Gait Analysis in Patients Recovering from Total Joint Replacement Using Body Fixed Sensors

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Mestrado Integrado em Bioengenharia

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Abstract

Locomotion is one of the most important functions of Human Being. It implies a complex interaction of joint movements controlled by muscle activity and positional perception which allows a human to walk at a desired speed and direction. This process is usually called gait control.

Human gait can be affected by several problems and diseases such as osteoarthritis, a progressive musculoskeletal disorder characterized by a gradual loss of articular cartilage. Due to the ageing process that reduces the ability of the cartilaginous tissue to withstand loads and stresses, this disorder is the most common cause of long-term disability for people over 65. The surgical procedure indicated for patients affected by severe osteoarthritis is the Total Knee Replacement (TKR) which aims to relieve pain and restore range of motion (ROM). However, after surgery, many individuals still experience an antalgic gait pattern and the motor pattern of the operated limb may remain slightly dysfunctional even years after the intervention and the contra-lateral limb may also adopt compensation strategies.

In order to evaluate the changes in post-operative stage of patients that underwent this surgery and to overcome the limitations of current analysis strategies, new community based solutions must be developed. In this project, a mobile phone combined with several inertial measurement units (IMU) and an electromyographic (EMG) monitoring system was purposed as a potential instrument to perform gait analysis. Several strategies and algorithms were then evaluated and adapted to analyze and compare the spatio-temporal signals as well as the EMG signals in a control group and a test group.

Inertial sensors were used to extract gait related parameters, including gait phases, step duration, cadence, step length, walking velocity and knee angle associated parameters. Muscle activation patterns were accessed using electromyographic data.

Results suggest that this approach can be used as a major strategy to evaluate changes in post-operative stage of patients affected by osteoarthritis. Moreover, this strategy revealed statistical significance changes in some gait parameters between the control group and the test group.

Keywords: Knee, Osteoarthritis, Total Knee Replacement, Human Gait, Gait Analysis, Inertial Measurement Unit, Electromyography
Resumo

A locomoção é uma das funções mais importantes do ser humano. Implica uma complexa interação de movimentos articulares controlada pela atividade muscular e percepção posicional, que permite ao ser humano andar a uma velocidade e direção desejada. Este processo é normalmente chamado de controlo da marcha.

A marcha humana pode ser afetada por vários problemas e doenças tais como a osteoartrite, uma desordem musculosquelética caracterizada pela perda progressiva de cartilagem articular. Devido ao processo de envelhecimento que reduz a capacidade do tecido cartilaginoso para suportar cargas e tensões, esta desordem é a causa mais comum de incapacidade a longo prazo para as pessoas com mais de 65 anos. O procedimento cirúrgico indicado para pacientes afetados por osteoartrite severa é a artroplastia, que visa sobretudo aliviar a dor e restaurar a amplitude do movimento. No entanto, após a cirurgia, muitas pessoas ainda apresentam um padrão de marcha antálgica que pode prolongar-se por meses ou mesmo anos e também o membro contra lateral pode adotar estratégias de compensação.

Para avaliar as mudanças na fase pós-operatória dos pacientes submetidos a esta cirurgia e ultrapassar as limitações das estratégias atuais de análise, devem ser desenvolvidas novas soluções. Neste projeto, um smartphone combinado com várias unidades de medição inercial (IMU) e um sistema de monitorização eletromiográfica foi proposto como um instrumento potencial para realizar análise de marcha. Várias estratégias e algoritmos foram avaliados e adaptados para analisar e comparar os parâmetros espacial-temporais e eletromiográficos para ambos, pacientes submetidos à cirurgia e grupo de controlo.

Os sensores inerciais foram usados para extrair parâmetros da marcha, tais como fases da marcha, duração do passo, cadência, comprimento do passo, ângulo do joelho, etc. Padrões de ativação muscular foram obtidos usando os dados eletromiográficos.

Os resultados obtidos sugerem que esta abordagem pode ser usada como uma estratégia de avaliação das mudanças da marcha, na fase pós-operatória dos pacientes afetados por osteoartrite. Além disso, esta estratégia evidenciou diferenças estatisticamente significativas em alguns parâmetros nos dois grupos testados.

Keywords: Joelho, Osteoartrite, Artroplastia, Marcha Humana, Análise de Marcha, Unidade de medição inercial, eletromiografia
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Last but not the least, I would like to thank my family: my parents and my grandparents for supporting me spiritually throughout writing this thesis and my life in general.

Pedro Pereira
“Learn from yesterday, live for today, hope for tomorrow. The important thing is not to stop questioning.”

Albert Einstein
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<td>Acc</td>
<td>Accelerometer</td>
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<tr>
<td>ANOVA</td>
<td>Analysis of variance</td>
</tr>
<tr>
<td>CHTMAD</td>
<td>Centro Hospitalar de Trás-os-Montes e Alto Douro</td>
</tr>
<tr>
<td>EMG</td>
<td>Electromyography</td>
</tr>
<tr>
<td>FEUP</td>
<td>Faculdade de Engenharia da Universidade do Porto</td>
</tr>
<tr>
<td>FhP AICOS</td>
<td>Fraunhofer Portugal research center for assistive information and communication solutions</td>
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<tr>
<td>IMU</td>
<td>Inertial Measurement Unit</td>
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<tr>
<td>IR</td>
<td>Infrared</td>
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<tr>
<td>LED</td>
<td>Light emission diode</td>
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<td>RMS</td>
<td>Root Mean Square</td>
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<tr>
<td>ROM</td>
<td>Range of motion</td>
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<tr>
<td>SNR</td>
<td>Signal-to-noise ratio</td>
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<tr>
<td>TKEO</td>
<td>Teager-Kaiser energy operator</td>
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<td>TKR</td>
<td>Total Knee Replacement</td>
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Chapter 1

Introduction

1.1 Overview and Motivation

Locomotion is one of the most important functions of Human Being. It implies a complex interaction of joint movements controlled by muscle activity and positional perception which allows a human to walk at a desired speed and direction. This process is usually called gait control.

Human gait can be affected by several problems and diseases. Osteoarthritis, the most common type of arthritis, is a progressive musculoskeletal disorder characterized by a gradual loss of articular cartilage. Due to the ageing process that reduces the ability of the cartilaginous tissue to withstand loads and stresses, this disorder is the most common cause of long-term disability for people over 65.

The surgical procedure indicated for patients affected by severe osteoarthritis is the Total Knee Replacement (TKR), a widely used intervention in the management of knee diseases, which aims to relieve pain and restore range of motion. However, after surgery, many individuals still experience an antalgic gait pattern and the motor pattern of the operated limb may remain slightly dysfunctional even years after the intervention. Moreover, the contra-lateral limb may also adopt compensation strategies.

The increasing prevalence of TKR highlights the need to appropriately assess post-operative outcome of this procedure [1]. Traditional methods include observational methods and self-questionnaires which are used in the follow-up of post-operative patients. Gait analysis is a tool that has been used by researchers to quantitatively measure functional outcome following TKR. It has been proposed that gait analysis is valuable in the clinical management of patients undergoing TKR through its ability to monitor forces through the knee [2].

Clinical gait analysis can be defined as the measurement, processing and systematic interpretation of bio-mechanical parameters that characterize human locomotion and ability to detect limitations in motion in order to identify appropriate procedures for rehabilitation [3].

Clinical gait analysis allows the physician to quantitatively assess the degree to which an individual’s gait was affected by a disease already diagnosed. This process involves the measurement of fundamental gait parameters, the processing of these data into valid and useful information.
and the systematic interpretation of the collected information. The objective is to understand the cause of gait abnormalities, such as a decrease in a joint movement, as well as to recommend an appropriate treatment based in each patient situation. Thus, clinical gait analysis is currently an evaluation tool and not a diagnostic tool, not only in orthopedics (mainly used in amputees who use prosthesis) but also in other diseases such as polio, cerebral palsy, multiple sclerosis, rheumatoid arthritis and muscular dystrophy [3,4].

Despite the potential usefulness of gait analysis, there are marked discrepancies in the research methods that have been reported. Variations in subject characteristics, prosthetic designs and methodology of gait analysis make comparison of findings between studies difficult [4].

Therefore, it’s of great interest the development of a quantitative solution with a different methodology, capable of analysing more than one kind of information, and reporting about the recovery process after the surgery.

