic muscles, originating on the dorsal or plantar aspects of the foot.

1.3 Function of the foot [2]

The main functions of the foot are its shock-absorbing capability during heel strike and its adaptation to the irregular surface of the ground during gait. During weight bearing activities, the plantar surface of the foot is exposed to ground reaction forces and consequently to deformation of the tissues. The relationship between force and deformation is expressed as stress-strain relations. When a person is standing, the magnitude of the GRFs is equal to body weight and each foot will experience approximately 50% of it, distributed over the plantar surface area. Obviously it will result in a predominantly vertically directed force. During walking however, for several reasons, the stresses applied to the feet are much higher. First, the amount of time in which both feet are simultaneously in contact with the floor is substantially reduced. Secondly, the rocker action of the ankle and heel allows different parts of the foot to come in contact with the floor during different phases of the stance phase: the support area changes in size and position while the ground reaction forces moves progressively from heel to hallux. Third, the ground reaction forces vary in magnitude, with one peak during heel landing and a second peak during push-off with the forefoot. Therefore, the heel and the forefoot experience much higher peak pressures than does the midfoot [2]. There are also ground reaction forces in the horizontal direction that cause a stress parallel to the foot skin. However, the magnitude
of the anterior-posterior-directed force and of the medio-lateral forces are much smaller than the vertical-directed force.

1.4 The gait cycle [4]

Gait is defined as a type of bipedal movement where a sequence of rhythmically repeated, alternating movements of the trunk, the upper limbs, the pelvis and the legs leads to forward locomotion. Gait is a complex activity that requires the work of the nervous system as a controller, of the muscles as motor generators which produce power and moments and of the bones as a system of levers which transmit motion. Disorders in any segment of these kinetic mechanisms that can restrict the normal joint range of motion, alter the normal lever morphology or interfere with the timing of muscle activation, will create change of the normal gait pattern. It is clear that the pathogenesis of gait disorders is multi-factorial and that the evaluation of gait disturbances implies a depth understanding of normal walking mechanisms [3].

To analyse the behaviour of the foot during walking the gait cycle is usually recorded and studied. The gait cycle consists of two basic components: the stance phase, during which the limb is in contact with the ground, and the swing phase, when the foot is in the air for the advancement of the limb.

![Figure 1.2: Dynamics of human gait [4]](image)

The stance phase may be subdivided into three separate phases as shown in figure 1.2:

1. *First double support*, when both feet are in contact with the ground;
2. *Single limb stance*, when the left foot is swinging through and only the right foot is in ground contact;

3. *Second double support*, when both feet are again in ground contact.

The nomenclature shown in the figure refers to the right side of the body; the same terminology would be applied to the left side, which for a normal person is half a cycle behind (or ahead of) the right side. In normal gait there is a natural symmetry between the left and right sides, but in pathological gait an asymmetrical pattern very often exists.

**Events** Traditionally the gait cycle has been divided into eight events or periods, five during stance phase and three during swing [5]. The names of these events are self-descriptive and are based on the movement of the foot, as seen in the figure below.

![Figure 1.3: Stance phases [5]](image)

The stance phase events, shown in figure 1.3, are as follows:

1. *Heel strike*, during which the foot through the heel touches the ground, and represents the point at which the body’s centre of gravity is at its lowest position.

2. *Foot-flat*, which is the time when the plantar surface of the foot touches the ground.

3. *Midstance*, that begins with opposite-side toe-off and full forefoot loading and terminates with heel lift, and represents the point at which the body’s centre of gravity is at its highest position.
4. *Heel-off*, occurs as the heel loses contact with the ground and pushoff is initiated via the triceps surae muscles, which plantar flex the ankle.

5. *Toe-off*, terminates the stance phase as the foot leaves the ground.

The swing phase events are as follows:

6. *Acceleration*, begins as soon as the foot leaves the ground and the subject activates the hip flexor muscles to accelerate the leg forward.

7. *Midswing*, occurs when the foot passes directly beneath the body, coincidental with midstance for the other foot.

8. *Deceleration*, describes the action of the muscles as they slow the leg and stabilize the foot in preparation for the next heel strike.

The traditional nomenclature is the best method to describe the gait of normal subjects but there are patients suffering several kind of pathologies whose gait cannot be described using this approach. For this reason an alternative nomenclature has been developed by Perry and her associates at Rancho Los Amigos Hospital in California. Here, too, there are eight events, but these are sufficiently general to be applied to any type of gait. Below there are the percentage of each phase during the gait time.

- Initial contact (0%)
- Loading response (0-10%)
- Midstance (10-30%)
- Terminal stance (30-50%)
- Preswing (50-60%)
- Initial Swing (60-70%)
- Midswing (70-85%)
- Terminal swing (85-100%)

Considering only the stance phase of the gait, other percentages could be calculated: initial contact (0-3%), loading response (3-17%), midstance (17-50%), terminal stance (50-83%), and pre-swing (83-100%) [6].
Chapter 2

The diabetic foot

2.1 Introduction [7]

Diabetes mellitus is a group of metabolic diseases in which a person has high blood sugar, either because the body does not produce enough insulin, or because cells do not respond to the insulin that is produced. This high blood sugar produces the classical symptoms of polyuria (frequent urination), polydipsia (increased thirst) and polyphagia (increased hunger).

There are three main types of diabetes mellitus:

- **Type 1 diabetes mellitus** results from the failure of the pancreas to produce insulin because of the destruction of the $\beta$-cellules of the islets of Langerhans which are the producer of that hormone. It requires the person to inject insulin and for this reason it is also referred to as insulin-dependent diabetes mellitus. Nevertheless given that it usually occurs in childhood or adolescence it is also called juvenile diabetes.

- **Type 2 diabetes mellitus** results from insulin resistance, a condition in which cells fail to use insulin properly, sometimes combined with an absolute insulin deficiency. Obesity is considered the main cause of it in individuals who are genetically predisposed to the disease. It is formerly referred to as non insulin-dependent diabetes mellitus or adult-onset diabetes.

- **Gestational diabetes** is when pregnant women, who have never had diabetes before, have a high blood glucose level during pregnancy. This event turns out to be quite transient and easily treatable but can cause problems for the infant ranging from an increased weight at birth until death of the unborn. For the mother it represents an important risk factor for developing type 2 diabetes mellitus.
Finally there are other forms of diabetes mellitus as for instance congenital diabetes. All forms of diabetes have been treatable since insulin became available in 1921, and type 2 diabetes may be controlled with medications but both types 1 and 2 are chronic conditions that usually cannot be cured.

### 2.2 Epidemiology [8]

Diabetes is rapidly emerging as a global health care problem that threatens to reach pandemic levels by 2030 [9]; the number of people with diabetes worldwide is projected to increase from 171 million in 2000 to 366 million by 2030 as it is shown in figure 2.1. This increase will be most noticeable in developing countries, where the number of people with diabetes is expected to increase from 84 million to 228 million.

![Figure 2.1: Millions of Cases of Diabetes in 2000 and Projections for 2030, with Projected Percent Changes [9].](image)

A recent Italian study has declared that 3 million Italian adults have diabetes, 4.9% of the total population. It is estimated that an additional million people have diabetes without knowing it and that 2.65 million people (6% of the population) suffer from impaired glucose tolerance (pre-diabetes). Furthermore in the future it is expected that nearly 5 million adult citizens will suffer from diabetes.

Diabetes mellitus is a disease known for its multifaceted complications like foot ulceration. The prevalence of foot ulcers ranges from 4% to 10% among persons diagnosed with diabetes. Foot ulcerations may lead to infections, lower extremity amputations and are major causes of disability to patients, often resulting in significant morbidity, extensive periods of hospitalization, and mortality. All these impairment lead to high costs for the SSN: it has been estimated that they
2.3. COMPLICATIONS

amount to 9% of the resources that means more than 9.22 billion of Euro a year. In general the cost of health care for an Italian diabetic patient is an average of 2600 Euro a year, more than double of the one for the citizens of the same age and sex, but without diabetes. It is not the treatment of the diabetic that expenses more but the healing of its complications, so the prevention is very important to cut costs.

2.3 Complications [10]

Diabetes can cause acute or chronic complications. Acute complications as for instance hypoglicemia are more frequent in type 1 diabetes and are related to the almost total lack of insulin. In these cases the patient may experience ketoacidotic coma, due to accumulation of products of altered metabolism, ketones, which cause loss of consciousness, dehydration and serious blood disorder. In type 2 diabetes acute complications are rare, but on the other hand chronic complications are very frequent affecting different organs and tissues, including eyes, kidneys, heart, blood vessels and peripheral nerves. Most chronic complications are:

- **Vasculopaty.** Peripheral vascular disease is an expression of a process of atherosclerosis which is localized in the lower limbs. In diabetic subject is one of the most frequent manifestations in terms of development of chronic complications due to poor glycemic control. Vascular disease also occurs in non-diabetic elderly men but in a patient with diabetes mellitus it has particular characteristics. First of all the frequency of occurrence is nearly similar in both diabetic male and female. The clinical picture has a classic debut in non-diabetics with pain that is localized to the calf when the subject after only a few meters of walk is forced to stop for cramps. In diabetic subject both legs are usually involved but the location is often the feet instead of the calves. This clinical situation if not attacked with specific therapies can worsen quickly to produce a state of total difficulty in walking also coming to the necessity of amputation of the affected limb, which is needed in only 1% of cases in non-diabetes patients but which is seven times more frequently in diabetes patients.

- **Retinopathy.** Diabetic retinopathy is a damage of the retina caused by complications of diabetes, which can eventually lead to blindness as it usually affects both eyes. It is the most common diabetic eye disease and it is caused by changes in the blood vessels of the retina. In some people blood vessels may swell and leak fluid in other instead abnormal new blood vessels
grow on the surface of the retina. The retina is the light-sensitive tissue at the back of the eye. A healthy retina is necessary for good vision. At first a subject suffering from the disease may not notice changes to his vision but over time, diabetic retinopathy can get worse and cause vision loss.

- **Nephropathy.** Diabetic nephropathy is a progressive reduction of the filter function of the kidney which, if untreated, can lead to renal failure until the need for dialysis or kidney transplant. It is the second most common cause of kidney failure. As the name suggests, the clinical picture is caused by the pathophysiological changes associated with diabetes.

- **Neuropathy.** Diabetic neuropathy is the most common chronic complication associated with Diabetes Mellitus, affecting 20-50\% of diabetic patients 10 years after their diagnosis [11]. It can cause numbness, pain of varying intensity and limb injuries, requiring amputation in severe cases. It can also lead to malfunctions of the heart, eyes, stomach and it is a major cause of male impotence.

- **Diabetic foot.** A diabetic foot is a foot that exhibits any pathology that results directly from diabetes mellitus. The most serious foot complications are the diabetic foot ulceration, the diabetic foot infections and the neuropathic osteoarthropathy of the foot.
2.4 The diabetic foot [10]

The diabetic foot, as stated above, is one of the most serious complications of diabetes mellitus as it compromises the function and structure of the foot. It may be the result of neuropathies or vasculopathies from which the name of neuropathic foot or ischemic foot and it also could develop the Charcot disease. The latter causes the fragmentation and destruction of the joint and the surrounding bones resulting in a severe foot deformity. Neuropathic and ischemic foot are deeply different, however in most subjects especially at advanced age, they coexist.

2.4.1 Neuropathic foot [8]

Diabetic neuropathy is defined as a damage to the peripheral nervous, somatic or vegetative system, merely attributable to diabetes. It may manifest with different clinical pictures but the most common form is the one that underlies the neuropathic foot and causes the impairment of all three components: sensory, motor and vegetative. Indeed there are three forms of diabetic neuropathy depending on which nerves are affected. If they are the sensory ones the perception is altered, if the motor ones the muscle balance is changed, finally if they are the autonomic nerves there could be a variation of the self-regulation of vegetative stimuli.

The sensory neuropathy affects the nerves that transmit feelings to the brain. The most serious consequence is the reduction of pain threshold that may have different levels of severity: some patients have feet less sensitive than the normal, others can endure surgery without anesthesia. The lack of stimulation of pain determines the inability to understand when something is hurting you in the foot. A neuropathic who wears too tight shoes will walk for a whole day without experiencing the formation of ulcers and blisters.

The motor neuropathy affects the nerves that innervate the muscles of the foot and are responsible for directing the commands of the brain to the muscles, thus leading the movement. When a nerve is injured, the muscle attached to it will suffer in turn manifesting hypotrophies or atrophies. Typically in diabetic patients with motor neuropathy, it creates a disparity between the extensor and flexor muscles that lead in an imbalance between the different tendons and their joints. In simpler words, when a muscle is "retract" because of it atrophies, the tendon of this muscle pulls back the joint on which it is inserted. All this leads to a deformation and an alteration of the bearing surface of the foot that will shrink to some particular points thus subjected to very high pressures.

Autonomic neuropathy is so called in regard to the damage of the autonomic
nerves and it results in the interruption of signals between the brain and parts of
the autonomic nervous system such as the heart, blood vessels and sweat glands,
resulting in poor performance or abnormalities of the same. It then determines
an alteration of the circulatory system in the foot by increasing cutaneous blood
flow and capillary permeability following the increase in hydrostatic pressure in
the microcirculation.

2.4.2 Ischemic foot

The poor blood supply (ischemia) is a consequence of the narrowing of blood
vessels that occurs in the ischemic diabetic foot due to plaque lipids and other
substances that accumulate in the vessel lumen, restricting the gauge. Peripheral
arterial disease is twice as common in persons with diabetes as in persons without
and is also a major risk factor for lower extremity amputation [12]. In diabetics
it is very frequent and early developed, it usually affects both legs and especially
the arteries below the knee. The latter is the most important characteristic for
the purposes of the treatment as the arteries of the leg and foot have a smaller
calibre than those of the thigh and is thus more difficult to act therapeutically
on them. Furthermore diabetics’ arteries are often calcified with a prevalence of
multiple occlusions. When circulation is compromised, the foot is less able to
react to situations such as cold, infections and trauma, and is more susceptible to
dry skin, and neuropathy affecting the shape. Typical of the diabetic disease is
the lack of the earliest symptom of peripheral arterial disease the “claudicatio” or
pain that occurs in the calf or thigh after a certain number of steps. It depends
on the fact that the arteries of the leg receiving less blood than necessary, fail to
increase blood flow essential during the physical effort of the deambulation. The
absence of claudicatio in diabetic exist for the concomitant presence of sensory
neuropathy: the pain will be muted or even absent and the patient will be unaware
of suffering peripheral arterial disease in the legs. This is one of the reasons of
why the diagnosis of peripheral arterial disease in diabetics is not easy. There is
a real risk that its first manifestation is a non-healing ulcer or in the most severe
cases a gangrene of the tissue. The figure 2.2 summarizes the differences between
neuropathic and ischemic disease.

2.5 Plantar ulcers and plantar pressures

From the above it is clear that diabetic patients suffer of changes in foot structure.
These can lead to high plantar foot pressure, which is an important predictive
2.5. PLANTAR ULCERS AND PLANTAR PRESSURES

risk factor for the development of diabetic foot ulceration that once opened can significantly increase the risk of infection [2].

These high values of pressures usually occur at sites with bony prominence as the first, second, or third metatarsal head, mainly as a result of excessive and repetitive pressure applied to the foot during walking. They are also associated with the plantar tissue thickness and the amount of cushioning (soft tissue) available. The fat pad of the human heel indeed functions to absorb shock and it provides protection against excessive local stress. However, its behaviour is affected by its material properties, its shape, and its thickness [18]. For instance, the diabetic patients often shown qualitative changes of the plantar fat due to a fibrotic process beneath the metatarsal.

Furthermore, high pressure is associated with the increased hardness of the skin and of the plantar fascia common in diabetic patients and with the deformation of the foot. It is clear that foot ulcers in diabetes result from multiple pathophysiological mechanisms, including roles for peripheral neuropathy, peripheral vascular disease, foot deformity, increased foot pressures, and diabetes severity.