1.2 Objectives

The objective of this dissertation is the development of a mobile system capable of evaluating gait changes in the postoperative phase of patients who underwent a Total Knee Replacement surgery, with simultaneous information from both spatio-temporal and electromyographic signals, using IMU and EMG sensors. The innovative aspects of this approach are the fact that it is a mobile system and it uses both spatio-temporal and electromyographic signals.

1.3 Structure

Apart from this Introduction, this dissertation consists of four chapters.

Chapter 2 presents a literature review focused on the gait analysis methodologies as well as some theoretical concepts related with human gait.

In Chapter 3, the methodology and tools used in this work are described. The dataset used for validation and its statistical comparison are presented.

In Chapter 4, the results of the implemented methodology are presented and discussed, including not only the performance of this system but also the statistical results. Finally, in the last Chapter the major conclusions are summarized, and some future work are also summed up.
Chapter 2

Background and Literature Review

Gait analysis involves many variables and the choice of appropriate ones implies to know the human gait basics. Thus an anatomical description of the lower limbs is presented in this chapter, both skeletal and muscular constitution, as well as the processes taking place in the gait cycle, mainly those ones affected by osteoarthritis and his consequent surgery. Gait analysis techniques are introduced based on a literature review.

2.1 Lower Limb Anatomy

The contents of this section and the next one are essential taken from [5].

In order to understand human gait it’s of great importance to consider the parts of the body involved in this process. The lower limbs and its anatomical and physiological constitution are therefore the object of study in this section.

Bones, joints and muscles are the three anatomical and physiological determinant factors for gait execution. The principle of a joint is the movement between two or more bones promoted by the muscles [5]. Without joints, we would not be able to move anyway. However not every joint provides motion because the structure of the joint relates directly to their degree of movement; in some cases two or more bones are “connected” but no movement is allowed, as we can see in the temporo-parietal junction of the skull.

In spite of having these different kinds of joint, only diarthrosis (free movable) will be considered in this study due to the limb structure [5][6]. The main articulations of lower limbs are the hip joint, knee joint and ankle joint. The hip joint is responsible for transferring the weight of the head, trunk and upper limbs to the lower limbs. The hip is capable of a wide range of motions, including flexion, extension, abduction, adduction, rotation and circumduction. The knee joint makes the connection between the thigh and the leg (upper and lower part of the limb, respectively). It is a complex joint which allows flexion, extension, and a small rotation of the leg [5]. The ankle joint allows the movement between the leg and the foot and is an important articulation since it allows a wide range of motions of the foot although imposing crucial limitations necessary to the equilibrium and gait control.
Beyond those joints, there are joints between the bones of the foot which are numerous and complex but will not be focused in this dissertation.

2.1.1 Skeletal System

All bones in the human body have influence in the gait process [7]. However, only bones of the pelvis and lower limb will be considered in this dissertation. These bones (Figure 2.1 and 2.2) support the body and are essential to stand, walk or run normally [5,7].

The right and left coxal bones join each other anteriorly and the sacrum posteriorly to form the pelvic girdle. The pelvis is formed by the pelvic girdle and the sacrum, a bone resulted of the fusion of five sacral vertebrae. The pelvic girdle is divided in three parts: the ilium, ischium and pubis. The intersection occurs at the center of the acetabulum, a fossa placed on the lateral surface of each coxal bone where the lower limb articulates with the trunk, more precisely the head of the femur.

The only real movement between the bones of the pelvis occurs in the sacroiliac joint and this movement is generally very small in adults. For the purpose of gait analysis, it is reasonable to consider the pelvis as a single rigid structure [7].

The thigh contains only one bone, the femur. The femur is the longest bone of the body and has a spherical head in his proximal end that articulates with the acetabulum of the pelvis establishing the hip joint. Its distal part has two condyles (medial and lateral), smooth and curvy surfaces that articulate with the leg and form the knee joint.
2.1 Lower Limb Anatomy

The patella, or kneecap, is a small flat bone inserted within the tendon of the femoral quadriceps muscle, covering the anterior part of the knee joint.

The leg is the part of the lower limb placed between the knee and the ankle and is composed of two bones: tibia and fibula. The tibia is much larger than the fibula and supports most of the weight of the leg. The proximal end of the tibia has two flat surfaces which articulate with the condyles of the femur. The fibula does not articulate with the femur but has a small head which articulates with the proximal tibia. The tibia and fibula are basically parallel bones which contact
with each other in superior and inferior part.

The proximal portion of the foot is composed of seven bones. The astralagus (or talus) connects with the tibia and fibula forming the ankle joint. The calcaneus is located below the astralagus and supports it. In the distal portion of the foot we have the metatarsal bones and phalanges.

### Muscular System

The muscles of the lower limb can be divided into the muscles involved in movement of the thigh, the leg and the foot [5].

Several thigh muscles have their origin on the coxal bone and fit into the femur (Figure 2.3). These muscles can be divided into three groups: anterior, posterolateral and deep. The anterior muscles, iliacus and psoas major, provide the flexion of the hip. As these muscles share a common insertion and produce the same movement, they are often called iliopsoas being for the most of the work when someone does sit-ups. Posterolateral muscles responsible for moving the thigh are the gluteal muscles and the tensor fasciae latae. The gluteus maximus is the muscle that contributes with the largest part of muscle mass of gluteal muscles. The deep hip muscles, as well as the gluteus maximus, work as lateral rotators of the thigh. The gluteus medius, gluteus minimus, and tensor fasciae latae are medial hip rotators, while the medius and minimus gluteus help tilt the pelvis contributing for the maintenance of the trunk in a straight posture during walking, being fundamental in gait control [5].

![Figure 2.3: Right hip and thigh muscles: anterior view (left); posterior view (right)](image)

Besides the hip muscles, some muscles of the thigh have their origin on the coxal bone and can cause movement of the thigh. There are three sections of muscles: the anterior section flexes the
2.1 Lower Limb Anatomy

hip and extends the knee; the medial section adducts the thigh; the posterior section extends the hip and flexes the knee. The anterior thigh muscles are the quadriceps femoris and the sartorius. More specifically, the quadriceps femoris is actually composed of four muscles: rectus femoris, vastus lateralis, vastus medialis and vastus intermedius. The quadriceps muscles make the extension of the knee while the rectus femoris also flexes the hip since it is long enough to be involved on the hip and knee joints. The quadriceps femoris is the heaviest muscle group on the anterior thigh and has an insertion in the patellar tendon, on and around the patella. The other thigh muscle, sartorius, is the longest muscle in the human body, across the lateral side of the hip to the medial side of the knee. Its contraction causes the flexion of the hip and the knee and laterally rotates the thigh, action required to cross the legs. The medial thigh muscles are mainly involved in adduction of the thigh, although some of these muscles are also responsible for lateral rotation of the thigh and flexion or extension of the hip. Additionally, the gracilis helps in the flexion of the knee. The posterior thigh muscles group, also called the hamstring, are composed by the biceps femoris, the semimembranosus and the semitendinosus. Their tendons are easily observed and palpated on the medial and lateral posterior part of the knee when slightly bent.

The leg muscles responsible for moving the ankle and foot can be divided into three groups, each located in separate compartments in the leg: anterior, posterior and lateral (Figure 2.4 and 2.5). The anterior muscles of the leg are extensors that are involved in dorsiflexion and foot inversion or eversion and extension of the toes. The superficial muscles of the posterior compartment are the gastrocnemius and the soleus which form the bulge of the calf. These muscles join with the small plantaris muscle to form the common calcaneal tendon, usually known as the Achilles tendon, and are involved in plantar flexion of the foot. The deep muscles of the posterior compartment are responsible for the flexion and inversion of the foot as well as toe flexion. The lateral muscles are primarily responsible for foot eversion, but also help in plantar flexion. These muscles are the fibularis brevis and the fibularis longus. The first one has an insertion on the fifth metatarsal bone and contributes for flexion and evasion of the foot. The second one inserts onto the first metatarsal bone and medial cuneiform. Both tendons can be observed on the lateral side of the ankle joint.

In summary, our anatomical and physiological constitution is designed to allow movement. The bones are responsible for structure and support the body weight whereas the muscles work together to control the joints between bones [5].

The hip, knee and ankle joint are the main joints of the lower limb. Since the knee is the joint focused in this project, in the next section a more detailed view of this articulation is presented.
2.2 Knee Joint

The knee joint is conventionally classified as a modified hinge joint positioned between the femur and the tibia \[5\]. Actually, it is a complex ellipsoid joint allowing flexion, extension, and a slight
rotation of the leg. The distal end of the femur has two great ellipsoid surfaces with a deep fossa between them. The femur articulates with the proximal end of the tibia, composed of flattened and smooth surfaces, with a crest called the intercondylar eminence in the center. The tibial plateau is built up by menisci, thick articular disks of fibrocartilage, which extend the articular surface. The fibula articulates only with the lateral side of the tibia, not with the femur.

The major ligaments which promote knee stability are the cruciate and collateral ligaments. Two cruciate ligaments extend between the fossa of the femur and the intercondylar eminence of the tibia, preventing anterior and posterior displacement of the tibia relative to the femur. The medial and lateral collateral ligaments stabilize the medial and lateral sides, of the knee, respectively. Articulation strength is also provided by popliteal ligaments and tendons of the thigh muscles that extend around the knee (Figure 2.6).

![Figure 2.6: Right knee joint](image)

The knee is surrounded by small fluid-filled sacs called bursae (Figure 2.7). The largest bursa
Background and Literature Review

is the suprapatellar, a superior extension of the joint capsule that allows the anterior thigh muscles to move over the distal end of the femur. Other bursae protecting the knee are the subcutaneous prepatellar bursa and the deep infrapatellar bursa, as well as the popliteal bursa, the gastrocnemius bursa, and the subcutaneous infrapatellar bursa.

![Figure 2.7: Right knee joint: photograph of anterior view (left); sagittal section (right) [5]](image)

2.2.1 Osteoarthritis

In a general view, arthritis is the inflammation of a joint, leading to pain, swelling and stiffness of the joint. Any joint in the body may be affected by the disease being related to over 100 causes, including infectious agents, metabolic disorders, trauma, and immune disorders. Knee arthritis can make it hard to do many daily activities, such as walking or climbing stairs. It is a major cause of lost work time and a serious disability for many people.

Osteoarthritis, the most common type of arthritis, is a progressive musculoskeletal disorder characterized by gradual loss of articular cartilage (Figure 2.8). As the cartilage of the junction wears away, it becomes frayed and rough, and the protective space between the bones decreases. This can result in bone rubbing on bone, and produce painful bone spurs. It is known that advancing age leads to a gradual degeneration of a joint. However this process can be delayed with exercise [5, 8].

Injuries to the medial side of the knee are much more common than injuries to the lateral side for several reasons [5]. First, the lateral (fibular) collateral ligament strengthens the joint laterally and is stronger than the medial (tibial) collateral ligament. Second, severe blows to the medial side of the knee are far less common than blows to the lateral side of the knee. Finally, the medial meniscus is fairly tightly attached to the medial collateral ligament and is damaged 20 times more often in knee injuries than the lateral meniscus, which is thinner and not attached to the lateral collateral ligament.
2.2 Knee Joint

Figure 2.8: Visual differences between a normal knee and a knee with osteoarthritis [8]

2.2.2 Total Knee Replacement

Currently, the most effective treatment for end-stage knee osteoarthritis is the Total Knee Replacement. This procedure is effective in providing pain relief and improving function in knee osteoarthritis patients with the assistance from post-operative rehabilitation programs [9][10].

This procedure requires an orthopedic surgeon to make precise measurements and skillfully remove the diseased portions of the bone, in order to shape the remaining bone to accommodate the knee implant (Figure 2.9) [11]. The surgeon makes an incision across the front of the knee to gain access to the patella. In a traditional TKR, the incision is usually about 20 to 25 cm long. Once the knee is open, the surgeon rotates the patella outside the knee area and makes a split incision in the quadriceps muscle. Once the knee joint is exposed, the surgeon will carefully measure the bones and make precise cuts using special instruments. Firstly, the damaged bone and cartilage from the end of the femur is cut away. Then a metal femoral component is attached to the end of the femur and bone cement is used to seal it into place. The same is done to the top of the tibia, shaping the bone to fit the metal tibial component, called tray. Once the tray is in place, the surgeon snaps in a polyethylene (medical-grade plastic) insert to sit between the tibial tray and the femoral component, and act as a kind of buffer. The patella is readjusted before returning to its normal position, ensuring a proper fit with the rest of the implant. Finally, the surgeon will bend and flex the knee to ensure that the implant is working correctly, and that alignment, sizing, and positioning is suitable. To complete the procedure, the surgeon will close the incision with stitches or staples, and then bandage it.

After any type of surgery for arthritis of the knee, there is a period of recovery. Recovery time and rehabilitation depends on the type of surgery performed. However, physical therapy to help the patient regain strength in the knee and to restore range of motion must be done. Depending upon the procedure, the patient may need to wear a knee brace, or use crutches or a cane for a time. In most cases, surgery relieves pain and makes it possible to perform daily activities more
Despite experiencing significant reductions in pain, many TKR patients do not achieve normal joint function when walking following surgery [4, 9, 10, 12, 13]. In most cases, gait remains slower than asymptomatic controls, with the treated knee exhibiting abnormal biomechanics [4, 13]. It can be caused by muscle handling namely those ones connected to the patella which needs to be “set aside” during surgery in order to access the interior of the articulation. Additionally, the procedure usually includes the use of Poly(methyl methacrylate) that provides prosthesis adhesion to the bone. This bone cement solidification is an exothermic reaction which means that can cause some damage in surrounding tissues.

By the other hand, an abnormal pre-surgery gait has been reported to affect the post-surgery gait pattern due to the patient attempt to reduce pain [14].

Given the importance of maintaining adequate mobility in people following TKR, identifying specific gait impairments following surgery may also help to define rehabilitation strategies. Therefore, in the next section a brief introduction to Human gait and gait analysis is presented.

2.3 Human Gait

Human gait is a very complicated coordinated series of movements. It is the basic way of human locomotion and is defined as a biphasic forward propulsion of center of gravity of the human body, in which there are alternate sinuous movements of different segments of the body with least expenditure of energy. This process involves not only a perfect coordination of the muscular-skeletal system, but also the contribution and interaction of the nervous system in planning, initiating and maintaining gait [15, 16].

Human gait analysis is, basically, the systematic study of human locomotion, using the eye in a rudimentary way or draw upon adequate instrumentation for measuring body movements, body
2.3 Human Gait

mechanics or the activity of the muscles. Clinical gait analysis is used to assess, plan and treat individuals with conditions affecting their ability to walk. Therefore, it provides objective criteria to assess performance and strength following TKR [7].

This section is organized to explain the gait cycle, especially the way it is divided, and which are the characteristic movements and events of each phase.

2.3.1 Gait Cycle

A gait cycle is defined as the time period or the sequence of events or movements during locomotion in which one foot contacts the ground to when that same foot again contacts the ground, and is also known as a stride. In average, a gait cycle has the duration of 1 second, where 60% is due to the stance phase and 40% of swing phase. While any event can be chosen to define the beginning of the gait cycle, the initial contact of a foot with the ground is usually used [4, 7].

Each cycle is divided into two main phases (Figure 2.10):

- The stance phase is the part while the foot remains in contact with the ground of each gait cycle. It is initiated by heel strike and ends with toe off of the same foot;
- The swing phase, which is the part while the foot is not in contact with the ground, is initiated with toe off and ends with heel strike.

Each gait phase can also be divided in sub-phases. The stance phase can be divided into five stages: Initial contact, Loading response, Mid stance, Terminal stance and Toe off; and the swing phase is divided into three sub-phases: Initial swing, Mid swing and Terminal swing (Table 2.1).

In each gait cycle, there are two periods of double support and two periods of single support, divided by the two legs (Figure 2.11).

2.3.2 General Movements

In the gait process, there are different contributions of different segments of the body, specifically in the lower limb. The foot can present a neutral position especially when is totally supported on the floor, a plantar flexion since the initial contact until the mid-stance, and dorsiflexion (Figure 2.12).

There is flexion of the leg during the stance phase and since the initial swing to mid-swing. Its extension takes place in the remaining periods of the cycle. The thigh is flexed whenever there is advancement of the lower limb and extended when performing the opposite movement.

All the movements described above occur in the sagittal plane (antero-posterior); however, there are also important movements in other planes, like pelvic rotation (on the vertical axis) which occurs earlier in the leg during the swing and then in the mid-stance. Pelvic rotation is maximal when the heel touches the ground [17, 18].