2.5.1 Plantar pressure measurement

The measure of the plantar pressure is important to prevent the breakdown of the plantar tissues that can be initiated by three main mechanisms:

1. increased duration of pressures which includes application of relatively low pressures for a long period of time resulting in ischemia. If the latter is prolonged, it leads to cell death and injury. As shown in figure 2.3 there is an inverse relationship between time and pressure that means that in several days pressure ulcers can also occur at a very low level;

2. increased magnitude of pressures which comprises the application of high
pressures in a short period of time. This only occurs if a large force is applied to a relatively small area of skin;

3. increased number of pressure which leads to failure of the biological structure for the continuous repetition of loads.

Anyway given that different plantar pressure measurement devices give data that are not always directly comparable, there is no universal threshold set for the ulceration but normal pressure ranges has to be obtained each time that a new system is used. There are significant limitations in using a single critical level of surface plantar foot pressure to identify patients at risk for neuropathic foot ulceration because different areas experience different pressures and might have different thresholds. Recent studies have suggested that peak plantar pressure may only be 65% specific for the development of ulcers [19]. These limitations are at least partially due to surface pressures not being representative of the complex mechanical stresses that developed inside the subcutaneous plantar soft-tissue, which are more likely to be the cause of tissue breakdown.

![Figure 2.3: Inverse relationship between force (pressure) and time (or repetition). As force (pressure) increases, the duration (time) or number (repetition) of force(s) required to cause tissue injury decreases [2].](image)

Anyway to cause tissue breakdown, trauma is needed in addition to neuropathy and vascular diseases [2]. A trauma may be intrinsic, due for example to repeated stress from high pressure or calluses, or extrinsic, due for instance to the rubbing on the skin of an object inside the shoe. The body responds to high pressures or to repeated micro-trauma with callus formation to protect the skin from further damage. However if the callus becomes excessive it also contribute to an increase in pressure and must therefore be removed at regular intervals.
Chapter 3
Motion analysis [15]

3.1 Introduction

Several studies have highlighted that biomechanical factors play a crucial role in the aetiology, treatment and prevention of diabetic foot ulcers. Accordingly to this a it is indispensable to understand the biomechanics of the normal foot before any treatment could be applied on it. To evaluate the conditions of a diabetic patient a study of his gait is then usually performed.

Motion analysis starts in the days of Aristotele (384-322 BCE) who can be considered the first to study human and animal locomotion in his work ”On animal locomotion”. It was not until the renaissance that further progress was made through the experiments and theorising of Giovanni Borelli (1608-1679) an Italian mathematician, doctor and physiologist who created models of human and animal bodies in the form of a lever system: by studying forces acting during motion he introduced the methodology of contemporary biomechanics. After him, the French E.J. Marey (1870) and the American E. Muybridge (1880) should be mention because they performed gait assessment using multiple camera technology. In 1945 Inmann and his fellows introduced the modern gait analysis starting the instrumented data collection and the application of biomechanical principles for gait evaluation. In the following years the contributions of Saunders et al (1953), Sutherland (1988), J. Perry (1992), J. Gage (1991) and others have increased our understanding on gait mechanisms and established modern gait terminology [16].

The study of the movement requires several measurements because both kinematic and dynamic variables are needed and should be integrated once obtained. The first ones are position, velocity and acceleration while the second ones regards the forces. For instance to perform a gait analysis, the kinematic data are
integrated with the measures obtained from force and pressure plates; to obtain a more complete study also other signals can be acquired as the electromyographic signals.

3.1.1 Stereophotogrammetric systems

There are several kinds of commercial motion capture systems with heterogeneous technologies and different degrees of invasiveness used to collect kinematic data. Electrogoniometers and accelerometers are examples of this instruments, which give direct measurements of the variables of interest. Anyway given that they are electrical devices in contact with the surface of the body, they can reduce the naturalness of the movement. Another example are the electromagnetic systems which provide direct kinematics measurements by an external magnetic field generator. Anyway the accuracy of their measure has a high sensitivity to the presence of ferromagnetic objects. There are also systems based on acoustic sensors which provide an indirect estimation of kinematic variables given the speed of sound in air. They assure good localization accuracy, but they also suffer from interference and inconstancy of speed of sound and echo.

Most commonly used systems for the estimation of human movement are the optoelectronic systems that use cameras operating in the range of the visible or near infrared. They ensure high accuracy even though they can provide only an estimate and not the direct measure of the kinematic variables. They take advantage of the geometric characteristics of the surface body which images, acquired by several cameras, are reconstructed in 3D. It is possible for instance, to analyse the time evolution of the position of the edges of the anatomical segments which move in the space. These are recognisable because they generate a gradient of light intensity in the image that can be detected by segmentation algorithms. The intrinsic limit of this process is the impossibility to measure the velocity of an edge in movement in a direction non orthogonal to the edge itself. A possible solution is the analysis of the vertex of the edges which allow to measure the movement of the object in all the direction. Anyway the vertex of the human body are not well marked and they can move during the acquisition so the accuracy of their determination is low. Even though actual researches are focusing on the develop of markerless technique, so far the the cooperative marker are the most used solution. Physical marker can be active or passive and they are easy recognizable in the images. Passive markers are small balls of plastic covered with reflector film. In contrast to the active ones they do not have a led that generate light thus they need an additional lighting device with specific wavelengths (780-820 nm). Their sphericity guarantees the best reflection of in-
3.1. **INTRODUCTION**

Infrared rays and with cameras provided of an an appropriate optical filter, they are immediately recognizable from the background. The active ones instead are LED (Light Emitting Diodes) and so they don’t need a device of external illumination but rather a power supply. They should be synchronized by cable but they do not need a sophisticate pre-elaboration as the passive ones to be identified and classified. Compared to the passive markers they have lower emission angles and so the setup of the camera is critical. The use of cameras with infrared emitters together with passive markers is called stereophotogrammetry. Figure 3.1 shows the Laboratory of Bioengineering of the Movement of the University of Padua with the infrared camera for the stereophogrammetric acquisition, and a patients with the markers attached on his limbs during an analysis.

![Figure 3.1: At left a picture of the laboratory with the stereophotogrammetric system and the pressure and force plates, at right a patient during an acquisition.](image)

Movements are thus acquired from a stereophotogrammetric system and the data are processed by an elaboration software that can detect markers. The result is a sequence of images.

Once the image is acquired it is subjected to a sequence of elaboration.

**Thresolding**

In the first step of elaboration of the image, it is the filtered with different gradients of illumination to improve the accuracy of the algorithm of thresholding. After that the histogram of the pixel that belong to the background and to the characteristic of interest is analysed to defined the best thresold to separate the markers. Given that a pixel can belong only to a two classes, marker or background, the problem can be treated by means of a statistical tests of hypotheses approximating the intensity histogram of gray with two gaussians. The threshold value is determined by means of a minimization algorithm that minimize the probability to recognize a pixel which belong to the background as one that belongs to a marker and vice versa the probability to recognize a pixel which belong
to a marker as a pixel that belongs to the background. After this operation the centre of the marker is estimated.

The simpler solution consists of the calculation of the barycentre of the pixel over threshold which belongs to the single marker weighted with an unitary value that means without consider the gray levels. A best solution makes use of the circle fitting that is based on the calculation of the centre of an hypothetical circle passing through the pixel on the edge of the image of the marker. This method gives wrong values of the centre of the marker in presence of distortions and occlusions.

**Blob Analysis**

A flexible approx is given by the Blob Analysis, which aims to extract from the images the characteristics of interest eliminating false measures mainly due to reflections. It checks the size of the blob to eliminate the problem of reflexes and the shape of the blob. To classify the markers the a priori information on their shape and dimensions are used. The projection of the marker on the image plane of cameras occupies an area of ellipsoidal shape of a certain number of pixels, function of the position. The ellipsoid that approximates the marker can be estimated from the experimental covariance matrix of the positions of the pixels occupied from the image of the marker with respect to its barycentre. From the analysis of the main components of the matrix a shape factor of the blob is estimated and it must be compatible with the projection of a sphere.

**Correlation**

An other methodology to calculate the centre of a marker is the cross-correlation. In few words a mask (kernel) of shape and size equal to the shape and size expected from the image of the markers is created and then compared to the real images obtained during the acquisition. Once the real marker are identified the barycentre is calculated, weighted by the value of the correlation. This approx ensures an high signal to noise ratio given that big luminous area or shape different from the kernel are transformed in areas with negative or zero value.

**Calibration**

Before starting the acquisition the cameras have to be calibrated so their geometrical parameters have to be settled. The reference system of the laboratory has to be determined and also the internal parameters of the cameras as the optic has to be fixed.

The calibration start with the setting of the cameras: the operator has to control if a subject that walk on the force and pressure platforms is inside the volume of
acquisition until the end of the walk: usually two steps should be acquired. Position, zoom and focus of the cameras can be change in order to obtain this results. After this, it should be controlled if all the cameras see the markers. Two axis of reference consisting of two chopsticks with several markers, are posed on the edges of the force platforms. The ability to see the markers is checked. Finally the operator has to move a chopstick with markers in the volume of interest during an acquisition of around 5000 frames, with horizontal and vertical movements. This is done in order to define the volume of acquisition desired.

Each time that a new acquisition is performed, an initial checking phase is done in order to control if all the markers are visible by the cameras. After these steps the system is able to reconstruct the position of a point in the 3D space.

**Tracking**

Once the system is calibrated it is possible to reconstruct the position of the markers in the space for each instant of time. The next problem is the tracking that is the determination of the trajectories of the markers during time. The information that can be used to solve the problem are essentially two: regularity of the trajectory and a priori information about the shape and type of movement of the subject. Usually models that visualized the normal position of the marker placed on the body are used: the operator who tracks all the acquisition defines the name of a marker checking its position on the model. The use of pre-defined model gives the possibility to predict the self occlusions of the markers as shown in figure 3.2.

![Figure 3.2: Self occlusions. Marker m is visible by the camera because the angle α is less than 90°. On the contrary marker n is not visible because the angle β is obtuse. Markers o and p should be both visible, but o is subject to self occlusion caused by the left arm [15].](image)
When the tracking is terminated, data can be exported to be elaborated and processed.

3.1.2 Force platforms

Force platforms are measuring instruments that detect the ground reaction forces generated by a body standing on or moving across them, to quantify balance, gait and other parameters of biomechanics. Measures are given according to the XYZ reference system associated with the platform and usually provided by the constructor. The simplest force plates measure only the vertical component of the force in the geometric centre of the platform. These are called three-component platforms because they allow to calculate, using the system of equations shown in figure 3.3, only the vertical force and the two horizontal moments.

$$
\begin{align*}
[F] = & \begin{bmatrix}
  k_{11} & k_{12} & k_{13} & k_{14} \\
  k_{21} & k_{22} & k_{23} & k_{24} \\
  k_{31} & k_{32} & k_{33} & k_{34}
\end{bmatrix}
\begin{bmatrix}
  f_1 \\
  f_2 \\
  f_3 \\
  f_4
\end{bmatrix}
\end{align*}
$$

Figure 3.3: In the figure a three-component platform with its system of equation is shown [15].

More advanced models measure the three-dimensional components of the single equivalent force applied to the surface and its point of application, usually called the centre of pressure, as well as the vertical moment of force. These are called six-component platforms. They do not disturb the free expression of the motor action and are employed for the study of locomotion (dynamic analysis reverse) and posture. Force platforms may be classified as single-pedestal or multi-pedestal and by the type of the force and moment transducer if strain gauge or piezoelectric sensors. Single pedestal models are suitable for forces that are applied over a small area but for studies of movements, such as gait analysis, force platforms with at least three pedestals are used to permit forces that migrate across the plate. Ideally the force transducer should have a linear response without phase distortion in the whole field of change of amplitude and speed of interest. The frequency response of a transducer depends on the type of sensing element, on its geometrical dimensions and on the influence the electronic signal. Generally in the design of the sensitive element, a compromise between flexibility and rigidity should be
3.1. INTRODUCTION

set: they are directly proportional respectively to the sensitivity and to the maximum frequency response. The frequency response is a parameter provided by the manufacturers, but it is easily measurable subjecting the transducer to a pulse of force. It is indicative of the speed of response and of the maximum frequency at which the transducer can be used.

Important static characteristics are the linearity of the response and the hysteresis. Usually the non-linearity is considered by the determination of the maximum deviation of the response to the straight line that best approximates it. If there is not hysteresis the non-linearity can be corrected during calibration. The hysteresis describe the dependence of the response of the transducer from its previous history of loading: for a force transducer it is define the maximum difference in response to two forces of equal intensity during a loading-unloading cycle. The hysteresis limits the accuracy of the measure, so its value is important to judge the quality of a measure instrument. The most common technologies used for the realization of force transducers are based on strain gauge or piezoelectric crystals.

**Force transducer**

A strain gauge enables the conversion of a variation of the length in an electrical signal by means of the variation of its electrical resistance. The sensitivity is maximum according to a main direction, which is that of prevalent development in length of the resistive element. The variation in resistance of a strain gauge divided for its total value is proportional to its deformation. Normally four strain gauge are used for each measurement channel configuring a Wheatstone bridge. This is the configuration of maximum sensitivity and it allows the minimization of thermal effects. The strain gauge must have the same nominal resistance to balance the bridge and to have the best sensitivity on the four branches. Furthermore the combination of the strain gauges must be such that the transducer is sensitive only to a component of the force.

The piezoelectric sensors take advantage of the piezoelectric effect which manifests itself in some crystals: when subjected to a mechanical deformation they generate a potential difference. Quartz is often used for the realization of this transducer because in addition to having this property, they are very stable. Piezoelectric transducer are not suitable for measuring static loads, because the electric charge disperses over time. In the commercially available transducer crystals are cut into disks: depending on their orientation with respect to the crystal structure, elements sensible to the force normal to the surface or to the tangential force can be obtained. Often to increase the sensitivity of the transducer, pair of discs are used.
Strain gauges have a frequency response lower than the piezoelectric crystals but they are cheaper and they can be used for the posturgraphic examination.

An ideal characteristic of these sensors would be the not-sensitivity to components of the load different from the desired one. Generally a measuring channel for a component of the load may show a certain sensibility also to other components and this is known as cross-talk of the latter on the desired one. Crosstalk can be caused by dimensional inaccuracies and tolerances in the orientation of strain gauges and in their sensitivity. If the transducer has been designed to measure all the components of the load that it can transmit, the crosstalk does not pose particular problems and it can be compensated during the calibration. Anyway it can not be compensated if relating to a component of the load that is not measured by the transducer.

3.1.3 Calibration of force platforms

The force plates are precision instruments whose response can, however, deteriorate with time. To avoid this problem, periodic calibration and testing protocols must be performed. Recently, two prototypes of calibration devices, an active type, oriented to six-component platforms, and a passive type for three-component platforms, have been developed.

Active device for six-component platforms

The test device shown in figure 3.4, consist of a three-feet platforms, on of which act on the force plate that has to be controlled. This foot is able to apply a variable force over time that has vertical and horizontal components measured by sensitive load cells. The vertical component, $F_v$, is generated by standing on the device and varying the distance of the center of mass from the unique support. The horizontal component $F_h$, is provided by a linear motor and transmitted to the support by means of a spring and a load cell.

The protocol requires the accurate positioning of the support in five known points of the platform. This allows to apply known moments. In each point, two test are done, orienting the horizontal component of the force along the direction of the two horizontal axes. Finally an algorithm estimates the calibration matrix with least squares approach on data collected in the trials. It allows to correct the error of alignment between the horizontal component of force applied in each
3.1. INTRODUCTION

Figure 3.4: At left the active test device for six-component platforms is shown. At right the procedure of calibration [15].

test and the corresponding axis of the force platform.

This method of calibration is suitable for the periodic verification of the platforms even in clinical environment, because it allows the manual and fast execution of the protocol also by non-technical staff.

Passive device for three-component platforms

The calibration device consists of a ballast with mass $M$ and of a smallest eccentric mass $m$ that rotates with respect to a vertical axis aligned with the axis of symmetry of the ballast. It is shown in figure 3.5.

Figure 3.5: The figure shows at left the passive test device for three-component platforms and at right the spirals before and after the calibration [15].

The position of the centre of pressures depends on $M$, $m$, the instantaneous angular speed $\omega(t)$ of the rotating mass and on geometrical parameters of the device. The trajectory of the center of pressure, once the rotating mass has been in motion manually, can be determined with great precision and it is described by a spiral. A visual check can be made at once: any anisotropies are evident symptoms of loss of calibration. More precisely, the eigenvalues of a covariance matrix are calculated and from their ratio and index of non-calibration is derived.