For a better understanding of which muscles and joints were involved in each phase of the gait cycle, a diagram of the muscles and joints that are active during the sub-phases of stance phase is
Table 2.1: Phases and sub-phases of gait cycle [7,17,18]

<table>
<thead>
<tr>
<th>Sub-phases</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Stance phase</strong></td>
<td></td>
</tr>
<tr>
<td>Initial contact (Heel strike)</td>
<td>The moment when the foot touches the ground. Typically, the heel is the first part of the foot touching the ground. The opposite leg is at the end of terminal stance sub-phase.</td>
</tr>
<tr>
<td>Loading response (foot flat)</td>
<td>The doubled supported period of time when the whole foot is on the ground. It ends when the opposite foot rises (toe-off). The weight of the body shifts between the legs.</td>
</tr>
<tr>
<td>Mid stance</td>
<td>The first half of single support, beginning with the elevation of the opposite leg (which is in mid swing), ending when the leg is approximately vertical.</td>
</tr>
<tr>
<td>Terminal stance</td>
<td>Begins when the heel of the foot (now in a posterior position) rises and ends when the heel of the front foot touches the ground.</td>
</tr>
<tr>
<td>Pre-swing (Toe off)</td>
<td>Begins with initial contact to the front foot and ends when the other foot rises, initiating the swing phase. There is again a short period of time of double support.</td>
</tr>
<tr>
<td><strong>Swing phase</strong></td>
<td></td>
</tr>
<tr>
<td>Initial swing</td>
<td>Begins when the foot leaves the ground and starts moving forward and ends when the other foot is on the end of a medium support.</td>
</tr>
<tr>
<td>Mid swing</td>
<td>The period of time when the leg advances by balance until is localized anterior to the body and the tibia is vertical.</td>
</tr>
<tr>
<td>Terminal swing</td>
<td>The anterior leg continues to move forward, in order to get into a position anterior to the thigh. The sub-phase ends when the front foot touches the ground, beginning a new cycle.</td>
</tr>
</tbody>
</table>
Gravity is responsible for leg movement during a great part of the swing phase.

In general, gait cycle has two phases (stance and swing), each one divided into subphases. The division into subphases is based mainly on foot’s movement and position. In each subphase, the movement is performed by different muscles which require different efforts in the joints. The combined knowledge of the gait cycle phases and the anatomical components involved in the movement are fundamental for understanding gait analysis and its techniques.
Figure 2.11: Gait cycle time dimensions [16]

Figure 2.12: Sagittal movements of some limb segments: (a) ankle movements; (b) knee movements [5]
2.3 Human Gait

Figure 2.13: Posterior and lateral views of muscles activity during a gait cycle [19]
2.4 Gait Analysis

People have been thinking about how they walk since the earliest times. Aristotle can be attributed with the earliest recorded comments regarding the manner in which humans walk. After him, a high number of individuals have worked on this field of study, contributing for theories formulation and, more recently, for computerized analysis techniques development [20].

Gait analysis is applied in the assessment of human gait and the accumulation of data that describes and characterizes it. Gait analysis helps distinguish between normal and pathological gait, estimate the course of an orthopedic problem, and assess the need for prosthetic and orthotic devices for the upper and lower limbs. Also it’s of great importance for athletes who look for a quantitative analysis in order to improve their performance [21].

Several techniques have been developed for gait analysis, differing in the type of information they offer, as well as in their methodology. Some of them are more applicable in a research laboratory, but less appropriate for routine clinical practice, and offer much information regarding human gait.

Gait analysis can use both qualitative and quantitative methods. Qualitative methods include scoring systems where clinicians and researchers have access to the effectiveness of the surgical intervention seen by the patient. These scoring systems, such as the Knee Society Score (KSS) and the Oxford Knee Score (OKS), are questionnaires designed to obtain a perspective of the patient regarding the surgical procedure [22]. As a qualitative method, it lacks scientific rigor and is considerably subjective. In an even more rudimentary way, observational methods are used in the follow-up of post-operative patients (for example at 3, 6 and 12 months after surgery).

Concerning scientific research, the main outcomes are summarized in objective and measurable variables, called gait parameters. Usually, studies try to obtain these parameters comparing between two groups of individuals, in order to evaluate gait changes or physical dysfunctions.

Surprisingly, for most of the reported outcomes, the findings were inconsistent between the studies. This may relate to discrepancies in the research methodologies, prosthetic designs and variations in subject characteristics [4]. Further research evaluating usefulness of findings from gait analysis may assist in determining which of the gait methods provide the most useful information about the gait.

Thus, regarding the purpose of this project, it becomes important to know the previous studies performed in this area and if there are really differences in gait of individuals who underwent a TKR. Thereby, a search for articles on gait analysis in patients following TKR was made.

This section presents the state of the art of gait analysis techniques and a description of the main and most used motion sensors and the gait parameters measured. Moreover, electromyography studies applied to gait analysis are also included.

2.4.1 Gait parameters

Most of the gait parameters measured have the influence of several factors related to the patient and the surgery itself. These factors include body mass, age, prosthesis design, cruciate and collateral
Regarding objective and measurable analysis, the most common techniques reported in several studies [4, 7, 12, 13, 22, 23, 24, 25, 26, 27, 28] include the measurement of physical parameters, listed in the table below (Table 2.2).

In these referred studies the gait parameters are, in general, divided in four categories: spatio-temporal, kinematics, kinetics and electromyography. Spatio-temporal parameters include spatial and temporal information about gait movement, such as step number, step length, stance time, etc. Kinematics parameters are related to joint angular information, such as range of motion, joint angle at several time-points, etc. Kinetics parameters report about the forces involved in the production of movements, while electromyography parameters aim to access muscle activity during gait.
Table 2.2: Summary of the most common physical parameters found in literature [4, 7, 12, 13, 22, 23, 24, 25, 26, 27, 28]

<table>
<thead>
<tr>
<th>Spatio-temporal</th>
<th>Kinematics</th>
<th>Kinetics</th>
<th>EMG</th>
</tr>
</thead>
<tbody>
<tr>
<td>Step number</td>
<td>Range of motion</td>
<td>Ground reaction force</td>
<td>Muscular activation time</td>
</tr>
<tr>
<td>Speed</td>
<td>Angle at initial contact</td>
<td>Flexion/extension moment</td>
<td>Muscular activation power</td>
</tr>
<tr>
<td>Step/stride length</td>
<td>ROM during loading</td>
<td>Abduction/adduction moment</td>
<td>Muscular max. activation peak</td>
</tr>
<tr>
<td>Step/stride time</td>
<td>Max. angle during loading</td>
<td>Internal/external rotation moment</td>
<td></td>
</tr>
<tr>
<td>Cadence</td>
<td>Max. angle during stance</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Angular velocity</td>
<td>Max. angle during swing</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stance time</td>
<td>Max. abduction angle</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Swing time</td>
<td>Max. adduction angle</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Single limb support time</td>
<td>Abduction/adduction ROM</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Double limb support time</td>
<td>Internal/external rotation ROM</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Time of weight acceptance</td>
<td>Angle at maximum angular velocity</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Step width</td>
<td>Step height</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Time of weight acceptance</td>
<td>Step height</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Time of weight acceptance</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
2.4 Gait Analysis

2.4.2 Monitoring systems

Stationary Systems

Nowadays, the gold standard of gait analysis is based on a stationary marker-based motion capturing gait lab [22]. These systems are able to capture human movements using a determined number of cameras and markers. Usually, the cameras are disposed in the room so that they can record the movement of all markers distributed by the human’s body.

An example of these systems applied to gait analysis is the VICON (Oxford Metrics, Oxford, England) [29], an infrared marker-tracking system composed by ten high-resolution cameras outfitted with IR optical filters, an array of IR LEDs, and a set of reflective dots. The IR radiation emitted by the LEDs is reflected by the dots, strategically disposed in the region of interest of the subject. This reflected radiation is then capture by the cameras and used to construct a three-dimensional representation of the markers.

Another example of this technology is Qualysis [30], similar to VICON, composed of retro-reflective ball markers that can be identified by the cameras surrounding the subject. This allows a 3-D digital reconstruction of the positions of the moving human limbs [31].

Ground reaction forces measurement, another analysis method, can be obtained using special force platforms like the AMT (Advanced Mechanical Technology, Newton, MA) [32] or the Kistler Type 9260AA (Kistler, Winterthur, Switzerland) [33]. By the use of a piezoelectric force sensors, these plates measure the forces applied to its top surface as a subject stands or steps.

There are many commercial systems and gait analysis laboratories which use different combinations of the above mentioned sensors and technologies, or similar ones. Some examples of systems situated and calibrated in laboratory or clinical environments are CONTEMPLAS (Figure 2.14): Clinical gait analysis based on a walkway, Tekscan: Pressure Mapping, GRAIL: Gait Real-time Analysis Interactive Lab, from Motek Medical and BTS GAITLAB [34].

![Figure 2.14: Example of a stationary system: BTS GaitLab configuration. (1) infrared videocameras; (2) inertial sensor; (3) GRF measurement walkway; (4) wireless EMG; (5) workstation; (6) video recording system; (7) TV screen; (8) control station. [34]](image-url)
In general, the aforementioned technologies are fast and very accurate. However they present some disadvantages comparing to mobile systems:

- Need of specific hardware and software
- Cost of software, equipment and personnel
- Space requirement
- Cannot reproduce daily living activities
- Physically demanding (necessity of active patients) . . .

Mobile Systems
In contrast with the stationary systems, inertial motion capture technology has been reported in several studies [22, 35]. These systems are based on miniature inertial sensors and sensor fusion algorithms. The motion data of the inertial sensors is usually transmitted wirelessly (Bluetooth) to a control device (Laptop, Smartphone, etc.), where the motion is firstly recorded and stored for a further analysis. Most inertial sensors are equipped with an accelerometer, gyroscope and magnetometer, being attached to the patient using some kind of tape in strategic positions. This placement ensures that the sensors are positioned identically for different test scenarios and minimize motion relative to the skeleton underneath. A preliminary calibration is usually needed to ensure that data is normalized between different subjects. Some of the systems reported by [35, 36, 37] use only one pelvic sensor, which allows the acquisition of some spatio-temporal parameters. However, the information about knee and other joints function is very limited. Their motion and stability, as well as prosthesis performance, cannot be evaluated.

The more sophisticated systems with more than one sensor are able to monitor knee range of motion and related parameters. One of these reported systems uses three SHIMMER 2R sensors (Shimmer, Realtime Technologies, Dublin, Ireland) [22] attached with therapeutic tape to three different anatomic parts: lumbosacral junction, lateral thigh and in the medial aspect of the shank. With this system, the lack of knee and joint function information is surpassed.

More recently have been developed an removable on-shoe device, GaitShoe [38], that can be used for continuous and real-time monitoring of gait. As this device is fixed on typical athletic shoes, its design was thought in a way to not interfere with the normal gait pattern. By the use of IMU sensors, this device provides information about the three-dimensional motion, position, and pressure distribution of the foot [38]. However, one downside of this technology is that does not give a complete analysis of the joints movement and consequently, the gait process [21].

Moreover, successful gait analysis systems based on wearable sensors have been commercialized, such as the widely used Xsens MVN [34], which uses 17 inertial trackers situated in the chest, upper and lower limbs to perform motion capture and six degrees of freedom tracking of the body with a wireless communicated suit. Another commercial package is the wireless M3D gait analysis system developed by Tec Gihan Co [34], which uses motion sensors on the lower leg, the
2.4 Gait Analysis

thigh, the waist and the back and wearable force plates on the toes and the heels. A similar wire-
less system, composed of 9 inertial sensors situated in the lower limbs and wearable force plates
with wireless force sensors, was presented by INSENCO Co. under the name Human Dynamics
Analysis (HDA) [34]. Figure 2.15 shows these two systems.

![Image of gait analysis systems](image)

Figure 2.15: Sagittal movements of some limb segments: (a) Commercial WS system based on
inertial sensors: Xsens MVN; (b) wearable system based on inertial sensors and wearable force
plates [34]

All systems previously refereed need some kind of software, not only to record data but also
to synchronize all signals and analyse the information. Stationary methods and marketed products
usually use self-developed software which are displayed as part of the system, while the other
systems use developed algorithms in IDE’s, using a programming language like Java, Python or
R.

Electromyography

The most comprehensive gait analysis includes the mentioned parameters as well as simultaneous
electromyographic monitoring of the muscles involved in the gait cycle [23, 24, 25]. This has
the advantage of providing information about muscular work in order to correlate the gait cycle
subphases with the moment that muscles are activated. Furthermore, EMG signals can be used to
measure different gait characteristics, such as correlating joint angular motion with EMG signals
recorded at the same time to see if one set of data can explain the other. The amplitude of EMG
signals derived during gait may be interpreted as a measure of relative muscle tension and it has
been found that it increases with increased walking speed and that the EMG activity is minimized
with subjects walking at a comfortable speed [34].

The basic of electromyography is the detection of the electrical potential generated by muscle
cells when these cells are electrically or neurologically activated. So, muscular activity can be
obtained from the collected signal, as well as the degree of activation, rate of force production, number of motor units recruited, etc.

The detection of electric signals is made with electrodes, firstly with intramuscular and more recently with surface electrodes [39]. Intramuscular electrodes are used to study individual motor units and have a better signal/noise ratio, but apart from being invasive, they are very restrictive to a small area of analysis. Surface electrodes are painless and detect a larger area of activity but are much more susceptible to noise and crosstalk (interference from other muscles). It has been shown that application of surface electromyography (SEM) is a useful in non-invasive assessment of relevant pathophysiological mechanisms potentially hindering the gait function such as changes in passive muscle-tendon properties (peripheral, non-neural component), paresis, spasticity, and loss of selectivity of motor output in functionally antagonist muscles [40].

Electric detection requires at least two electrodes: positive and negative. Nevertheless a third electrode is often used as reference. In most cases, the connection between the electrodes and the processing unit is made with wires and the connection to the laptop/smartphone by Bluetooth. To obtain data is necessary to take into account various factors such as skin preparation, the place of the electrodes and the orientation of the electrodes [41].

Usually the frequency of an EMG signal can go from 5 to 450Hz and the amplitude from 0.1 to 1 mV when using surface electrodes. Due to the small amplitude of the signals and the usage of electrodes, signals must be filtered to avoid low and high frequencies noise contamination. The main source of signal noise is the power supply, which introduces a 50-60Hz noise that can be minimized with a notch filter. The analysis of the signal should be preceded by a rectification (transformation of the signal to its module) in order to estimate the average value of the signal.

Some studies apply a root mean square (there’s no need of rectification) that evidences in a more precise way the behavior of the motor unit during activation [40].

Some of the selected muscles include the quadriceps femoris (specifically the rectus femoris, vastus medialis and vastus lateralis), hamstring, gastrocnemius and tibialis anterior [23, 24, 25]. This selection allows the understanding of the lower limb movement: flexion and extension of the leg.

**Literature Review**

A group of studies have analysed the pre and postoperative gait in order to detect which gait parameters show more significant differences in each patient.

Wilson *et al.* [23] reported that no significant differences were found in spatio-temporal parameters using both VICON and AMTI technologies. Of all four gait parameters categories, kine- matics and kinetics stood out, as the most significant differences were detected in the knee ROM of the patients that underwent knee surgery. However, Smith *et al.* [14] studies, who also used VICON and AMTI technologies to collect data in patients pre and postoperative stages, reported significant differences not only in kinetics but also in spatio-temporal parameters, where an increase of walking speed was found.
In Levinger et al. [27], both stationary motion system and force plates were reported. They obtained significantly lower velocity with shortened stride length and reduced cadence for the surgical group before the surgery compared to a control group. Moreover, all spatio-temporal parameters revealed to remain significantly lower at 12 months post-surgery compared to the control group, despite the pain relief and function. However, only few significant improvements were found between the surgical group before and after the surgery, due to the compensatory response before the surgery.

In Calliess et al. [22], post-surgical improvements were found in walking speed, cadence and step length in spite of differing in every patient. Kinematics values revealed to be significantly identical in patients before and after the surgery.

In Fuchs et al. [25], a stationary camera-based motion system, two force plates and an EMG monitoring system were used. Nevertheless the objective was to determine gait differences between two different prosthesis, an inconsistency between quantitative outcomes and clinical scores was observed between control and TKR patients.

In Mandeville et al. [28], a stationary camera-based motion system coupled with two force plates (AMTI) were used. Post-surgical improvements were found in gait velocity, stride length, step width and stride time. Despite this, significant differences were still found between these results and the standard values of the control group.