3.1.4 Pressure Platforms

During locomotion forces between the human body and the ground are spread over various structures. It is important to know how loads are distributed for a lot
of application as the evaluation of the pressures in patients suffering from diabetic foot. The force measured by force plates does not give this information and so different low-cost technologies have been developed. To be able to detect the real disposition of loads, multiple sensors that measure only the vertical component of the force are used. Once the area of the sensor is known, measuring the force it is possible to calculate the pressure on it. Systems realized in this way provide measurement of the pressure distribution on a bearing surface with a detail that can be established in the construction phase according to requirements. Typically the pressure sensors are arranged in arrays to form platforms of pressure. The characteristics to take into account about pressure platforms are:

- spatial resolution, that is the distance between the centres of two adjacent sensors. It is important for the suitability to provide the information below smaller structures, such as the metatarsal heads;

- area of the sensor. Ideally, sensors should be point. In the reality the actual pressure is measured as the average force / area. The resulting error is greater the greater the surface is;

- Frequency of sampling, that should be the highest possible but usually meet its limit in the great quantity of data to treat. For instance a frequency of 50 sample/s for a matrix of 2000 elements results in a flow of 100000 data/s.

- Type of sensors, if piezoelectric, resistive or capacitive.

**Sensors of pressure**

The technologies of transduction adopted by the sensors of pressure are based on resistive, capacitive or piezoelectric principles.

The capacitive sensors, as shown in figure 3.6, are made putting two metallized plates on both sides of a dielectric and elastic material; when a force is applied, the elastic material deforms resulting in a diminution of the distance between plates and in an increase of the capacity. The elastic material is chosen so that the ratio between the dielectric constant and the thickness is a linear function of the global force applied on all the surface of the sensor.

The resistive sensors are made putting two metallized plates in contact with a conductive polymer as shown in figure 3.6. While the compression force applied on the plates increase the electrical resistance decreases mainly for contact
3.1. **INTRODUCTION**

phenomena. There are two possible constructive schemes: the two plates can be in contact on the opposite surfaces of a polymer conductive sheet or on the same side of the polymer material. In the last case the plates are generally very close and the small distance (< 1 mm) is critical for the response of the sensor. The polymer can be a sheet or just ink deposited on an insulating support.

![Construction scheme of the main pressure sensors](image)

**Figure 3.6:** Construction scheme of the main pressure sensors where F is the force applied and d the dielectric material. [15]

Piezoelectric sensors does not yet have a significant commercial deployment for the use in pressure platforms.

**Response of the sensors and main error sources**

Pressure sensors are not very accurate and the inaccuracy mostly depends on non-linearity and hysteresis. The most accurate sensors available are the capacitive sensors, indeed they have an hysteresis of 3% and an accuracy of ±5% full scale. In terms of dynamic response, it is important to know what is the maximum rate of change of force or pressure that sensors are able to capture. Usually the phenomena under study have a frequency limited to 50 Hz and all the sensors are capable to respond to this need. The elastomers suffer the phenomenon of creep that is a gradual deformation which lasts for a long time. This results in a typical drift of the response and so the use of sensors based on elastomers is not recommended in cases where the load does not return to zero for long times as the posturographic examinations.

To evaluate what should be the necessary spatial resolution, usually what happen under the metatarsal heads is studied. This because the region is often overloaded as for instance in the diabetic foot. In order to record the pressures in that zone,
the distance between the centre of adjacent sensors, should not be more than 3-5 mm: this requirement is not fulfilled by a lot of commercial pressure systems. Finally another important aspect regards the area of the sensors that, as already said, should ideally be a point. The real sensors indeed measure a value of the pressure averaged on the sensible surface. Anyway it is not possible to construct point sensors because of the electrical principles under their functionality so when choosing a pressure platform, it should be take into account that the one with the smaller sensors is always the best.

3.1.5 Pressure Insoles

Sometimes when there is the need to record the patients’ feet pressures in normal conditions such as using their own shoes and walking on the real ground, pressure insoles are used. They can be resistive or capacitive insoles and are introduced into the shoes of the subject. The subject wears a belt in which the recording instrument and the battery are placed. Some cables provide the connection among all the elements. The data collected can be sent to the software of analysis via Bluetooth or other technologies. Unfortunately pressure insoles cannot be considered as substitute for pressure platforms due to their less accuracy. The major causes of these inaccuracy are:

- preloads and not real loads variable even in a synchronous manner with the gait cycle, due to the inevitable deformation for adaptation to the inner surface of the shoe;
- possible alteration of the value of pressure due to the damping effect of the insole, if it is made of materials with viscoelastic characteristics;
- lack of knowledge and possible change during the movement, of the position of the individual sensors with respect to the anatomic sites of interest. To overcome this problem insoles with a high number of small sensors can be used, but these usually have a limited temporal resolution and they generate a great amount of data that limits the portability of the equipment.

All the pressure measurement devices consist of a number of force sensor elements and the variation in size of these elements is a delicate parameter as already attested: it has heavy consequences for the calculation of pressure since a focal area of pressure under the foot will appear to have a lower value on a device with a larger element size. This means that different plantar pressure measurement devices give data that are not always directly comparable and so there is no universal threshold but normal pressure ranges has to be obtained each time that a
3.2 Protocols

As already said in the introduction 3.1, plantar pressure, ground reaction force and body-segment kinematics measurements are largely used in gait analysis to characterise normal and abnormal function of the human foot. The combination of all these data together provides a more exhaustive and detailed view of foot loading during activities than traditional measurement systems alone do. Diabetic patients usually exhibit a displacement of the fulcrum of the step from the tibio-tarsal to the coxofemoral joint and posture modifications [45]. A further limited joint mobility is caused by additional alterations of the soft tissues, tendons and ligaments and it occurs mainly to 1st metatarsophalangeal and subtalar joints [46]. Especially the plantar fascia behaves like one rigid lever during the step and this results in a reduction of the adaptability to the ground [47, 48]. This particularity should request special studies of of structure and function of the diabetic foot but instead they have received little attention in the literature, while most of the studies have concentrated on the kinetic analysis by means of force and plantar pressure plates. On the other hand kinematic analysis would be clinically very important for diabetic neuropathic patients in order to appreciate the supination-pronation and inversion-eversion movement of forefoot vs midfoot and hindfoot.

Unfortunately, there are still no many movement analysis protocols suitable for this purpose because the majority utilize rigid mounting plates by means of elastic bandages and lengthy anatomical calibration procedures that cannot be easily applied in patients with peripheral artery disease or neuropathies [7, 49, 50].

A work presented in literature suitable for this intent is the protocol of Giacomozzi et al. [52]. They proposed a study that aimed to design and test an integrated use of different devices for a more comprehensive description of the loading characteristics of the foot sole. More specifically, the authors synchronised a 3D anatomical foot tracking system with a piezo-dynamometric platform. An anatomical landmark projection on the footprint was exploited to perform an anatomically based selection of the subareas of interest and, in turn, to estimate the relevant local forces and moments. Their Pressure-Force-Kinematic measuring system was installed at the Movement Analysis Laboratory of the Istituti Ortopedici Rizzoli (Bologna, Italy). A pressure platform made of a matrix of $81 \times 121$ resistive sensors, 5 mm spaced in
both directions, was incorporated in the piezo-dynamometric platform and data were transferred to a personal computer (PC) through a dedicated board. The pressure platform was rigidly fastened on top of a commercial $0.4 \times 0.6m$ force platform. The ELITE stereophotogrammetric system was used to track foot bone positions. Two cameras were arranged in a line almost parallel to the line of gait progression, both seeing all the reflective markers during the stance phase. The markers were plastic hemispheres, 6 mm in diameter, covered with retro-reflective tape. The system is shown in figure 3.7.

![Figure 3.7: Diagram of experimental setup, with three instruments for routine integrated gait tests' [52.]](image)

The synchronisation of all the instruments was obtained by means of a trigger signal from the pressure platform PC to the ELITE PC, which activated both the Kistler force plate and the stereophotogrammetric system. They performed various test to certify the improved features of this integrated system and finally they attested that the integration of the pressure and force platforms delivered the complete characterisation of foot subareas and that the integration of the two platforms with the 3D anatomical tracking system made it possible to link the loading distribution under the foot and the position of the overlying anatomical structure [52].

Another work that use a similar protocol has been the one of Mac Williams et al. [55]. Nineteen 10 mm diameter reflective markers were used to identify eight segments of the foot plus the shank. Six cameras were used to record the spatial positions of the markers throughout stance phase. In this work they collected two separate sets of data. In the first set, subjects walked over a pedobarograph and so foot pressure and kinematic data were simultaneously collected. In the second set, subjects walked over a force plate and so ground reaction force and
3.2. PROTOCOLS

kinematic data were simultaneously collected.
Finally another similar work has been presented by Stebbins et al. in 2005. Also in this case they integrated three different source of data, based of the work of Giacomozzi, in order to assess foot pressure measurement in healthy children, using an automatic technique of sub-area definition measurement. They made use of a stereophotogrammetric system with 12 camera to collect the 3D kinematic data, and of a piezo-resistive pressure platform with a spatial resolution of 5 mm, sampling at 100 Hz. This was rigidly mounted to and time synchronised with an AMTI force plate, with a minimum sampling frequency of 500 Hz.

3.2.1 Protocol used in this work

The protocol used in the Laboratory of Bioengineering of the Movement of the University of Padua is born with the aim to combine 3D motion, ground reaction forces, and plantar pressures in order to obtain the simultaneous assessment of kinematics, kinetics, and plantar pressures on foot subareas of diabetic subjects. This is achieved by means of commercially available systems [53].

Movement analysis is carried on using a 60 Hz 6 cameras stereofotogrammetric system (BTS S.r.l, Padova), 2 force plates (FP4060-10, Bertec Corporation,USA), 2 PP systems (410 × 410 × 0.5mm, 0.64cm² resolution, 150Hz, Imagortesi, Piacenza). The stereophotogrammetric system is used either to perform the automatic footprint subareas subdivision or to compute the 3D foot subsegment kinematics.

The signals coming from all systems are temporally and spatially synchronized in post processing thus avoiding the need of modifying either the hardware or the software of the employed systems. The temporal synchronization between force plates and motion capture system is provided by the motion capture company. The temporal synchronization of pressure and force plates instead is achieved normalizing the output of them with respect to stance phase of gait. Therefore the sample frequency of the systems has not to be modified and the method can be transferred to different bands of pressure and force plate systems.

With regard to the spatial synchronization each pressure plate system is mounted onto each force plates by means of double-sided tape and the spatial alignment of the two platforms is assured by comparing the two center of pressures.

3D foot kinematics and plantar foot subarea definition

A 3D kinematic model already established in this laboratory is used for the subsegment angles estimation during gait [6]. It requires that the skin markers are attached through double sided tape on the anatomical landmarks described in
Figure 3.8: The model anatomical landmarks identified on a skeleton foot (black circle), the projection of them on the footprint in order to obtain the sub-area division, the kinematics model definition [53].

Table 3.1 and shown in figure 3.8.

<table>
<thead>
<tr>
<th>Anatomical Landmark</th>
<th>Segment</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>HF = Head of the Fibula</td>
<td>Tibia = tibia-fibula</td>
<td>Proximal tip of the head of the fibula.</td>
</tr>
<tr>
<td>TT = Tibial Tuberosity</td>
<td></td>
<td>The most anterior border of the proximal extremity of tibial tuberosity.</td>
</tr>
<tr>
<td>LM = Lateral Malleolus</td>
<td>Hindfoot = calcaneus and astragalus</td>
<td>The lateral apex of the external malleolus.</td>
</tr>
<tr>
<td>MM = Medial Malleolus</td>
<td></td>
<td>Medial apex of the internal malleolus.</td>
</tr>
<tr>
<td>ca = Calcaneus</td>
<td></td>
<td>Lower ridge of the calcaneus posterior surface.</td>
</tr>
<tr>
<td>PT = Peroneal Tubercle</td>
<td></td>
<td>Sitting with unloaded foot placed at 90° with respect to the sagittal axis of the fibula. Following the prolongation of inferior apex of the lateral malleolus, aligned with the longitudinal axis of the tibia, place the marker on the first bone prominence below the lateral malleolus.</td>
</tr>
<tr>
<td>ST = Sustentaculum Talii</td>
<td></td>
<td>Sitting with unloaded foot placed at 90° with respect to the sagittal axis of the fibula. Following the prolongation of inferior apex of the medial malleolus, aligned with the longitudinal axis of the tibia, place the marker 2 cm under the distal border of the lateral malleolus: in correspondence of the last medial bone prominence before the medial muscle-tendon insertion of the calcaneus.</td>
</tr>
<tr>
<td>NT = Navicular Tuberosity</td>
<td>Midfoot = scaphoid, cuboid, 1st, 2nd, 3rd cuneiform, 1st, 2nd, 3rd metatarsus</td>
<td>Sitting with his unloaded foot placed at 90° with respect to the sagittal axis of the fibula. Ask the subject to relax the foot and find the proximal epiphysis of the 1st metatarsal. Following the line between the proximal epiphysis of the 1st metatarsal and the lower ridge of the calcaneus the first prominence that you palpate is the cuneiform and the second is the navicular. Once found the navicular bone on that line place the marker on the navicular following the line orthogonal to the floor on the interior side of the extensor longus of the allux (ask the subject to rise the allux to find the extensor longus).</td>
</tr>
<tr>
<td>C = Cuboid</td>
<td></td>
<td>Sitting with his unloaded foot placed at 90° with respect to the sagittal axis of the fibula. In correspondence of the proximal aspect of the 5th metatarsal base following the direction of the tibia axis (orthogonal to the floor) place the marker on the first bone prominence you palpate on the cuboid. The most external surface of the base of the fifth metatarsus.</td>
</tr>
<tr>
<td>VMH = Fifth Metatarsal Heads</td>
<td>Forefoot = 1st, 2nd, 3rd, 4th, 5th metatarsal heads and phalanxes</td>
<td>Head of the 1st metatarsus.</td>
</tr>
<tr>
<td>IMH = First Metatarsal Heads</td>
<td></td>
<td>Head of the 1st metatarsus.</td>
</tr>
<tr>
<td>VMB = Fifth Metatarsal Base</td>
<td></td>
<td>Head of the 1st metatarsal.</td>
</tr>
<tr>
<td>IIT = Proximal epiphysis of second toe phalanx</td>
<td></td>
<td>Choose the 2nd ray with the left hand, and with the right hand move the proximal phalanx of the second toe in dorsiflexion and plantarflexion; place 1 cm distal from the joint interstice.</td>
</tr>
</tbody>
</table>
The foot and ankle complex are divided into sub-segments. A three segment model for the plantar sub-area definition is obtained by means of projecting the anatomical landmarks of the kinematics protocol onto the footprint. The following foot subareas are defined:

- **hindfoot**: the area between the line connecting both the vertical projection of the sustentaculum tali and the throclea peronealis and the vertical projection of calcaneus;

- **midfoot**: the area between the anterior reference line of the hindfoot and the line connecting the vertical projection of the first and fifth metatarsal head;

- **forefoot**: the area between the anterior reference line of the midfoot and the end of the anterior border of the footprint. The plantar surface is compartmentalized so that sensors did not overlap across segments.

Relevant anatomical bone embedded frames are defined for each segment and sub-segment as described in table 3.2 following international conventions.

Furthermore for the kinematic assessment, the following model segments and joints relative motion are considered: motion of the ankle joint as complete foot vs. tibia, motion of the hindfoot vs. tibia, motion of the midfoot vs. hindfoot, motion of the forefoot vs. midfoot. Dorsi-planatarflexion (D/P) motion is considered as the distal segment rotation around the mediolateral axis of the proximal one, inversion-eversion (I/E) angle as the distal segment rotation around its anteroposterior axis, internal-external (Int/Ext) rotation as the segment rotation around the axis obtained as cross product between the other two axis. Model segments and joints rotation angles are calculated as described in table 3.2 according to Cardan convention.
SEGMENT | AXES | JOINT | COORDINATE SYSTEM
--- | --- | --- | ---
Tibia | y | The two malleoli and the head of fibula define a quasi frontal plane, the y axis is parallel to the line connecting the midpoint between LM and MM and the projection of the tibial tuberosity (TT) on this plane with its positive direction upward.
x | The line connecting lateral and medial malleoli (LM e MM) and y axis define a plane: x is orthogonal to that plane with its positive direction forward (obtained as product between the two above described lines).
z | Product between axis x and y.
Origin | Midpoint between LM and MM.

Hindfoot | z | Parallel to the line connecting ST and peroneal tubercle PT with its positive direction from left to right.
y | The line connecting calcaneus (CA) and subtentaculum talii (ST) and the z axis define a plane: y axis is orthogonal to that plane with its positive direction upward (obtained as product between the two above described lines).
x | Product between axis y and z.
Origin | CA.

Midfoot | z | Parallel to the line connecting NT and C with its positive direction from left to right.
y | The line connecting (NT), and fifth metatarsal base (VMB) and z axis define a plane: y axis is orthogonal to that plane with its positive direction from proximal to distal segment (obtained as product between the two above described lines).
x | Product between axis y and z.
Origin | Midpoint between NT and C.

Forefoot | z | Parallel to the line connecting IMH and VMH with its positive direction from left to right.
y | The line connecting VMH and IIT and the z axis define a plane: y is orthogonal to the plane with its positive direction upward (obtained as product between the two above described lines).
x | Product between y and z.
Origin | Midpoint between IMH e VMH.

Foot | z | Parallel to the line connecting IMH e VMH with its positive direction from left to right.
y | CA, IMH and VMH define a plane; the line connecting IIT and CA belong to a plane perpendicular to the previous one; z axis is parallel to the line intersection between the two planes with its positive direction forward.
x | Product between axis y and z.
Origin | CA.

Table 3.2: Anatomical bone embedded frames [6].

Subsegment ground reaction forces

The vertical component of the ground reaction forces and the centre of pressure coordinates are available from both platforms and this gives the possibility to define a global coefficient as the ratio between the resultants of the vertical ground reaction forces measured by means of the force and pressure platforms, respectively. The former is then applied to correct the absolute force value delivered by the pressure platform. Thus the instruments simultaneously measures, for each sample, the ground reaction forces resultant and the plantar pressure distribution.

Local sub-segment vertical ground reaction forces are computed as the summation of the forces measured by each sensor of the pressure platform belonging to the same foot subareas. The anterior-posterior and medio-lateral ground reaction forces are calculated assuming that they are distributed proportionally to the vertical ground reaction forces. This is done under the simplified assumption that a foot loaded to half of the body weight can generate half the shear force of a foot with 100% body weight. The same assumption has been previously made in the literature [54], and yielded acceptable results.

Even if soft tissues are inelastic, with a nonlinear stress-strain relationship, the
3.2. PROTOCOLS

Hookean linear approximation is adopted as in [54]. The tissue is considered to be locally isotropic and the Poisson ratio is supposed to be a constant in all the tissue volume. Thus it is possible to calculate the shear forces acting on an elementary foot-to-floor contact area. Finally by assuming that the elementary area corresponds to the area covered by a sensor of the pressure platform located at a distance "d" from the centre of pressure, it is possible to define three different shear forces acting on the area: \( f_{xi}, f_{zi} \) that contribute to the resultant shear forces \( F_x \) and \( F_z \) respectively, and \( fM_i \) generated by the free moment applied to centre of pressure. By applying the above mentioned assumptions and given:

\[
 k_i = \frac{f_{yi}}{F_y} \tag{3.1}
\]

the following equations can be written as in Uccioli et al. [54]:

\[
 F_x = \sum k_i F_x - \sum |fM_i| \sin \alpha_i \sin x \tag{3.2a}
\]

\[
 F_z = \sum k_i F_z - \sum |fM_i| \sin \alpha_i \sin z \tag{3.2b}
\]

where \( x \) and \( z \) are the unit vector of the x and z axes respectively, \( Fx, Fy, \) and \( Fz \) are the components of the ground reaction force measured by means of the force platform, and \( f_{yi} \) is the elementary vertical ground reaction force measured by each pressure sensor. Ground reaction forces are then normalized to body weight.

**Plantar pressures**

The elaboration of plantar pressures distribution concentrates on the analysis of some significant parameters:

- the centre of pressures mean medio-lateral/anterior-posterior excursions and the curve integral are evaluated. Each footprint is compared with the others after rotating each one according to its longitudinal axes. The latter is defined as the line connecting the projection of the 2nd metatarsal head and the calcaneus markers on the footprint. Then the lateral side of the foot is considered as the positive medio-lateral direction, and the medial side of the foot as the negative one.

- peak and mean pressure curves obtained by linearly interpolating respectively the successive maximum or mean values of pressure during the whole stance phase (normalized to body weight);
• loaded surface curve obtained by linearly interpolating the successive medium values of surface covered respectively by the three foot subareas during the whole stance phase (normalized to the foot length).

**Motor tasks**

The protocol requires a static and a dynamic assessment of the subject. During the static trial subjects are asked to assume an upright posture with their feet placed with ankles together, toes pointed 30 degrees apart and the arms along the body for 60 seconds. To ensure similar angles during all the analysis, a guides made of heavy cardboard is placed between the performer’s feet to set them at the correct angle. Subject’s feet should line along both arms of the footguide.

In the dynamic trials indeed subjects are asked to walk at their normal speed of progression. Velocity, stride, and step parameters are calculated. At least three force-plate strikes of each limb (entailing simultaneous acquisition of both ground reaction forces and plantar pressures data) are recorded for each patient.

For each trial, all angular displacements are plotted over one stance phase.

The static acquisitions are used to figure out the analysed joints neutral orientations.

The problem of skin artefacts that create a bias which affect the real position of the bones, is minimized by the use of the static acquisition combined with a specific algorithm to define each segment anatomical bone embedded frames. This solution also results in a prevention of errors related to markers occlusion. The algorithm is based on the hypothesis that every segment behaves like a rigid body and so it checks for the mutual distances between markers placed on the same anatomical segment during the walking trials comparing them to the values obtained through the static analysis. At this point the distances between each marker belonging to the same segment are computed and there are different possible scenario. If the deviations are significant, the operator could decide either to correct the marker position through an interpolation procedure or, in the worst case, to exclude the trial from the analysis.

Summarizing, this protocol allowed the description of the complementary role of kinematics to kinetics, and plantar pressures in diabetic subjects gait by means of commercially available systems without applying any additional change to the original systems and this is a valuable capability of the system as the outcome measures are released from the specific system employed the acquisition.
Chapter 4

Foot Finite Element Model

4.1 Introduction

All the techniques seen in the previous chapter are generally used to predict joint kinetics and to quantify plantar pressure distributions though bones, soft tissue, and associate joint stresses inside the foot are still less investigated and remained uncertain. This is because of the difficulties in quantifying the in vivo bone and soft tissue stress even with the in vitro studies: the loading conditions are often different from the real physiological loading situation resulting in a compromised foot structure.

To overcome the problem many theoretical models such as kinematic models, mathematical models and finite element models of the foot had been developed. Finite element method (FEM) has been used more and more in many biomechanical investigations with great success due to its ability of modelling structures with irregular geometry and complex material properties, and the ease of simulating complicated boundary and loading conditions in both static and dynamic analyses. Furthermore it allows internal stress and strain analyses Consequently, it is now appreciate as an appropriate method for the investigation of foot stress distributions.

4.2 Finite Element Models

The figure 4.1 shows the steps that have to be done in order to obtain a FE model.

First of all a model is a symbolic device built to simulate and predict aspects of behaviour of a physical system [23]. Engineers and scientists use math models to perform simulation of physical systems in order to reduce costs. These models have an infinite number of degrees of freedom but however they are idealization
of the reality and so the characteristics of the system are usually simplified. The processing of a model often involved coupled partial different equation in space and time. Analytical solutions are more intellectually satisfying, but they tend to be restricted to regular geometries and simple boundary conditions, so usually numerical solutions are preferred. To make numerical simulations practical it is necessary to reduce the number of degrees of freedom to a finite number and this reduction is called discretization: the result is finally the discrete model.

4.2.1 Discretized geometry

The development of a FEM model starts always with the discretization of the geometry of interest using a collection of finite elements. Each element in the model represents a discrete portion of the physical structure and it is connected to other elements by shared nodes. The collection of all the elements and nodes is called the mesh. Generally the mesh is only an approximation of the actual geometry of the structure [24].

The element type, shape, and location, as well as the overall number of elements used in the mesh, affect the results obtained from a simulation. The number of elements per unit of length, area, or in a mesh is referred to as the mesh density. The greater it is, the more accurate the results. As the mesh density increases, the analysis results converge to a unique solution, and the computer time required for the analysis increases.

Type of mesh for the realization of FEM model

Given that the accuracy of a model especially depends on the mesh it is important to have an overview of how it could affect. There are different types of elements available to discretize the geometry of interest: for a 2D model triangular or quadrilateral elements, for 3D models tetraedral or hexahedral finite elements.

From analysts’ experience it has been proved that in 2D FE modeling, if there is a choice between triangles and quadrilaterals with similar nodal arrangement, quadrilaterals are always to be preferred. Triangles are quite convenient for mesh generation, mesh transitions and rounding up corners but in finite element anal-
yses with a given number of degree of freedom, 4-noded quadrilateral elements provide better results than 3-noded triangular elements [25]. In 3D FE modeling instead the preference is for the hexahedral elements. The tetrahedral mesh is fully automated, it allows the generation of millions of elements in minutes sometimes even seconds. It is adequate for some analysis but inaccurate for others. Hexahedral mesh instead is only partially automated, it usually requires user intervention and it is labor intensive. To obtain millions of elements, not only days but even weeks or months are needed. As a result most finite element models of the foot developed to date are built using tetrahedral elements [19–21, 27, 30, 31].

Another interesting approach now under study is the use of quadratic tetrahedral elements. They are different from the linear elements because they have an higher number of nodes, as shown in figure 4.2, indeed they are also called ten-node tetrahedron.

Figure 4.2: The figure shows the different disposition of the nodes in linear and quadratic tetrahedrals.

It has been proved that they performed as well as the hybrid hexahedral elements in terms of contact pressure and contact shear stress predictions. The limit is given by the fact that the simulations are more computationally expensive than the hexahedral element simulations [22]. Nevertheless this solution seems to be very promising as a result of decreased labour and expedited model development, all related to facilitated mesh generation.

A relatively fine discretization should be use in regions where a rapid variation of strains and/or stresses is expected. Regions to watch out for this are for instance the surrounding area of concentrated loads or near entrant corners or sharply curved edges findable for example in the geometry of the bones. When modelling complex structure, several materials can be put in contact and the elements of the mesh in these cases must not cross interfaces. Also shape of the element is an element of which to put attention. When discretizing two and three dimensional problems, finite elements of high aspect ratios (the ratio between the largest and smallest dimension of an element) as those shown in figure 4.3 should be avoided [26].
Such elements will not necessarily produce bad results but they introduce the potential for trouble. Anyway sometimes in thin structures modelled as continuous bodies, the use of such elements is inevitable on account of computational economy reasons.

All the suggestions provided should be respected: the choice of a mesh rather than another one can affect the accuracy of a model.

### 4.3 FEM models of the foot

Musculoskeletal models of the human foot are used as powerful tools to study the biological structures and the consequences of many diseases [29]. However they frequently lack the geometry detail necessary to provide meaningful insights into biomechanical behavior. The foot is a part of great interest because of its role as primary physical interaction between the body and the environment during gait. How variations in foot structure affect the interaction between the body and the environment is a field now under research. The first 3-D model of the foot published in literature which incorporates realistic geometric and material properties of both skeletal and soft tissue components of the foot, was developed for biomechanical analysis by Gefen et al. 2000 [27]. Nevertheless, it is difficult to obtain a nonlinear, three-dimensional FE model of the soft tissue-skeletal foot complex and so far only simplified biomechanical models have been generated lacking the necessary details which ensures accurate results. This is because models should represent the complex interplay of skeleton, cartilages, muscles, ligaments, fascia, and the external environment but the foot contains 26 bones, 33 joints, 107 ligaments and 19 muscles as stated in 1.1.
There are several studies that present 2D models [33–35] and in this case the simplification made are very big and they don’t give the possibility to study the individual motions of the bones.

Goske et al. [33] presented a 2D model of a slice of the heel pad in which the bones were unified in a unique structure. Their aim was to study the peak plantar pressures prediction and the behaviour of different insole materials. Agić et al. [34] instead modelled the first medial planar cross-section of the foot in standing posture with the objective to develop a patient-specific biomechanical model of the foot able to predict mechanical stresses transferred through the soft tissues padding, the medial metatarsals and calcaneous during standing, and the deformations of these tissues of diabetic patients. Erdemir et al. [35] developed a two-dimensional finite element model of the second metatarsal bone, plantar and dorsal soft tissue in order to use it for quasi-static simulation of the phase of gait approximating the time of the second peak in vertical ground reaction force. All the models are shown in figure 4.4.

Figure 4.4: A) 2D model of the heel pad developed by Goske et al. [33]. B) 2D model developed by Agić et al. [34]. C) 2D model developed by Erdemir et al. [35]. The figure on the top shows the model description, the one below the meshed model.

Trying to overcome the limit and to reduce the simplification of the 2D-model, several 3-D finite element models of the human foot have also been developed, but most of them has been uniquely tailored to study special conditions.

Qiu et al. presented a study in which the goal was to adopt the finite element modelling and analysis approaches to create a state-of-the-art 3D coupled foot-boot model for future studies on biomechanical investigation of stress injury mechanism, foot wear design and parachute landing fall simulation. In this case
the FE model of the foot and ankle consisted of tibia, fibula, talus, calcaneus, cuboid, navicula, three cuneiforms, five metatarsals and 14 components of the phalanges. They included also 70 ligaments and the plantar fascia. This structure was then bonded to the encapsulated soft tissues to form the model of the lower limb and finally this was combined with the model of the boot to create the coupled model [29].

Gu et al. developed a 3D FE model of the hindfoot incorporating a separate heel skin layer with the aim to gain a better insight into the biomechanical behaviour of the heel skin layer. The main objective was indeed to characterise the biomechanical responses of the hindfoot system during heel strike with potential variation of the skin stiffness [32]. He also developed a 3D FE model to investigate the effect of inversion positions on stress distribution and concentration within the metatarsals [38].

Zhi-hui Qian et al. presented a 3D FE model of the foot with the aim to describe in more details subject-specific representation of all major musculoskeletal structures, which could be used to investigate the interactions and responses inside of the foot musculoskeletal complex [36]. In this study, twelve major muscle groups around talocrural, subtalar and metatarsal-phanlangeal joints were constructed [36].

Pi-Chang et al. presented a study aimed to implement the finite element method to analyse the influence of different foot arches [37]. Some of these models are shown in figure 4.5.

An other example of a 3D model which helps to understands what are the typical simplification in the development of a model, is the one of Chen et al.: the bones in the five phalanges have been modelled to be five integrated parts and the rest of the metatarsals and tarsal bones have been modelled with two rigid columns (medial and lateral) [19]. The medial column consisted of the first three metatarsals, three cuneiforms, navicular, talus, and tibia. The lateral column consisted of the fourth and fifth metatarsals, cuboid, calcaneus, and fibula. The joint spaces between each of the five phalanges and its connective metatarsal bones has then been modeled with cartilage elements in an attempt to allow deformation and to simulate the metatarso-phalangeal joints. Also the joint space between the medial and the lateral columns was modeled with cartilage elements. The figure 4.6 shows triangular surface models for each of the bone and cartilage parts using assembled and exploded parts.

Sometimes also to FEM software are useful to reduce the complexity of the model because they offer the possibility to define contact surfaces which allow relative movement between the bones. Furthermore also the connection between ligaments or plantar fascia with the bones can be modelled by defining the at-
4.3. **FEM MODELS OF THE FOOT**

Figure 4.5: A) 3D coupled foot-boot model developed by Qiu et al. [29]. At right the detailed model of the foot, at left the complete model with the soft tissues and the boot. B) 3D Finite element model of the hind foot incorporating a detailed skin structure developed by Gu et al. [34]. C) General 3D foot model with a zoom on the metatarsals developed by Gu et al. [32]. D) 3D FE model with the muscles developed by Qian et al. [36].