In Schache et al. review [26], TKR patients had reduced strength of multiple limb muscle groups when compared to control groups, particularly evident for the quadriceps and hamstring muscle groups.

In Hilding et al. [2], a gait analysis was performed before operation and at six months and two years after TKR. On all three occasions they found significant differences in the mean sagittal plane moments of the knee joint and peak flexion moments.

In Davis et al. [3], a video-based motion collection system, force plates and EMG transducers were used in order to obtain spatio-temporal, kinematics and kinetics information of the gait. Its performance in providing quantified assessments of human locomotion was evaluated.

In Bade et al. [9], TKR patients performed significantly worse for all measures compared to healthy adults, except for single-limb stance time. Persistent impairments and functional limitations 6 months after TKR with standard rehabilitation were reported.

In McClelland et al. [12], knee kinematics of patients following TKR and unimpaired controls during comfortable and fast walking speeds was studied. The TKR group walked with significantly reduced cadence, stride length, less knee flexion during stance and swing phases, less knee extension during stance phase and less peak knee external rotation than controls.

In Bolanos et al. [24], a stationary motion analysis system, two force plates and an EMG monitoring system were used. The patient group walked slower than the control group and had shorter step length. Knee range of motion of test group was lower than in the control group.

In Huddlestone et al. [37], a stationary video-camera based system was used to test the performance of a mobile sensor-based system in identifying ordinary activities. Measures of knee
motion and gait were assessed, revealing that mobile systems can be useful as a clinical tool for evaluating knee function.

In Wang et al. [42], a stationary camera-based system (VICON), two force plates (AMTI) and an isokinetic dynamometer were used to study the knee mechanics during gait after bi-compartmental knee replacement. The surgical knee exhibited less peak torque and initial abduction moment than both the non-involved and control limbs. The non-involved limb had less knee extension at stance and greater knee extensor moment during push-off than both the surgical and control limbs. No differences were found for other typical knee mechanics among the surgical, non-involved, and control limbs during walking.

In Benedetti et al. [43], a lower limb functional evaluation after a TKR was performed using a stationary camera-based system and a force platform for the acquisition of kinematic and kinetic variables. EMG signals were recorded from eight muscles: the ipsilateral and contralateral erector spinae, the gluteus medius, the rectus femoris, the medial and lateral hamstrings, the gastrocnemius, and the tibialis anterior. The stance phase was significantly increased after the surgery, while stride length, cadence and speed of progression were significantly reduced. Kinematics parameters were in general reduced.

2.5 Proposed method

Only in the most severe cases (when people do not adapt or experience pain) patients are redirected to physical therapy appointments. However, the most rudimentary methods are not an effective way in determining patient status. Hence the importance of this study where there is a quantitative and more complete analysis, combining spatio-temporal parameters, kinematic, and also EMG.

The proposed system is a mobile-based low-cost solution which does not need expert personnel to be handled and can be used anywhere without laboratory restrictions. This solution will facilitate and assist in the understanding of postoperative improvements and if people can restore the "normal range" of the gait.

In the next Chapter, the proposed system and its methodology is presented.
Chapter 3

Methodology

The methodology used to extract several gait parameters from the IMU and EMG data is detailed in this chapter. The first part includes the Section 3.1 and Section 3.2 where the used sensors are described and the methods for recording data reported. The second part includes the Sections 3.3 and 3.4 where the dataset collection procedures and the statistical evaluation used to compare between two groups of subjects are described.

3.1 Data acquisition

Since the proposed system works with two kinds of signals, inertial and electromyographic, it becomes necessary to use two sub-systems responsible for collecting simultaneously the two different sources of information.

![Figure 3.1: Overview of the system](image)
Methodology

The first sub-system is composed by a smartphone and 4 IMU devices, while the second sub-system is formed by a laptop and an EMG monitoring device.

3.1.1 Inertial Sensors

IMU is an electronic device that usually combines three sensors - accelerometer, gyroscope and magnetometer [44]. With a IMU it is possible to measure the gravitational forces acting in the device, usually using three axis. The accelerometer is a sensor that measures the linear acceleration caused by the movement or the earth gravitational acceleration. The gyroscope is a sensor that measures the angular velocity. Lastly, a magnetometer is a sensor that measure the local earth magnetic field vector, proving additional information about orientation [45].

Using the data provided by IMU it is possible to estimate the posture or movement of the human body. Currently, several applications in medical field are using this technology [44, 46], e.g., comparing the movements between normal persons and pathological persons, or in Human Computer Interface where the human movements are used to control electronic devices. The usage of this device is becoming increasingly popular, due to their low cost, small size, light weight, and limited power requirements compared to traditional approaches of quantitative motion analysis [47].

In this project, only the accelerometer sensor is used. Some preliminary tests revealed that results are acceptable using the accelerometer in order to not compromise the acquisition of four IMU's simultaneously. An accelerometer basically uses the fundamentals of Newton’s Laws of Motion, which says that the acceleration is proportional to the force acting on the body [34]. The signal obtained with accelerometers has two components, a gravitational acceleration component (static) - provides information on the postural orientation of the subject - and a body acceleration component (dynamic) - provides information on the movement of the subject [34].

The chosen IMU was the Pandlets Fraunhofer Portugal developed at FhP AICOS Institute (Figure 3.2). It consists of a small box with dimensions of 28.4x 28.4mm and 10 mm height.

Figure 3.2: Pandlets Fraunhofer Portugal (IMU used in this project)

This device includes several sensors, namely an accelerometer with 16 bits of resolution, a sampling frequency of 4 kHz (100Hz was used in this project) and 2 to 16 g range. It is connected to a smartphone (LG Nexus 5) by Bluetooth using an improved version of an Android application developed by FhP AICOS, capable to connect to more than one device at the same time (in this case, 4 devices). Figure 3.3 shows the accelerometer axis of the IMU’s as well as their positions.
These positions were chosen based on literature [22] in order to avoid some noise from the skin movement. In Figure 3.4 some layouts of the Android application used to record data are shown.

Figure 3.3: IMU’s positioning and Acc axis of the two different orientations: (1) Right thigh, (2) Right shank, (3) Left thigh, (4) Left shank

![Image](image1.png)

Figure 3.4: Android application used for data recording: (a) initial interface; (b) sensors selection

![Image](image2.png)

The data recording application used in the Smartphone had already been developed by FhP
Methodology

AICOS, however it was suitable to record only from one device. It was necessary to modify the application in order to accept to 4 devices. Moreover, devices position can be established each time it starts to record. The user is free to choose how many devices he wants to connect simultaneously and the directory name to save the files. Then, a list of all available devices is displayed and the user can choose the ones to connect and record the data. Finally, information from the accelerometer is saved as a ".csv" file to be opened and processed in a computer.

The four IMU units were positioned as shown in figure 3.3 and 3.6 and described below:

- Device 1 - Right lateral thigh.
- Device 2 - Right anterior shank.
- Device 3 - Left lateral thigh.
- Device 4 - Left anterior shank.

The usage of four devices, two on each leg, is due to the need of determining the knee angle, being necessary to place one sensor on each leg segment (thigh and shank). These positions are intended to reduce artifacts and noise from the surface where they are attached, since skin in these positions is more steady.

3.1.2 Electromyography Monitoring System

BITalino, shown in Figure 3.5, was the device used to collect the data from the muscles activity. It is a non-invasive low-cost modular toolkit based on Arduino made explicitly for applications using physiological signals [48]. This device has 8 channels - 6 analogical and 2 digital - and supports different types of sensors, namely EMG. The EMG sensors record at a sampling frequency of 1000 HZ and a bandwidth between 20 and 400 Hz [49].

Figure 3.5: BITalino (EMG monitoring system used in this project)
3.2 Inertial Data Analysis

The interface between the sensors and the skin is made by surface electrodes specially designed by its developers with a conductive and adhesive hydrogel, to maximize electrical conduction. Its dimensions are 24 mm in diameter and 1 mm in thickness \[50\].

The electrodes were placed as shown in figure 3.6 and described below:

- Channel 1 - Right posterior shank (Gastrocnemius muscle).
- Channel 2 - Right anterior thigh (Rectus femoris muscle).
- Channel 3 - Left posterior shank (Gastrocnemius muscle).
- Channel 4 - Left anterior thigh (Rectus femoris muscle).
- Channel 5 - Right posterior thigh (Hamstring muscles).
- Channel 6 - Left posterior thigh (Hamstring muscles).