Figure 4.6: Explode images showing the various parts of the foot model [19].
tachments bones. For instance Cheung et al. developed a finite element model using this simplifications [30].

All these models have increased the knowledge of the internal structure and of the behaviour of the foot, in special condition or in special disease. The challenge for the future is to continue to decrease the number of simplifications in order to obtain detailed specific-subject models which allows to achieve the best matching between reality and simulations.

4.4 Workflow to obtain a model of the foot

To obtain a model the first step is always the creation of the geometry based on the volume reconstruction of the coronal Computer Tomography (CT) or Magnetic Resonance Imaging (MRI) images of the foot in non-weight-bearing condition. Once collected, the images are segmented in order to separate each part of the foot complex. Then they are processed into a CAD environment to obtain the volume of interest and starting from this and using a FEM software it is possible to develop the model on which perform all the simulation of interest. All the steps are summarized in figure 4.7.

Figure 4.7: The figure, inspired by Cheung et al. [30], shows all the steps to perform in order to obtain a FEM model of the foot.
4.4. WORKFLOW TO OBTAIN A MODEL OF THE FOOT

4.4.1 Computed tomography and Magnetic Resonance Imaging

As said above the images used for the 3D reconstruction are usually CT or MRI scan images. The two techniques offer very different kind of results. MRI is a medical imaging technique used in radiology to visualize internal structures of the body in detail [17]. It makes use of the fact that body tissue contains lots of water (and hence protons) which gets aligned to large magnetic field to produce net average magnetic moment vector. The decay of MDM is detected as MR signal. MRI provides a very high ability to distinguish the differences between two arbitrarily similar but not identical tissues. The basis of this ability is the complex library of pulse sequences that the modern medical MRI scanner includes, each of which is optimized to provide image contrast based on the chemical sensitivity of MRI. In order to obtain the best images matching the requirement for a good segmentation and geometry reconstruction, it is possible to change different parameters. The two basic parameters of image acquisition are the echo time (TE) and the repetition time (TR). With particular values of those, a sequence takes on the property of T2-weighting. On a T2-weighted scan, water- and fluid-containing tissues are bright and fat-containing tissues are dark. The reverse is true for T1-weighted images. The typical MRI examination consists of 5-20 sequences, each of which are chosen to provide a particular type of information about the subject tissues. Other parameters to consider are the distance between the slices, usually around 2 mm, and what accuracy it is possible to obtain.

Computed Tomography (CT) is a powerful non-destructive evaluation technique for producing 2-D and 3-D cross-sectional images of an object from flat X-ray images. X-ray slice data is generated using an X-ray source that rotates around the object; X-ray sensors are positioned on the opposite side of the circle from the X-ray sour. Characteristics of the internal structure of an object such as dimensions, shape, internal defects, and density are readily available from CT images because of its high-contrast resolution: it allows to distinguish the differences between tissues that differ in physical density by less than 1 mm. The interval between two slides varies from 1 mm to 2 mm. One advantage of an MRI scan is that it is harmless to the patient. It uses strong magnetic fields and non-ionizing radiation in the radio frequency range, unlike CT scans and traditional X-rays, which both use ionizing radiation. However the average cost of a MRI acquisition is higher than the CT. For a FEM model of a part of the body, it could be necessary to have a good definition of all the different tissue but also a good spatial resolution if there is the aim to obtain a complete model with all the biological structures.
Thus the consideration that drives the choice of what techniques to use regards the safety of the patients. As a matter of fact MRI is often the choice because of its characteristic to no determine biological hazards even if the images could be more difficult to segment and so they can cause lower precision of a FEM model. CT is mostly used in the studies carry on cadaveric part of the body, such as cadaveric foot. Anyway the part under modelling should not be weight bearing and in the case of the foot that means that the patient must not be standing but lying in supine position or seating. Sometimes devices are used to keep the foot in position as ankle-foot ortoshes. This wariness is very important in order to obtain a real representation of the foot, no suffering the actions of any forces.

4.4.2 Segmentation

Segmentation is the process of subdividing a digital image into several portions. The aim of segmentation is to simplify or change the representation of an image into something that is more meaningful and easier to analyse. In the field of FEM models it is used to obtain the geometry of interest from the clinical images locating objects and boundaries of the various structures. The result of image segmentation is a set of segments that collectively cover the entire image. All the pixels in a region are similar with respect to some characteristic as the belonging of a specific bone or soft tissue. Adjacent regions are significantly different with respect to the same characteristic. To develop a model the segmentation is applied to a stack of images and so the resulting contours obtained can be used to create 3D reconstructions. During this phase a lot of different filter can be used. For instance with the CT images is possible to extract the part of interest just setting a threshold to contrast bones. This is because in this images the contrast among the different tissues is very high and so a manual pixel-level user input is required only to define some suboptimal delineated border. MRI images instead have a larger grayscale and so segmentation required more manual work than a simple setting of a grey threshold. For this reason the geometry obtained by MRI images is sometimes less accurate than the one from CT. The software more used in the literature for the segmentation is MIMICS. It is specially developed for medical image processing. It should assure highly accurate 3D model of patients’ anatomy when used for segmentation of a large range of medical images, coming from CT, MRI, micro CT, Ultrasound and so on.

4.4.3 Cad modelling

After the segmentation of MRI/CT images there is always the needing for another step before starting to use a FEM simulator. From the surfaces achieved with the
images indeed, it is necessary to obtain clean geometries for adequate discretization. Various tools exist for smoothing and obtaining triangulated or parametric surfaces. One of the most used in literature is Solidworks that allows to form solid models of each part, but other choices can be Rhinoceros or SALOME.

4.4.4 Fem Modelling

A FEM model could always be divided into three step, one of pre-processing, one of process and one of post-processing. The first one regard the definition of all the properties of the model that means materials, mesh, boundary and loading conditions. The latter instead is about the variables obtained with the simulation, the control of the convergence and the validation.

Pre-processing - Material

With the solid model of the foot it is possible to begin the build of the FEM model. First of all the definition of the materials should be done. Usually bones and cartilages are considered as linearly elastics and so Young’s modulus and Poisson’s ratio are the only parameters requested. Sometimes the bones are considered only as rigid body. The soft tissue instead, as the fat pad and the skin, is a non linear material and has to be modelled with hyperelastic models. In the literature several model are available for this purpose, but one of the most used, in particular for 2D FEM models of the foot, is the Ogden’s model [32,33]. The first-order Ogden model describes the strain energy potential (U) in the form of:

\[ U = \frac{2\mu}{\alpha^2}(\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3) \] (4.1)

where \( \lambda_{1,2,3} \) are the deviatoric principal stretches and \( \mu \) and \( \alpha \) are the material properties representing the hyperelastic behavior of the heel pad. \( \mu \) is the initial shear modulus and therefore it is linearly related to initial elastic modulus. The \( \alpha \) value is a measure of increase in tangential modulus with increased strain and therefore describes a change in high strain behaviour [24]. Other studies prefer the use of the general hyperelastic material model employing a second-order polynomial strain energy potential. This choice is usually made for the 3-D models. The form of the potential is:

\[ U = \sum_{i+j=1}^{2} C_{ij}(I_1 - 3)^i(I_2 - 3)^j + \sum_{i=1}^{2} \frac{1}{D_i}(J_{el} - 1)^{2i} \]

Where \( U \) (Nm\(^{-2}\)) is the strain energy density, and \( I_1, I_2 \), and \( J_{el} \) (dimensionless)
are the first and second deviatoric strain invariants and elastic volume ratio, respectively. The coefficients $C_{ij}(N m^{-2})$ and $D_i(m^2 N^{-1})$ correspond to material characteristics [24]. Finally when the studies include the test of material of the shoes, an hyperfoam strain energy function is used in the form of:

$$U = \frac{2\mu}{\alpha^2} [\lambda_1^\alpha + \lambda_2^\alpha + \lambda_3^\alpha - 3 + \frac{1}{\beta} J^{-\alpha\beta} - 1]$$

(4.2)  

Where  

$$\beta = \frac{\nu}{1 - 2\nu}$$

$\lambda_{1,2,3}$ are the principal stretches and $\mu$, $\alpha$ and $\nu$ the effective Poisson’s ratio) are the material properties.

**Pre-processing - Mesh**

As already said above the choice of what mesh to use in a model is a sensitive phase because it influences the results of the simulation. FEM software usually provides an internal part dedicated to the construction of the mesh but anyway there are several software dedicated. An example of a real good tools for hexahedral mesh of structures is IA-FEMesh. In an attempt to facilitate anatomic FE model development this is a freely available software toolkit developed by the university of Iowa. IA-FEMesh employs a multiblock meshing scheme aimed at hexahedral mesh generation. The goal is to provide an efficient and reliable method for model development, visualization, and mesh quality evaluation.

**Pre-processing - Assembly, interaction, loading and boundary conditions**

A model of the foot comprises different parts as bones, cartilages and ligaments and each part has its own description regarding material and structure. Then model components has to be spatially arranged and it is necessary to define the interaction among them. So geometric constraints have to be defined as tissue-device coupling, contacts and interactions between different physical domains. Once obtained the whole assembly it is possible to continue with the application of the loading and boundary conditions. Movement of components, what surfaces are fixed and what type of loads are applied (if point forces or surface pressures) have to be taken into account.

**Processing and post-processing**

At this point the model can be processed and the results evaluated. The convergence of the model depends on different factors and one of these is the number
of elements. The latter is related with the problem under analysis and so it is not possible to define a priori what is its value. To know if the calculations are correct simpler problems with previously verified results have to be processed. Finally a validation should be done, comparing the experimental measurements with the results obtained: if the predictions are correct the model is capable of reproducing experiments.

The solution obtained from the numerical model is generally an approximation to the solution of the physical problem being simulated. The extent of the approximations made in the geometry of the model, material behaviour, boundary conditions, and loading determines how well the numerical simulation matches the physical problem.
Chapter 5

Materials and methods

5.1 Introduction

In this chapter all the steps to develop the models used for the simulations in this study are shown. Starting from the acquisition of foot images and going through the construction of the geometry and the definition of the characteristics of the FEM model, all the phases will be shown in detail. Two subjects have been analysed, a diabetic and an healthy one, and all the step have so been repeated two times in order to obtain two specific models. Because magnetic resonance imaging has emerged as the most useful non-invasive tool with which biological structures can be studied [41], also in this work MRI technique has been used to record the images of interest.

5.2 Subjects

Two subjects have been used in this study, an healthy and a diabetic one. For both a model has been developed using their own MRI of the foot and then all the simulations have been done with their specific force and pressures data acting on their whole foot and hindfoot, collected in the Laboratory of Engineering of the Movement of the University of Padua. Their physical characteristics are summarized in table 5.1.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Sex</th>
<th>Age</th>
<th>Weight (kg)</th>
<th>Height (cm)</th>
<th>BMI</th>
<th>Foot number</th>
<th>Type of foot</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy</td>
<td>Female</td>
<td>29</td>
<td>61</td>
<td>174</td>
<td>20.2</td>
<td>40</td>
<td>Dx, sx: normal</td>
</tr>
<tr>
<td>Diabetic</td>
<td>Male</td>
<td>72</td>
<td>79.5</td>
<td>175</td>
<td>25.1</td>
<td>42.5</td>
<td>Dx, sx: cavus 1°</td>
</tr>
</tbody>
</table>
Table 5.1: Data of diabetic and healthy subjects.

The diabetic subject suffers from diabetes mellitus type 1 for 45 years. He has also developed a retinopathy and a vasculopathy and furthermore he suffers from an hypertensive pathology.

5.3 MRI acquisition

The acquisitions of the MRI of the foot of the subjects described in 5.2, have been done in two different places and so the characteristics of the instruments used are different. The MRI of the healthy subject has been recorded in the Abano Terme General Hospital which is provided with radiology and diagnostic imaging services lead by MRI and CT scans. The department has at its disposal a 1.5 Tesla MRI machine, (PHILIPS-E7A0411), capable of capturing high-resolution images of the body as well as sophisticated angiographical and functional studies [40]. This machine has been used in this work with particular attention in the setting of the parameters to obtain the best images to segment in the following step. In the literature several studies prefer the CT technique even tough it is more invasive for the patients. The desired space between slides is less than 0.5\text{mm} (0.3\text{mm} best) and the slice thickness less than 1\text{mm} (0.5\text{mm} best). The value of the resolution findable in the literature with a CT scan is 0.25\text{mm} and the value of the slice thickness is 0.4\text{mm} [19]. This small values are difficult to obtain with the MRI but other studies that use MRI state that a resolution of 1\text{mm} and a slice thickness of 7\text{mm} were found to be optimal in terms of quality and scanning time [42].

For the MRI of the healthy subject the acquisition mode used has been a multi fast field echo (mFFE), T2 weighted. Spacing between slides obtained has been of 0.6\text{mm} and the slice thickness of 1.2\text{mm}. The MRI of the diabetic subject instead, has been recorder at the Regional Hospital of Padua. The ward of radiology is provided with a 1.5 Tesla machine (SIEMENS. The acquisition mode has been a me\text{3d} sequence, T2 weighted. The value of slice thickness was 1.5\text{mm}. In both cases the foot has been placed in non weight-bearing condition. Figure 5.1 shows two DICOM images used in this work.

5.4 Segmentation [43]

Once obtained the DICOM images of both subjects, next step has been the segmentation of the region of interest: a slice of the heel pad. The software adopted has been Simpleware, a worldwide leader tool for converting images into 3D CAD and finite element models. It is based on a central platform of image
5.4. SEGMENTATION

Figure 5.1: A: DICOM image of the diabetic foot. B: DICOM image of the healthy subject. It is possible to note the difference of the quality depending on the acquisition mode adopted.

processing, ScanIP, with optional modules for the mesh generation and CAD integration. The relations of these products are shown in the figures below.

Figure 5.2: Simpleware software products [43].

ScanIP offers a wide selection of image processing tools to assist the user in the visualization and segmentation of the regions of interest from all the 3D volume data as MRI and CT. Then it allows the export of .STL or .IGES files for CAD analysis or RP manufacturing. A file .STL is dedicated to stereolithography CAD software and it represents a solid whose surface has been discretized into triangles.

The module +FE instead allows the conversion of 3D segmented images in volumes and/or surface mesh. The generated mesh can also be imported directly from a series of commercial FE and CFD software.

The +CAD module allows the import of CAD models and their interactive positioning within the images. The resulting model can then be exported as a multipart CAD model or converted with ScanIP into a FE model. All the components are shown in figure 5.2.
The process used to analyse the images recorded in the previous step involves some phases also shown in figure 5.3:

- preparation of data and filtering;
- segmentation;
- filtering of the masks;
- extraction of surface or mesh generation.

Figure 5.3: Work flow of the segmentation of the DICOM images. The figure is inspired by the guide of ScanIP [43].

Once the images to be analysed such as in this case a stack of DICOM has been imported, there is the stage of pre-processing the data. At this phase it is already possible to make a first filtering, with actions of cutting and resampling. The first consist simply in cutting parts of the volume to retain only the data of interest in order to obtain the minimum use of the memory. It is absolutely not necessary to cut too close to the object, but it is actually better to leave some space to avoid trouble during the mesh phase. The resampling instead can be used when it is necessary to reduce the size of an image by adjusting the resolution. This can result in a loss of information so it has not been used, but in any case it does not modify the size of the objects.

Next step is the segmentation. In ScanIP segmentation consists in the generation of one or more volumes called "masks" from the data volume in grayscale (in figure 5.4 an example is shown). With CT images the segmentation can be very fast and easy because the software gives the possibility to automatically subdivide the regions with grey thresholds. This solution is possible because with the CT the distinction between different tissues is very clear and well determinable. On the other hand the segmentation of MRI images requires the hand work of the user. For this purpose ScanIP offers several tools, as the paint and paint with threshold instruments. The first one is fully manual the second instead gives an additional threshold to restrict the voxels which can be painted. When the
5.5. **CONSTRUCTION OF THE GEOMETRY**

slice of interest has been chosen, these two painting instruments has been used to highlight all the biological structures of the foot such in this case the fat pad, calcaneus, fibula, anklebone and tibia. When masks have been obtained, they have been compiled, edited and filtered until considered satisfactory.

![DICOM images of the foot with masks applied](image)

Figure 5.4: DICOM images of the foot with masks applied. At left the healthy subject’s foot, at right the diabetic’s foot completely segmented.

One of the more used filters in this phase has been the Recursive gaussian filter which reduces image noise and detail levels. The visual effect of this blurring technique is a smooth blur of the image. Before this, a dilate filter which makes the region grow, has been applied because in this way the Gaussian filter eliminates less pixels of the real boundary.

When the filtering phase has been terminated the mesh has been obtained with the possibility to export the .stl file to use in the next step.

5.5 **Construction of the geometry**

The use of the representation of the structure so far obtained in the development of FEM models, requires a further processing software of the geometry. This should allow to obtain a solid model usable in the FEM modeling. In this work a trial version of Rhino 4.0 has been used. This software allows the conversion of the mesh file .stl in files .igs then importable into Abaqus for the FEM model. When the .stl file has been imported from ScanIP, the curves of the surface have been tracked and then a solid surface has been obtained. The figure 5.5 shows a screenshot of the work done with Rhino.
Figure 5.5: Screenshot of the software Rhino used in order to obtain the geometry of the foot.

5.6 Finite Element Models

5.6.1 COMSOL Multiphysics

Initially COMSOL Multiphysics was considered for the creation of the FEM model of this study but then it has been abandoned because of too many drawbacks. The main limitations encountered are likely explained by the multidisciplinary nature of the software that is not useful for this study given that it requires only structural mechanic: the module related to the latter is not as developed as the dedicated FEM applications such as Abaqus.

The first problem encountered in the development of the models of this study using COMSOL, regards the import of geometry. The 2D geometry that has been imported in a .DXF file, had been obtained with Rhino. The process is very similar for all FEM software, however COMSOL is less intuitive than others because it is not possible to define at the beginning for each imported item its unique features. This results in going through all the modules to add every time the desired properties.

COMSOL has an extensive library of materials already available for the user, but being a software used only partially in the field of bioengineering it has some deficiency in the definition of hyperelastic materials such as human soft tissues. Attested that in literature the law used to described the behaviour of soft tissue in 2D model is mainly the Ogden’s law [33], the absence of it in the library has created some difficulties. Anyway COMSOL gives the possibility to manually define a new material. The tutorial provided by COMSOL explains how to define an hyperelastic material through the density energy function that in this case is defined by the equation 4.1. As a first step, the procedure requires the definition of the parameters $\mu$ and $\alpha$ in the global definitions where there is the possibility
to insert variables and constants. The next step causes instead some uncertainty: it regards the inclusion of the entire law in the modulus of structural mechanics, choosing to add a hyperelastic model. Hence it has to be found what in COMSOL language corresponds to $\lambda_{1,2,3}$, the principal stretches. In the guide there are some definitions from which by calculation the identity of them within the software can be traced (solid.stchelp1, solid.stchelp2, and solid.stchelp3). Finally at this point it is possible to complete the definition of the material.

From what said above the direct creation of an hyperelastic material described by the Ogden’s law in the modulus of the material is not allowed and this makes all the process confusing and difficult to understand.

Once obtained the geometry and the description of the materials, the establishment of contacts between the various elements should be done. This creates significant difficulties that compromise the convergence of the model. There are only two types of contact, the identity pair and the contact pair, but the distinction is not clear. The first, with the condition of setting the continuity in the modulus of structural mechanics, should ensure a smooth transition of the properties between two different elements, with the second instead it is possible to set a coefficient of friction.

A limit of COMSOL is its inability to distinguish between two surfaces when they coincide. This probably happens because of the impossibility to define for each element its own characteristics as already said. A partial solution is provided by the use of the Form Assembly, a command in the modulus of geometry, that automatically defines the identity pair between two elements in contact. Unfortunately it does not consider the whole surface but only single points. Another idea could be to perform this step manually, but again a difficulty arise because the surface has not been internally drawn: the edge are imported in COMSOL as if they were made of an infinite number of points. So if the model includes contacts, first of all the groups that include full arches has to be created with the function explicit provided by the software.

To perform some tests a trial model has been created in which two rectangles drawn within COMSOL were put in contact. Also in these simplified conditions the model did not converge when a force was applied unless both materials used were already in the library of COMSOL. This problem with the materials prevented the execution of tests useful to understand the basic operation of the simulator. Given that in this model the use of the user defined Ogden’s law was mandatory the final decision has been to choose another software more specific
64 CHAPTER 5. MATERIALS AND METHODS

for the structural mechanics applications.

5.6.2 Abaqus/ Cad Environment Complete

Abaqus is a FEM simulation software dedicated to structural mechanics which consists of three basic product: Abaqus / Standard, Abaqus / Explicit and Abaqus / CAE (Abaqus Environment Complete). Each one offers optional modules that cover specialized skills required by some clients:

- Abaqus / Standard provides the technology for the traditional finite element analysis, including static, dynamic and thermal studies, enhanced with a wide range of options for contact and nonlinear materials;

- Abaqus / Explicit analysis provides the technology focused on transient and quasi-static dynamics;

- Abaqus / CAE provides a simple and consistent interface for creating, submitting, monitoring and evaluating the simulation results obtained with Abaqus / Standard and Abaqus / Explicit. It is divided into modules and each one deals with logical aspect of the modeling process: the definition of the geometry or of the material properties, the generation of a mesh and so on. Moving from module to module the model is compiled resulting in an input file that will be submitted to Abaqus / Standard and Abaqus / Explicit. Once the analysis is completed the information are sent back to Abaqus / CAE. This allows to monitor the progress of the work and to get an output database from which the result of processing can be displayed.

The choice of using Abaqus in this study has been driven from the high number of scientific publications available where its use is documented [19], [33]. Furthermore a quick comparison with what said about COMSOL in the previous paragraph 5.6.1 emphasizes his superior performance for bioengineering applications. First with regard to the import of geometries, also in this case it is necessary an intermediate step to another software in order to manipulate the geometry and to obtain a file .igs or .step. The difference is given by the modular structure in which you can define a model: it is possible to form the assembly but still modify and work on the parts separately.

In this study two loading conditions have been simulated, one with the bare-foot in contact with the force platform and one with the foot interacting with various shoe components in realistic geometries. The geometry of the slice of the heel has been imported from Rhino in two separated parts: one representing the
tissues and the other one all the visible bones simplified in an unique structure. The force platform with which the foot was in contact has been sketch in the dedicated part of the software as also the shape of the shoes.

For each element specific characteristics have been defined from the beginning, such as surfaces, materials and mesh. Abaqus does not provide a library of materials but the manual definition is very easy. In the Materials module indeed there are a lot of mechanical law to use depending on the characteristic of the material. If it is an hyperelastic one, also the Ogden’s law is available and the user has only to fill the empty field of the parameters as in this case $\alpha$ and $\mu$. For this study these value for the healthy subject has been found in literature from Goske et al. [33], table 1. The $\alpha$ and $\mu$ values for the diabetic patients instead has been found in Erdemir et al. [44], table 1.

Bone was assumed to be rigid. The stress-strain response of the insole materials was represented by a hyperfoam strain energy function with the equation 4.2 and the parameters found in literature [33]. Finally the leather and the material of the force platform, Aluminium 6061, were assumed to be linearly elastic. All the values of the parameters are listed in table 5.6.2.

<table>
<thead>
<tr>
<th>Material</th>
<th>$\mu$(kPa)</th>
<th>$\alpha$</th>
<th>$\nu$</th>
<th>E(kPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Heel pad</td>
<td>16.45</td>
<td>6.82</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Diabetic heel pad</td>
<td>16.88</td>
<td>7.02</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Insoles:</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Microcel Puff</td>
<td>1340</td>
<td>28.14</td>
<td>0.054</td>
<td>-</td>
</tr>
<tr>
<td>Microcel Puff Lite</td>
<td>1220</td>
<td>48.29</td>
<td>0.028</td>
<td>-</td>
</tr>
<tr>
<td>Poron Cushioning</td>
<td>620</td>
<td>34.46</td>
<td>0.037</td>
<td>-</td>
</tr>
<tr>
<td>Midsole: firm crepe</td>
<td>4240</td>
<td>28.59</td>
<td>0.076</td>
<td>-</td>
</tr>
<tr>
<td>Sidewalls: leather</td>
<td>-</td>
<td>-</td>
<td>0.3</td>
<td>200</td>
</tr>
<tr>
<td>Force platform (Alu6061)</td>
<td>-</td>
<td>-</td>
<td>0.3</td>
<td>68900</td>
</tr>
</tbody>
</table>

Table 5.2: Coefficients ($\mu$, $\alpha$) represent the stress-strain relationship of the components used in finite element simulations of barefoot, foam mat and shoe insert loading. The heel pad has been modelled as hyperelastic with the Ogden’s equation 4.1, [33]; insole foams and the midsole were represented by hyperfoam material models with the equation 4.2, obtained by fitting stress-strain data measured under compression from [33]. The sidewalls and the force platform were assumed to be linear elastic [33].

The definition of the contact and the constraint has also been simpler than using COMSOL. For each part indeed it is possible to define the surfaces that
will be in contact with the surfaces of other elements. The heel pad has been tied together with the bone in order to have a continuity. A tie constraint indeed allows you to fuse together two regions even though the meshes created on the surfaces of the regions are dissimilar, so that there is no relative motion between them [24]. For the bone a rigid body constraint has been applied.

In the first condition with the barefoot on the platform, a penalty contact with an estimated coefficient of friction of 0.5 has been used to model the interaction between the heel pad and the contacting surface. In the second condition other constraints have been defined for the components of the shoes.

Figure 5.6: Plane strain finite element models of the heel: (A),(C) Barefoot models to validate the lumped representation of the heel pad in predicting plantar pressures respectively of diabetic and healthy subjects. (B),(D) Footwear models interacting with the heel to investigate the influence of insole parameters on heel pressure relief respectively of diabetic and healthy subjects.
A midsole made out of Firm Crepe has been included in the footwear. The sidewalls of the shoe have been modeled to represent stiff leather and have been attached to the midsole by binding the base nodes to the midsole boundary using tie constraints. An insole has been placed on top of the midsole and in between the sidewalls to investigate pressure reduction by changing insole parameters. The insole has been tied to the inside of the shoe and penalty contacts with an estimated friction coefficient of 0.5 have been used to model the interactions between the heel pad and insole, and the midsole and floor.

All the components of the model have been meshed with linear quadrilateral elements in an hybrid formulation. The size of the elements is different for each part. To reach the convergence the size of the mesh elements of the diabetic tissue has been set larger than the one of the healthy subject.

The procedure described above has been repeated two times in order to obtain a model of the healthy subject and one of the diabetic patient. A screenshot of all the models realized is shown in figure 5.6.

### 5.7 Kinematic and kinetic data

In order to evaluate the capacity of the models to simulate loading applied on the foot during walking, kinematic and kinetic data have been collected in the Laboratory of Bioengineering of the Movement at the University of Padua for both the subjects described in 5.2. The protocol used has been the one of Sawacha et al. [53] presented in the paragraph of protocols 3.2. The choice of using integrated data acquired by three different instruments derives by all the scientific publications about the best performance in terms of providing a more exhaustive and detailed view of foot loading during activities. Furthermore it is possible to obtain subject-specific data usable for further analysis as in this study a FEM model. Indeed it has been possible to obtain all the forces and pressures of the whole foot, but also for each part of the foot described in 1.2.

The laboratory is provided with a BTS motion capture system, a digital optoelectronic system with six cameras operating in the range of 60 – 120 Hz, that allows all types of movements to be analysed with an high level of precision and accuracy. A landscape of the laboratory is shown in figure 5.7.

All its technical details are listed in the table 5.3.
CHAPTER 5. MATERIALS AND METHODS

Figure 5.7: On the left one of the six infrared camera of the laboratory, on the right a view of the laboratory.

<table>
<thead>
<tr>
<th>Specifications</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Resolution</td>
<td>$768 \times 576\text{pixel}$</td>
</tr>
<tr>
<td>Accuracy</td>
<td>$&lt; 0.2mm$ in a volume of $3 \times 3 \times 2m$</td>
</tr>
<tr>
<td>Lenses</td>
<td>Interchangeable, C-mount compatible</td>
</tr>
<tr>
<td>TVC power</td>
<td>by the dedicated hub</td>
</tr>
<tr>
<td>Marker</td>
<td>Passive and retro-reflecting markers from 3 to 20mm φ</td>
</tr>
<tr>
<td>Marker detection system</td>
<td>Enhanced Blob Analysis</td>
</tr>
<tr>
<td>Data station</td>
<td>dual-core Intel® XEON® and PCI-X architecture</td>
</tr>
<tr>
<td>Included software</td>
<td>BTS SMART Suite</td>
</tr>
</tbody>
</table>

Table 5.3: Smart technical specifications.

Subjects have been asked to wear a pair of black shorts and then skin passive markers have been attached through double sided tape. The choice of using markers has been motivated from the fact that they are easy recognizable by the camera and anyway they are the most common solution utilized in motion analysis as explain in paragraph 3.1.1. After the acquisition a tracking step has been done using the SMART tracker software provided by BTS, where all the markers has been given a name.

The forces developed during walking have been recorded with two Bertec force plates (FP4060-10). They use the strain gauge technology and an high accuracy with no drift is documented. They have an high resolution with 0% cross-talk and they don’t suffer from signal interference from outside sources. Finally they also have an high load capacity of 10000N. Placement and orientation of the force platforms (FP) has been chosen such that ground reaction forces during gait can be acquired from each foot individually. This configuration of the FPs
was found suitable for healthy young and older subjects. Linoleum tile sample floors matching the rest of the walkway flooring material are attached to the FPs. An image of the force plate is shown in figure 5.8

Figure 5.8: Bertec force plate [56].

All the technical specifications about the Bertec force plates are listed in the table ??.

<table>
<thead>
<tr>
<th>Specifications</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Type</td>
<td>4060H</td>
</tr>
<tr>
<td>Dimensions</td>
<td></td>
</tr>
<tr>
<td>Width</td>
<td>400mm</td>
</tr>
<tr>
<td>Length</td>
<td>600mm</td>
</tr>
<tr>
<td>Height</td>
<td>100mm</td>
</tr>
<tr>
<td>Force range</td>
<td></td>
</tr>
<tr>
<td>$F_z$ (vertical force)</td>
<td>10KN</td>
</tr>
<tr>
<td>$F_x$, $F_y$ (shear forces)</td>
<td>5KN</td>
</tr>
<tr>
<td>Overload capacity</td>
<td>50%</td>
</tr>
<tr>
<td>Sensitivity $F_z$</td>
<td>0.5mV/N</td>
</tr>
<tr>
<td>$F_x$, $F_y$</td>
<td>0.85mV/N</td>
</tr>
<tr>
<td>$M_x$</td>
<td>1.6mV/Nm</td>
</tr>
<tr>
<td>$M_y$</td>
<td>2.25mV/Nm</td>
</tr>
<tr>
<td>$M_z$</td>
<td>2.75mV/Nm</td>
</tr>
<tr>
<td>Linearity</td>
<td>$\pm 0.2%FSO$</td>
</tr>
<tr>
<td>Hysteresis</td>
<td>$\pm 0.2%FSO$</td>
</tr>
<tr>
<td>Cross-talk</td>
<td>$\pm 0.2%$</td>
</tr>
<tr>
<td>Natural Frequency</td>
<td>1000Hz</td>
</tr>
<tr>
<td>Mass</td>
<td>25kg</td>
</tr>
</tbody>
</table>

Table 5.4: Bertec technical specification.

The laboratory is also provided with two pressure platforms of size $410 \times 410 \times 0.5\text{mm}$ produced by Imargotesi in Piacenza: one of them is shown in figure 5.9. They ensure a $0.64cm^2$ resolution, and a frequency of $150Hz$. They have an high
number of resistive sensors (2304) electronically calibrated and extra-flat. They are also light and powered directly from the PC via USB cable that results in a great transportability.

![Figure 5.9: The figure shows at left one of the two pressure platforms used, at right an example of image obtainable during a dynamic acquisition.](image)

The WINPOD software which belongs to this system platform is very powerful and ergonomic. It gives the possibility to apply filters to reduce the noise. It performs static and dynamic analysis. The first provide a recording of the pressure performed by upright people on both feet with no support, with or without shoes. Such analysis allows the collection of a lot of different data such as the barycentre of the body and the maximum and average pressures of the feet. Dynamic analysis instead record the foot pressures during walking.