![Figure 3.6: Positioning of the IMU’s and EMG electrodes](image)

The connection between BITalino and the laptop is made by Bluetooth, using a free-software named OpenSignals, developed to acquire data and show it in real-time.

3.2 Inertial Data Analysis

After data acquisition being made by the smartphone, files are moved to a laptop to be processed. This processing includes several steps as shown in figure 3.7.

Signal processing is made in Python with the software PyCharm.

3.2.1 Pre-processing

The pre-processing starts with some steps which allows all the signals to be normalized not only in time but also in orientation.
In a first stage, as the sensors placed on the thighs are rotated 90 round the x-axis relative to the ones placed on the shanks (Figure 3.3), there is a need to rotate the y-axis and z-axis in order to have the same reference orientation in all sensors.

Next, the time is normalized so the first sample is at $t = 0$. This is made subtracting from each timestamp the value of the first timestamp.

The baseline is removed subtracting from each sample the average of the first 200 samples corresponding to 2 seconds (for each separate axis). This assures that small deviations to the standard orientation is eliminated.

Finally, all signals are truncated to the same number of samples.

A low-pass Butterworth filter (sixth order) is applied in order to remove noise and smooth the signal. The applied filter is adapted to the sampling frequency of 100Hz with a cutoff frequency of 4Hz. This filter removes the high frequencies caused by noisy movements (Figure 3.8).

### 3.2.2 Signal Segmentation

The signal segmentation is performed in order to remove the initial and final part of the signal, so that only the part when the individual is walking is maintained. This is achieved by firstly calculating the linear acceleration of the sensor placed on the shank (using Equation 3.1) because the steps are easier identified.

$$lin_{acc} = \sqrt{x^2 + y^2 + z^2} - 1 \quad (3.1)$$

Two signals are obtained, one from each limb. These signals are filtered with a low-pass Butterworth filter (cutoff frequency = $2Hz$), as the normal walking doesn’t usually exceed this frequency and artifacts are then removed.

Establishing a threshold equal to the average of the signal, a peak detection easily detects the steps taken in a time interval (Figure 3.9).
3.2 Inertial Data Analysis

Figure 3.8: Filtering process on the 3-axial signal: (a) before; (b) after

The segmentation is then done, more precisely cutting the signal between the first and the last step, assuring that the beginning and the end is excluded and not analysed in the following steps.

3.2.3 Angle Calculation

The assessment of the knee angle as well as some gait parameters implies knowing the angle of the thigh and shank. For this purpose is necessary to calculate the angle of the different IMU’s from the data of accelerometers.

The inner product of vectors can be used to calculate the angle between the gravitational vector measured by the accelerometer and the initial orientation with the gravitational field pointing
Methodology

Figure 3.9: Step detection based on linear acceleration: Horizontal blue line are the threshold (average); vertical red lines are steps downwards along the z-axis. The inner product of vectors is defined by the Equations 3.2 and 3.3:

\[ A \cdot B = ||A|| \cdot ||B|| \cos \theta \quad (3.2) \]

\[ A \cdot B = A_1B_1 + A_2B_2 + \cdots + A_nB_n \quad (3.3) \]

where \( n \) is the dimension of the vector space.

If \( A(Ax, Ay, Az) \) is the accelerometer reading and \( B(0, 0, 1) \) is the gravitational force, then:

\[ A \cdot B = A_z \]

and

\[ ||A|| \cdot ||B|| = \sqrt{Ax^2 + Ay^2 + Az^2} \]

Solving the Equation 3.2 in order to \( \theta \), and applying the Equation 3.4 to the signals of each accelerometer, the angles of the thigh and shank of both limbs are obtained.

\[ \text{Angle} = \arccos \left( \frac{Az}{\sqrt{Ax^2 + Ay^2 + Az^2}} \right) \quad (3.4) \]

where \( Ax \) represents the x axis, \( Ay \) the y axis and \( Az \) the z axis of the accelerometer.

Finally, the knee angle can also be estimated subtracting the thigh angle with the shank angle, with the Equation 3.5 (Figure 3.10):

\[ \text{Knee\_angle}[i] = \text{Thigh\_angle}[i] - \text{Shank\_angle}[i] \quad (3.5) \]
3.2 Inertial Data Analysis

3.2.4 Event Detection

The event detection steps were used to detect relevant moments of the gait cycle.

After the detection of the step, the toe of and the heel strike events were also detected, seeing that these moments are crucial in determining some gait parameters, such as the stance phase and double support. For its determination the accelerometer signals previously converted to the respective angle were used.

The toe off is detected by following these steps:

- Detection of minima of the thigh angle

- Detection of maxima of the thigh angle, with the threshold equal to the sum of the mean with the standard deviation

- Deletion of false minima: in the case two or more minima are detected between two maxima, only the last one will be maintained.

The heel strike is detected by following these steps:

- Detection of maxima of the shank angle

- Detection of minima of the shank angle, with the threshold equal to the difference of the mean by the standard deviation

- Deletion of false maxima: in the case two or more maxima are detected between two minima, only the first one will be maintained.

The previous listed steps for both events detection is exemplified on figure 3.11.
3.2.5 Parameters Estimation

Based on literature, the spatial-temporal parameters listed in Table 3.1 are extracted from different sources: linear acceleration of the shank accelerometers (described in section 3.2.3), thigh, shank and knee angles (described in section 3.2.2) and events detected (described in section 3.2.4).

These parameters were chosen since in some studies, after the surgery, people still remain with walking problems: lower speed, reduced knee angle, smaller step length, etc.

Spatio-temporal parameters and foot events were validated using a video recording. A subject was asked to walk while being filmed. In the end, the video was analysed and some spatial and temporal measurements were made.
Table 3.1: Chosen parameters and its description

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Steps</td>
<td>Number of maxima detected on the linear acceleration of the shank accelerometers</td>
</tr>
<tr>
<td>Step duration (s)</td>
<td>Average time interval between maxima detected on the linear acceleration of the shank accelerometers</td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>Quotient between 60s and the step duration</td>
</tr>
<tr>
<td>Velocity (m/s)</td>
<td>Quotient between the walked distance and the total time walked (last step - initial step)</td>
</tr>
<tr>
<td>Step length (m)</td>
<td>Product of the velocity by the step duration</td>
</tr>
<tr>
<td>Maximum knee angle</td>
<td>Average of the maxima detected in the knee angle signal</td>
</tr>
<tr>
<td>Minimum knee angle</td>
<td>Average of the minima detected in the knee angle signal</td>
</tr>
<tr>
<td>Knee angle ratio*</td>
<td>Ratio of the maximum knee angle between the two limbs</td>
</tr>
<tr>
<td>Gait Cycle Time</td>
<td>Average time interval between maxima detected on the knee angle signal</td>
</tr>
<tr>
<td>Gait Cycle Time ratio*</td>
<td>Ratio of the Gait Cycle Time between the two limbs</td>
</tr>
<tr>
<td>Stance/swing proportion</td>
<td>Quotient between the stance time (time interval between the heel strike and toe off in a gait cycle) and the Gait Cycle Time</td>
</tr>
<tr>
<td>Stance/swing ratio*</td>
<td>Ratio of the Stance/swing proportion between the two limbs</td>
</tr>
<tr>
<td>Double limb support time</td>
<td>Quotient between the double support time (time interval between the heel strike and toe off in common for the two limbs) and the Gait Cycle Time</td>
</tr>
<tr>
<td>Limbs angle during heel strike</td>
<td>Difference between the thigh angle of the two limbs during the heel strike</td>
</tr>
<tr>
<td>Limbs angle during heel strike ratio*</td>
<td>Ratio of the limbs angle during the heel strike between the two limbs</td>
</tr>
</tbody>
</table>
3.3 EMG Data Analysis

After data acquisition being made by BITalino using the OpenSignals software and stored in the laptop, recorded data is processed. This processing includes several steps as shown in figure 3.12. Signal processing is made at the same time as the inertial data.

![Figure 3.12: EMG data analysis scheme](image)

3.3.1 Pre-processing

With the objective of distinguish the active movements of the non-active, pre-processing is needed in order to eliminate useless information. In most of the cases we need to perform some noise reduction in order to obtain a compact representation of the gait pattern, that will facilitate the segmentation task.