The three systems have been temporally and spatially synchronized and all the data have been processed together in order to obtain the data for the FEM simulations. Vertical, medio-lateral and antero-posterior forces has been determined for the whole foot, hindfoot, midfoot and forefoot as well as the maximum pressures for both feet.
Chapter 6

Results

6.1 Introduction

Once the models was developed and all the kinetic and kinematic data of both the healthy and the diabetic subjects have been acquired, lots of simulations have been performed. First of all the results obtained for each one are presented, then a comparison between the two is shown. It should be underlined that some tests have also been done with both the medio-lateral and the vertical forces applied on the foot. Anyway given that the medio-lateral force has always much lower values compared to the vertical force as it is shown in figure 6.1, the simulated pressures remained always the same in both situations.

Figure 6.1: In the figure the vertical and medio-lateral forces developed on the hind-foot during the stance phase of gait of the diabetic subject are shown. It should be underlined the difference in the scale of the Body Weight between the two graphics: the medio-lateral force is always lower than the vertical one.
6.2 Healthy subject

In this section all the simulations that have been made using the data of the healthy subject are shown. The first four simulations have been conducted in barefoot condition, the last one with the footwear. Models are shown in figure 6.2.

Figure 6.2: The two figures at left show the models developed for the healthy subject, the one on the top in barefoot conditions and the one on the bottom with the sketch of the shoes. The figures at right show the models with a force applied.
6.2. HEALTHY SUBJECT

6.2.1 Experimental vs simulated data in barefoot condition

In this test the maximum pressures borne by the whole foot and the hindfoot have been determined from the collected data for all the dynamics. Then the force corresponding to these pressures have been extracted from the experimental data to be applied in the FEM model in order to obtain the simulated pressures. Finally a comparison has been done.

Figure 6.3: In the figure the histograms showing the experimental data compared to the simulated data are displayed.

As it is shown in figure 6.3 model predicted plantar pressures are in good agreement with those measured during compressive loading of the heel. All the peak plantar pressures values are reported in tables 6.1 and 6.2.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>579.4129</td>
<td>609.3861</td>
<td>505.5654</td>
</tr>
<tr>
<td>Simulated</td>
<td>391.4090</td>
<td>468.7390</td>
<td>365.4710</td>
</tr>
</tbody>
</table>
Table 6.1: Experimental and simulated peak plantar pressure values of the healthy’s whole foot.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>550.5045</td>
<td>609.3861</td>
<td>483.6330</td>
</tr>
<tr>
<td>Simulated</td>
<td>385.6440</td>
<td>459.2830</td>
<td>428.6320</td>
</tr>
</tbody>
</table>

Table 6.2: Experimental and simulated peak plantar pressure values of the healthy’s hindfoot.
6.2.2 Experimental vs simulated data of all the phases of the stance phase in barefoot condition

In this test the maximum pressures borne by the whole foot and the hindfoot have been determined from the collected data for all the dynamics and for each phase of the stance described in paragraph 1.4. Then the force corresponding to these pressures have been extracted from the experimental data to be applied in the FEM model in order to obtain the simulated pressures. Finally a comparison has been done.

The results represented in figure 6.5 and in figure 6.6 show again a good agreement between predicted and experimental pressures for all the cycles.

![Graphs showing vertical force applied on the hindfoot and whole foot.](image)

Figure 6.4: In the figure the trend of the vertical force applied on the hindfoot and on the whole foot of the healthy subject are shown.

Furthermore the percentages of the stance phase where the forces applied on the hindfoot are maximum (Figure 6.4) correspond to the ones where the pressure are maximum, and this is well documented by both experimental and simulated data. On the contrary the trend of the forces to which the whole foot is subjected is different.

The tables below shows all the peak pressure values of the healthy subject’s whole foot.

<table>
<thead>
<tr>
<th>Initial contact</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>430.4809</td>
<td>374.6199</td>
<td>367.9868</td>
</tr>
<tr>
<td>Simulated</td>
<td>370.6950</td>
<td>288.3370</td>
<td>193.7650</td>
</tr>
</tbody>
</table>

Table 6.3: Experimental and simulated peak plantar pressure values of the healthy subject’s whole foot during the initial contact.
### Loading Response

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>456.7744</td>
<td>569.0061</td>
<td>454.2560</td>
</tr>
<tr>
<td>Simulated</td>
<td>379.8320</td>
<td>492.6580</td>
<td>392.6580</td>
</tr>
</tbody>
</table>

Table 6.4: Experimental and simulated peak plantar pressure values of the healthy subject’s whole foot during the loading response.

### Midstance

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>550.5045</td>
<td>609.3861</td>
<td>483.6333</td>
</tr>
<tr>
<td>Simulated</td>
<td>467.6600</td>
<td>518.7390</td>
<td>445.9080</td>
</tr>
</tbody>
</table>

Table 6.5: Experimental and simulated peak plantar pressure values of the healthy subject’s foot during the midstance.

### Terminal stance

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>579.4129</td>
<td>523.7024</td>
<td>496.4181</td>
</tr>
<tr>
<td>Simulated</td>
<td>481.4090</td>
<td>404.3070</td>
<td>555.9640</td>
</tr>
</tbody>
</table>

Table 6.6: Experimental and simulated peak plantar pressure values of the healthy subject’s whole foot during the terminal stance.

### Preswing

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>557.1177</td>
<td>562.8222</td>
<td>505.5654</td>
</tr>
<tr>
<td>Simulated</td>
<td>500.8510</td>
<td>443.6640</td>
<td>257.1080</td>
</tr>
</tbody>
</table>

Table 6.7: Experimental and simulated peak plantar pressure values of the healthy subject’s whole foot during the preswing.

The tables below shows all the peak pressure values of the healthy subject’s hindfoot.

### Initial contact

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>430.4809</td>
<td>374.6199</td>
<td>367.9868</td>
</tr>
<tr>
<td>Simulated</td>
<td>164.0340</td>
<td>228.3370</td>
<td>193.7650</td>
</tr>
</tbody>
</table>

Table 6.8: Experimental and simulated peak plantar pressure values of the healthy subject’s hindfoot during the initial contact.
### 6.2. HEALTHY SUBJECT

<table>
<thead>
<tr>
<th>Loading Response</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>456.7744</td>
<td>569.0061</td>
<td>454.2560</td>
</tr>
<tr>
<td>Simulated</td>
<td>317.5120</td>
<td>392.6580</td>
<td>391.0990</td>
</tr>
</tbody>
</table>

Table 6.9: Experimental and simulated peak plantar pressure values of the healthy subject’s hindfoot during the loading response.

<table>
<thead>
<tr>
<th>Midstance</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>550.5045</td>
<td>609.3861</td>
<td>483.6333</td>
</tr>
<tr>
<td>Simulated</td>
<td>385.6440</td>
<td>459.2830</td>
<td>428.6320</td>
</tr>
</tbody>
</table>

Table 6.10: Experimental and simulated peak plantar pressure values of the healthy subject’s hindfoot during the midstance.

<table>
<thead>
<tr>
<th>Terminal stance</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>255.5920</td>
<td>134.9713</td>
<td>124.9088</td>
</tr>
<tr>
<td>Simulated</td>
<td>169,4339</td>
<td>93,1487</td>
<td>84,2617</td>
</tr>
</tbody>
</table>

Table 6.11: Experimental and simulated peak plantar pressure values of the healthy subject’s hindfoot during the terminal stance.

<table>
<thead>
<tr>
<th>Preswing</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>0.0000</td>
<td>52.6786</td>
<td>0.0000</td>
</tr>
<tr>
<td>Simulated</td>
<td>0.0000</td>
<td>7.9869</td>
<td>0.0000</td>
</tr>
</tbody>
</table>

Table 6.12: Experimental and simulated peak plantar pressure values of the healthy subject’s hindfoot during the preswing.
CHAPTER 6. RESULTS

Figure 6.5: A) Trend of pressures during stance for dynamic 1. B) Trend of pressures during stance for dynamic 2. C) Trend of pressures during stance for dynamic 3. D) Averaged trend of the pressures developed in the hindfoot for the three dynamics.
Figure 6.6: A) Trend of pressures during stance for dynamic 1. B) Trend of pressures during stance for dynamic 2. C) Trend of pressures during stance for dynamic 3. D) Averaged trend of the pressures developed in the whole foot for the three dynamics.
6.2.3 Experimental vs simulated data of the initial contact and the loading response phases in barefoot condition

In this test, the forces corresponding to the maximum pressure values in the whole foot and in the hindfoot from the beginning of the stance to the end of the loading response have been determined from the experimental data. Then the comparison with the predicted pressures has been done.

![Graph showing experimental vs simulated pressures of the whole foot.](image1)

(a) Experimental vs simulated pressures of the whole foot.

![Graph showing experimental vs simulated pressures of the hindfoot.](image2)

(b) Experimental vs simulated pressures of the hindfoot.

Figure 6.7: In the figure the histograms showing the experimental data compared to the simulated data of the whole foot and the hindfoot are displayed.

As it is shown in figure 6.7 model predicted plantar pressures are in good agreement with those measured during compressive loading of the heel. In the tables below all the simulated and experimental peak plantar pressures of the healthy subject from the beginning of the initial contact to the end of the loading response are listed.
### 6.2. HEALTHY SUBJECT

#### Table 6.13: Experimental and simulated peak plantar pressure values of the healthy subject’s whole foot from the beginning of the initial contact to the end of the loading response.

<table>
<thead>
<tr>
<th>Dynamic</th>
<th>Experimental</th>
<th>Simulated</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic 1</td>
<td>456.7744</td>
<td>329.8320</td>
</tr>
<tr>
<td>Dynamic 2</td>
<td>569.0061</td>
<td>442.0220</td>
</tr>
<tr>
<td>Dynamic 3</td>
<td>454.2560</td>
<td>393.4380</td>
</tr>
</tbody>
</table>

#### Table 6.14: Experimental and simulated peak plantar pressure values of the healthy subject’s hindfoot from the beginning of the initial contact to the end of the loading response.

<table>
<thead>
<tr>
<th>Dynamic</th>
<th>Experimental</th>
<th>Simulated</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic 1</td>
<td>456.7744</td>
<td>317.5120</td>
</tr>
<tr>
<td>Dynamic 2</td>
<td>569.0061</td>
<td>442.6580</td>
</tr>
<tr>
<td>Dynamic 3</td>
<td>454.2560</td>
<td>391.0990</td>
</tr>
</tbody>
</table>
6.2.4 Experimental vs simulated data in the frame of maximum medio-lateral force in barefoot condition

In this test the vertical forces corresponding to the maximum medio-lateral forces values in the whole foot and in the hindfoot have been used for the simulation.

![Histograms showing experimental vs simulated pressures of the whole foot and hindfoot](image)

(a) Experimental vs simulated pressures of the whole foot in the frame of maximum medio-lateral force.

(b) Experimental vs simulated pressures of the hindfoot in the frame of maximum medio-lateral force.

Figure 6.8: In the figure the two histograms showing the experimental data compared to the simulated data of the whole foot and the hindfoot are displayed.

As it is shown in figure 6.8 model predicted plantar pressures are in good agreement with those measured during compressive loading of the heel.

In the tables below all the simulated and experimental peak plantar pressures of the healthy subject in the frame of maximum medio-lateral force are listed.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>547.2099</td>
<td>529.3341</td>
<td>526.1576</td>
</tr>
<tr>
<td>Simulated</td>
<td>406.7100</td>
<td>553.5730</td>
<td>445.9080</td>
</tr>
</tbody>
</table>

Table 6.15: Experimental and simulated peak plantar pressure values of the healthy subject’s whole foot in the frame of maximum medio-lateral force.
6.2. HEALTHY SUBJECT

Table 6.16: Experimental and simulated peak plantar pressure values of the healthy subject’s hindfoot in the frame of maximum medio-lateral force.

<table>
<thead>
<tr>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>547.2099</td>
<td>487.9578</td>
</tr>
<tr>
<td>Simulated</td>
<td>373.1930</td>
<td>449.8390</td>
</tr>
</tbody>
</table>

6.2.5 Experimental vs Simulated healthy subject’s data in barefoot condition

In the histogram 6.9 all the percentages and the standard deviation of the difference between experimental and simulated data in all the conditions previously described are shown.

Figure 6.9: The histogram shows the percentage differences between simulated and experimental data of the healthy subject.
6.2.6 Comparison among the possible materials for the design of the insoles

In this test different materials suitable for the construction of insole has been tried. This in order to see what material ensures the maximum reduction of pressure and so that can be preferred for the realization of insoles.

(a) Comparison of the pressures to which the whole foot is subjected, obtained using insoles made of different materials.

(b) Comparison of the pressures to which the hindfoot is subjected, obtained using insoles made of different materials.

Figure 6.10: In the figures the simulated pressures obtained with the different materials in the three dynamics are shown.

The results are in agreement with what published in a previous work of Goske at all [33]. Indeed the material that presents the best ability to reduce the plantar pressures is the Poron Cushioning followed by the Microcell Puff Lite and finally the Microcell Puff.

In the tables below all the peak plantar pressure values for the different materials in the three dynamics for both the whole foot and the hindfoot are listed.
6.2. HEALTHY SUBJECT

Table 6.17: Simulate peak plantar pressures of the healthy subject’s whole foot using different insole materials.

<table>
<thead>
<tr>
<th>Dynamic</th>
<th>Microcell Puff</th>
<th>Microcell Puff Lite</th>
<th>Poron Cushioning</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic 1</td>
<td>202.4220</td>
<td>189.2540</td>
<td>177.2560</td>
</tr>
<tr>
<td>Dynamic 2</td>
<td>267.2790</td>
<td>253.0610</td>
<td>244.3010</td>
</tr>
<tr>
<td>Dynamic 3</td>
<td>163.1750</td>
<td>151.3680</td>
<td>141.3320</td>
</tr>
</tbody>
</table>

Table 6.18: Simulate peak plantar pressures of the healthy subject’s hindfoot using different insole materials.

<table>
<thead>
<tr>
<th>Dynamic</th>
<th>Microcell Puff</th>
<th>Microcell Puff Lite</th>
<th>Poron Cushioning</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic 1</td>
<td>224.9560</td>
<td>211.3610</td>
<td>200.2730</td>
</tr>
<tr>
<td>Dynamic 2</td>
<td>262.4510</td>
<td>248.3200</td>
<td>239.2330</td>
</tr>
<tr>
<td>Dynamic 3</td>
<td>246.8260</td>
<td>232.9030</td>
<td>222.9150</td>
</tr>
</tbody>
</table>
6.3 Diabetic subject

In this section the same simulation seen in paragraph 6.2 are reported for the diabetic subject. All the simulations at barefoot conditions and with the shoes has been processed using the specific models of the diabetic shown in figure 6.11.

Figure 6.11: The two figures at left show the models developed for the diabetic subject, the one on the top in barefoot conditions and the one on the bottom with the sketch of the shoes. The figures at right show the models with a force applied.
6.3. DIABETIC SUBJECT

6.3.1 Experimental vs simulated data in barefoot condition

In this test the maximum pressures borne by the whole foot and the hindfoot have been determined from the collected data for all the dynamics. Then the force corresponding to these pressures have been extracted from the experimental data to be applied in the FEM model in order to obtain the simulated pressures. Finally a comparison has been done and the results are shown in figure 6.12: once again the model predicted peak plantar pressures are in good agreement with those measured during compressive loading of the heel.

![Chart showing experimental vs simulated pressures of the whole foot in barefoot condition.](chart1)

![Chart showing experimental vs simulated pressures of the hindfoot in barefoot condition.](chart2)

Figure 6.12: In the figure the histograms showing the experimental data compared to the simulated data are displayed.

All the peak plantar pressures values are reported in tables 6.19 and 6.20.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>644.7859</td>
<td>686.469</td>
<td>682.3159</td>
</tr>
<tr>
<td>Simulated</td>
<td>574.1110</td>
<td>609.1360</td>
<td>612.4560</td>
</tr>
</tbody>
</table>

Table 6.19: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot.
<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>644.7859</td>
<td>686.4697</td>
<td>682.3159</td>
</tr>
<tr>
<td>Simulated</td>
<td>533.4730</td>
<td>601.3470</td>
<td>602.8160</td>
</tr>
</tbody>
</table>

Table 6.20: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot.
6.3.2 Experimental vs simulated data of all the phases of the stance phase in barefoot condition

In this test the maximum pressures borne by the whole foot and the hindfoot have been determined from the collected data for all the dynamics and for each phase of the stance described in paragraph 1.4. Then the force corresponding to these pressures have been extracted from the experimental data to be applied in the FEM model in order to obtain the simulated pressures. Finally a comparison has been done.

The results represented in figure 6.14 show again a good agreement between predicted and experimental pressures for all the cycle. Furthermore the percentages of the stance phase where the forces applied on the hindfoot are maximum (Figure 6.13) correspond to the ones where the pressure are maximum, and this is well documented by both experimental and simulated data. On the contrary the trend of the forces to which the whole foot is subjected is different.

![Image of graph showing forces applied on hindfoot and whole foot](image)

Figure 6.13: In the figure the trend of the vertical force applied on the hindfoot and on the whole foot are shown.

The tables below shows all the peak pressure values of the diabetic’s whole foot.

<table>
<thead>
<tr>
<th>Initial contact</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>315.9235</td>
<td>539.2517</td>
<td>534.3008</td>
</tr>
<tr>
<td>Simulated</td>
<td>240.0770</td>
<td>433.8380</td>
<td>410.3860</td>
</tr>
</tbody>
</table>

Table 6.21: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot during the initial contact.
Table 6.22: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot during the loading response.

<table>
<thead>
<tr>
<th>Loading Response</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>561.6408</td>
<td>639.9083</td>
<td>600.9760</td>
</tr>
<tr>
<td>Simulated</td>
<td>481.9400</td>
<td>541.2460</td>
<td>548.4480</td>
</tr>
</tbody>
</table>

Table 6.23: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot during the midstance.

<table>
<thead>
<tr>
<th>Midstance</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>644.7859</td>
<td>686.4697</td>
<td>682.3159</td>
</tr>
<tr>
<td>Simulated</td>
<td>567.6550</td>
<td>609.1360</td>
<td>612.456</td>
</tr>
</tbody>
</table>

Table 6.24: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot during the terminal stance.

<table>
<thead>
<tr>
<th>Terminal stance</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>543.4530</td>
<td>404.8506</td>
<td>509.9891</td>
</tr>
<tr>
<td>Simulated</td>
<td>558.4230</td>
<td>531.0360</td>
<td>414.1900</td>
</tr>
</tbody>
</table>

Table 6.25: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot during the preswing.

<table>
<thead>
<tr>
<th>Preswing</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>450.1812</td>
<td>511.5313</td>
<td>404.9697</td>
</tr>
<tr>
<td>Simulated</td>
<td>568.3820</td>
<td>509.4070</td>
<td>552.6820</td>
</tr>
</tbody>
</table>

The tables below shows all the peak pressure values of the diabetic’s hindfoot.

Table 6.26: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot during the initial contact.

<table>
<thead>
<tr>
<th>Initial contact</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>315.9235</td>
<td>539.2517</td>
<td>534.3008</td>
</tr>
<tr>
<td>Simulated</td>
<td>140.0770</td>
<td>432.5070</td>
<td>410.3860</td>
</tr>
</tbody>
</table>

Table 6.27: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot during the loading response.

<table>
<thead>
<tr>
<th>Loading Response</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>561.6408</td>
<td>639.9083</td>
<td>600.9760</td>
</tr>
<tr>
<td>Simulated</td>
<td>481.9400</td>
<td>540.5300</td>
<td>545.5250</td>
</tr>
<tr>
<td>Midstance</td>
<td>Dynamic 1</td>
<td>Dynamic 2</td>
<td>Dynamic 3</td>
</tr>
<tr>
<td>-----------------</td>
<td>-------------</td>
<td>-------------</td>
<td>-------------</td>
</tr>
<tr>
<td>Experimental</td>
<td>644.7859</td>
<td>686.4697</td>
<td>682.3159</td>
</tr>
<tr>
<td>Simulated</td>
<td>558.4230</td>
<td>601.3470</td>
<td>602.8160</td>
</tr>
</tbody>
</table>

Table 6.28: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot during the midstance.

<table>
<thead>
<tr>
<th>Terminal stance</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>418.5058</td>
<td>256.7596</td>
<td>190.9137</td>
</tr>
<tr>
<td>Simulated</td>
<td>329.5310</td>
<td>139.5330</td>
<td>148.5410</td>
</tr>
</tbody>
</table>

Table 6.29: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot during the terminal stance.

<table>
<thead>
<tr>
<th>Preswing</th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>9.5407</td>
<td>10.3489</td>
<td>13.2515</td>
</tr>
<tr>
<td>Simulated</td>
<td>0.0000</td>
<td>8.7777</td>
<td>9.3338</td>
</tr>
</tbody>
</table>

Table 6.30: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot during the preswing.
Figure 6.14: A) Trend of pressures during stance for dynamic 1. B) Trend of pressures during stance for dynamic 2. C) Trend of pressures during stance for dynamic 3; D) Averaged trend of the pressures for the three dynamics.
Figure 6.15: A) Trend of pressures during stance for dynamic 1. B) Trend of pressures during stance for dynamic 2. C) Trend of pressures during stance for dynamic 3; D) Averaged trend of the pressures for the three dynamics.
6.3.3 Experimental vs simulated data of the initial contact and the loading response phases in barefoot condition

In this test, the forces corresponding to the maximum pressures which act on the whole foot and the hindfoot from the beginning of the stance to the end of the loading response have been extracted from the experimental data. Then the comparison with the predicted pressures has been done.

Figure 6.16: In the figure the histograms showing the experimental data compared to the simulated data of the whole foot and the hindfoot are displayed.

As it is shown in figure 6.16 model predicted plantar pressures are in good agreement with those measured during compressive loading of the heel. In the tables below all the simulated and experimental peak plantar pressures of the diabetic subject from the beginning of the initial contact to the end of the loading response are listed.
### Table 6.31: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot from the beginning of the initial contact to the end of the loading response.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Experimental</strong></td>
<td>588.6744</td>
<td>676.0568</td>
<td>626.3715</td>
</tr>
<tr>
<td><strong>Simulated</strong></td>
<td>510.1190</td>
<td>569.8420</td>
<td>572.6840</td>
</tr>
</tbody>
</table>

### Table 6.32: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot from the beginning of the initial contact to the end of the loading response.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Experimental</strong></td>
<td>588.6744</td>
<td>676.0568</td>
<td>626.3715</td>
</tr>
<tr>
<td><strong>Simulated</strong></td>
<td>508.7080</td>
<td>569.8420</td>
<td>567.6550</td>
</tr>
</tbody>
</table>
6.3.4 Experimental vs simulated data in the frame of maximum medio-lateral force in barefoot condition

In this test the vertical forces corresponding to the maximum medio-lateral forces to which the whole foot and the hindfoot are subjected, have been used for the simulation.

![Whole Foot Pressure Comparison](image)

(a) Experimental vs simulated pressures of the whole foot in the frame of maximum medio-lateral force.

![Hindfoot Pressure Comparison](image)

(b) Experimental vs simulated pressures of the hindfoot in the frame of maximum medio-lateral force.

Figure 6.17: In the figure the two histograms showing the experimental data compared to the simulation data of the whole foot and the hindfoot are displayed.

As it is shown in figure 6.17 model predicted plantar pressures are in good agreement with those measured during compressive loading of the heel.

In the tables below all the simulated and experimental peak plantar pressures of the diabetic subject in the frame of maximum medio-lateral force are listed.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>492.8895</td>
<td>586.7787</td>
<td>398.2722</td>
</tr>
<tr>
<td>Simulated</td>
<td>417.2630</td>
<td>503.7410</td>
<td>456.8240</td>
</tr>
</tbody>
</table>
Table 6.33: Experimental and simulated peak plantar pressure values of the diabetic’s whole foot in the frame of maximum medio-lateral force.

<table>
<thead>
<tr>
<th></th>
<th>Dynamic 1</th>
<th>Dynamic 2</th>
<th>Dynamic 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Experimental</td>
<td>609.9611</td>
<td>586.7787</td>
<td>420.1815</td>
</tr>
<tr>
<td>Simulated</td>
<td>447.0460</td>
<td>498.8350</td>
<td>381.6110</td>
</tr>
</tbody>
</table>

Table 6.34: Experimental and simulated peak plantar pressure values of the diabetic’s hindfoot in the frame of maximum medio-lateral force.

6.3.5 Experimental vs Simulated diabetic subject’s data in barefoot condition

In the histogram 6.18 all the percentages and the standard deviation of the difference between experimental and simulated data in all the conditions previously described are shown.

Figure 6.18: The histogram shows the percentage differences between simulated and experimental data of the diabetic subject.
6.3.6 Comparison among the possible materials for the design of the insoles

In this test different materials suitable for the construction of insoles has been tried. This in order to see what is the material that ensures the maximum reduction of pressure and so that can be preferred for the realization of insoles.

(a) Comparison of the pressures to which the whole foot is subjected, obtained using insoles made of different materials

(b) Comparison of the pressures to which the hindfoot is subjected, obtained using insoles made of different materials

Figure 6.19: In the figures the simulated pressures obtained with the different materials in the three dynamics are shown.

The results are once again in agreement with what published in a previous work of Goske at all [33]. Indeed the material that presents the best ability to reduce the plantar pressures is the Poron Cushioning followed by the Microcell Puff Lite and finally the Microcell Puff.

In the tables below all the peak plantar pressure values for the different materials in the three dynamics for both the whole foot and the hindfoot are listed.
6.3. **DIABETIC SUBJECT**

<table>
<thead>
<tr>
<th></th>
<th>Microcell Puff</th>
<th>Microcell Puff Lite</th>
<th>Poron Cushioning</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic 1</td>
<td>371.6370</td>
<td>361.7890</td>
<td>354.2120</td>
</tr>
<tr>
<td>Dynamic 2</td>
<td>393.0870</td>
<td>383.3130</td>
<td>374.9930</td>
</tr>
<tr>
<td>Dynamic 3</td>
<td>394.8420</td>
<td>385.0780</td>
<td>376.6990</td>
</tr>
</tbody>
</table>

Table 6.35: Simulate peak plantar pressures of the diabetic’s whole foot using different insole materials.

<table>
<thead>
<tr>
<th></th>
<th>Microcell Puff</th>
<th>Microcell Puff Lite</th>
<th>Poron Cushioning</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dynamic 1</td>
<td>363.7840</td>
<td>353.9160</td>
<td>346.6100</td>
</tr>
<tr>
<td>Dynamic 2</td>
<td>388.2620</td>
<td>378.4730</td>
<td>370.3060</td>
</tr>
<tr>
<td>Dynamic 3</td>
<td>389.1390</td>
<td>379.3520</td>
<td>371.1570</td>
</tr>
</tbody>
</table>

Table 6.36: Simulate peak plantar pressures of the diabetic’s hindfoot using different insole materials.
6.4 Healthy vs diabetic subject

After the presentation of all the test for each subject, in this section a comparison between them will be shown.

6.4.1 Experimental vs simulated data in barefoot condition

Figure 6.20 shows the comparison between the pressures developed from experimental and simulated data for both the subject.

(a) Experimental vs simulated pressures of the whole foot for both the diabetic and the healthy subject.

(b) Experimental vs simulated pressures of the hindfoot for both the diabetic and the healthy subject.

Figure 6.20: In the figure the histograms showing the experimental data compared to the simulated data for both the subjects are displayed.

The pressures that act on the diabetic foot are higher then the ones that act on the healthy foot. Furthermore the model of the diabetic subject gives best prediction compared to the other one: difference among predicted and experimental pressures is lower in the diabetic subject compared to the healthy one.
6.4.2 Experimental vs simulated data of all the phases of the stance phase in barefoot condition

Figure 6.21 shows the comparison for all the dynamics and for hindfoot and whole foot, of all the experimental and simulated pressures for both the diabetic and the healthy subjects. Once again the pressures of the diabetic are higher than the one of the healthy subject but the trend during the different phases is similar. Furthermore the model of the diabetic subject gives again best prediction compared to the other one.

Figure 6.21: Trend of pressures during stance for dynamic 1, 2 and 3 for the whole foot and the hind foot of the diabetic and the healthy subject are displayed together.

6.4.3 Other tests

Also for the other test, the comparison between the diabetic and the healthy subject gives the same results: the pressure that the first bears are always higher. Furthermore the model of the diabetic foot gives the best results in terms of agreement between experimental and simulated pressures. Figure 6.22 shows for instance the comparison for the maximum pressures in the initial contact and loading response phases.
(a) Experimental vs simulated pressures of the whole foot for both the diabetic and the healthy subject.

(b) Experimental vs simulated pressures of the hindfoot for both the diabetic and the healthy subject.

Figure 6.22: In the figure the histograms showing the experimental data compared to the simulated data of the whole foot and the hindfoot are displayed.
Conclusions

The work presented in this thesis has been performed at the Laboratory of Bioengineering of the Movement of the University of Padua.

The aim was to obtain two 2D Finite Element Models of the hindfoot for both a diabetic and an healthy subject in order to be able to see the consequences in terms of peak plantar pressures, of the application of forces during gait. Models have been obtained starting from the MRI acquisition of the foot and then the geometry has been reconstructed by means of Simpleware and Rhino 4.0. The modelling phase has then been done with Abaqus/CAE.

Kinetic and kinematic data have been collected by means of a stereophotogrammetric system (BTS, Padova), two pressures platforms (Imagortesi, Piacenza) and two Bertec force plates (FP4060-10) temporally and spatially synchronized. Compared to the literature the thesis aims to test the performance of a subject specific model on which the patient’s own data are applied. Furthermore for the first time foot subsegments forces have been used for the simulations.

Different situations have been compared in both subjects using the forces corresponding to the whole foot or to the hindfoot:

1. experimental vs simulated peak plantar pressures in barefoot condition;
2. experimental vs simulated peak plantar pressures of all the phases of the stance phase;
3. experimental vs simulated peak plantar pressures from the beginning of the initial contact to the end of the loading response;
4. experimental vs simulated peak plantar pressures corresponding to the frame of max medio-lateral force.

For the healthy subject the difference between the experimental and the simulated data in the three dynamics of the whole foot in the first situation, varies
from 23% to 32%; of the hindfoot it varies from 11% to 30%. Furthermore
the difference between the experimental and the simulated pressures in the three
dynamics of the whole foot in the third simulation varies from 13% to 28%;
of the hindfoot it varies from 14% to 30%. Finally the difference between the
experimental and the simulated data in the three dynamics of the whole foot
 corresponding to the frame of max medio-lateral force varies from 5% to 26%;
of the hindfoot it varies from 8% to 32%.

For the diabetic subject the difference between the experimental and the sim-
ulated data in the three dynamics of the whole foot in the first situation varies
from 10% to 11%; of the hindfoot it varies from 12% to 17%. Furthermore the
difference between the experimental and the simulated data in the three dynamics
of both the whole foot and the hindfoot in the third simulation varies from 9%
to 16%. Finally the difference between the experimental and the simulated data
in the three dynamics of the whole foot varies from 14% to 15%; of the hindfoot
it varies from 9% to 27%.

The best matching has been obtained with the diabetic’s model. A possible
explanation could be given by the different size of mesh used. Indeed for the dia-
abetic’s model, larger elements had to be chosen because of problems of excessive
distortion of the tissue during the applications of the forces.

By taking into account that a possible solution in order to prevent ulcer for-
mation on the foot of diabetic subject is to prescribe plantar foot orthosis, the
effect of different sole materials has been tested. In this contest results confirmed
the literature [33].

The next step to be achieved is to increase the number of subject in order to
obtain a larger samples of subject specific fem models. Furthermore some param-
eters can be changed as the mesh size, in order to check if the accuracy of the
prediction changes. Finally an analysis of the position of the slice of the heel in
the space should be performed to test if a change of its angulation affects also
the predicted pressures.
Bibliography


[56] BERTEC. Website: http://bertec.com