The EMG signal has high levels of interference originated in different sources [51, 52] including the noise caused by the overlap of the muscles, by motion artifacts, the environmental noise and noise caused by the electronic equipment itself. All can be reduced with analog or digital filters [52, 53].

Using a high pass filter it is possible to attenuate the noise caused by motion artifacts. The corner frequency of the high pass filter is frequently set at 10 Hz and generally should be set no higher than 20 Hz [52]. Another possible way to attenuate motion artifacts is to apply averaging, reducing the undesired motion artifacts.

The AC power supply frequency (60 Hz in USA or 50 Hz in Europe) and its harmonics can result in a power line interference signal which can be much larger than EMG itself. This type of noise can be reduced by shielding the recorder device and applying a notch filter centred at the fundamental frequency (50 or 60 Hz) [52].

In the pre-processing a low-pass Butterworth filter (sixth order) is applied to all EMG channels in order to remove noise and smooth the signal. The applied filter is suitable to the sampling
3.3 EMG Data Analysis

frequency of 1000Hz with a cutoff frequency of 450Hz. Then, a notch filter was applied with a
cutoff frequency equal to 50 Hz. An high-pass Butterworth filter (sixth order) is also applied with
a cutoff frequency of 10Hz. To finish, a Moving Average with a window of 100 samples is used.

![Filtering process on the EMG signal: before (above) and after (below)](image)

3.3.2 Signal Segmentation

Signal segmentation distinguishes the time intervals that need to be recognized from the overall
acquired signal corresponding to the active movements.

In order to enhance the active segments relative to non-active segments several techniques
are proposed, either based on Root Mean Square (RMS) (see Equation 3.7) or the Teager–Kaiser
energy operator (TKEO) (see Equation 3.6) [54].

Based on a threshold, the signal segmentation includes some steps:

- Application of the Teager-Kaiser energy operator in order to enhance the active segments

  \[ TKEO = x^2(n) - x(n+1)x(n-1) \]  

  (3.6)

- Application of the Root Mean Square (window = 100 samples), intensifying the previous
  step

  \[ x_{rms} = \sqrt{\frac{1}{n} (x_1^2 + x_2^2 + ... + x_n^2)} \]  

  (3.7)

- Calculation of the mean of each channel to be used as a threshold

- Identification of the time intervals in which the signal is higher than the threshold
### 3.3.3 Parameters Estimation

After removing the noise from the EMG data, the muscle activation time, time ratio, mean value and mean value rate were calculated. The parameters are listed in Table 3.2 as well as a brief description of each one.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Muscle activation time (for each limb and muscle)</td>
<td>Average value of time intervals higher than the threshold</td>
</tr>
<tr>
<td>Muscle activation time ratio (for each muscle)</td>
<td>Ratio of Muscle activation time between the two limbs</td>
</tr>
<tr>
<td>Muscle activation mean value (for each limb and muscle)</td>
<td>Average value of electrical potential on time intervals higher than the threshold</td>
</tr>
<tr>
<td>Muscle activation mean value ratio (for each muscle)</td>
<td>Ratio of Muscle activation mean value between the two limbs</td>
</tr>
</tbody>
</table>

### 3.4 Dataset Collection

On a first stage, in order to visually analyze the connection of the devices and the effectiveness of the algorithm in development, some tests were done in random individuals with apparent normal gait. This dataset collection consisted on the individuals walking for some meters.

Afterwards, a small test was performed while recording an individual walking and making some leg movements in order to validate the developed algorithm.

The final dataset was collected comprising 8 patients recovering from the surgery and 9 patients with apparent normal gait. Patients were attending regular orthopedic appointments in "Hospital de Trás-os-Montes e Alto Douro", after the surgery. The control group were healthy individuals without problems or difficulties with walking, who were recruited by the "Colaborar", a network of contacts FhP has developed with elderly people in order to participate in research projects.

These two groups are composed of individuals with substantially the same characteristics, except that they have undergone the surgery or not (brief description in Appendix A).

The preparation before the test includes to clear the skin and attach the several IMU’s and the EMG electrodes (Figure 3.14). Then some instructions about the test execution are given.

Finally, the individual was asked to stand still for some seconds before walking at a comfortable speed in a path shown in figure 3.15. This path has a total distance of 12.5m and includes walking in straight line with some turns, restricted by the available space during the test. At the end point, he (or she) is asked to remain still again for some seconds.
3.4 Dataset Collection

Figure 3.14: Individual prepared for the test

Figure 3.15: Path performed by individuals during the test
3.5 Statistical Evaluation

In order to compare gait parameters between a control group and a group of individuals who underwent a TKR, a statistical analysis was performed.

Mean values and standard deviation were obtained for all measurements. Between group, control and test, differences for age were checked using Student t-tests. Univariate ANOVA was used to test for significant differences in spatiotemporal parameters as well as electromyographic parameters. For parameters involving two measures (for injured and non-injured leg), a ratio between the two was used in the test group and compared with the control group. Student paired t-tests were used to determine differences between the injured and non-injured leg. Statistical analyses were performed using IBM SPSS Statistics for Windows (version 23). A statistical significance level was set at $p \leq 0.05$. 
Chapter 4

Results and Discussion

On this chapter, results concerning the methods described in chapter 3 are presented. These results provide the necessary data to evaluate the algorithm performance on parameters estimation, and also to discuss the possibilities to adapt this system to other devices, for gait analysis purposes. The results of the gait analysis with the chosen dataset are explained and the statistical analysis made with the collected dataset is also presented.

4.1 Dataset description and Test execution

The final dataset, from where the parameters were extracted and the statistical evaluation was performed, comprises 8 patients recovering from the TKR surgery and 9 patients with apparent normal gait. These two groups are composed of individuals with substantially the same characteristics, except that they have undergone the surgery or not.

Table 4.1: Dataset individuals number and age

<table>
<thead>
<tr>
<th></th>
<th>Control group</th>
<th>Test group</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number</td>
<td>9</td>
<td>8</td>
</tr>
<tr>
<td>Age</td>
<td>70.33 ± 1.73</td>
<td>70.50 ± 2.78</td>
</tr>
</tbody>
</table>

These two groups, as seen in table 4.1 have approximately the same number and the same age (Student’s t-test revealed no statistically difference on age variable: p_value = 0.88).

The time elapsed after the surgery until the tests were around 11 ± 2 months for the test group.

For the data acquisition, the subject was asked to stand still for some seconds before walking at a normal desired speed in a determined path. This path has a total distance of 12.5m and includes a walking in straight lines with some turns, restricted by the available space during the test. At the end point, the subjects are asked to remain still again for some seconds. Some deviations to the real value of walked distance can happen due to unexpected turns at the path corners. Moreover, as the individual is asked to walk at a self desired speed, some parameters can not be a good marker to identify gait changes.
4.2 Spatio-temporal data

Figure 4.1 shows a triaxial typical acceleration signal obtained during gait (already pre-processed), positioned on the front of the right thigh.

As the pre-processing includes an axis rotation step, the sensors positioned on the thigh see its axis rotated 90° the same positioned on the shank. This means that every sensor signal will be analysed assuming that they have the same axis. Initially, sensors of the thigh were placed in the front and with the same orientation as the ones of the shank, but this placement was very susceptible to noise triggered by the muscle and skin movements, much more attenuated on the lateral part of the thigh.

Signals have a sampling rate of 100Hz and, when stationary and positioned as described in figure 3.3, the x-axis has a value of 0g, while the y-axis and the z-axis have a value of 1g. As can be observed, signals present repetitive patterns, as is expected, since gait is a repetitive activity.

Linear acceleration is used to segment the signal into steps segments, since this signal has its peaks more pronounced, and, for that reason, it was also used to estimate some gait parameters, as described in Section 3.2.5. An example of the linear acceleration obtained from the raw acceleration signals is shown in figure 4.2.

Linear acceleration was chosen to perform a step detection, identifying the peaks of the signal above a threshold (average of the signal), presenting an accuracy of 97% in determining parameter "steps" (Table 3.1). Errors are caused by very small steps done at the beginning and the end of the walking, as well as in the path turns.

The figure 4.3 shows an example of the steps detection.

Thigh and shank angles were calculated using the accelerometer readings and converting them as presented in section 3.2.3. The figure 4.4 shows these angles varying on time as well as the knee angle obtained from the two last signals.
However, recorded signals do not always present the desired characteristics. Some of them present quite sporadic characteristics, being difficult to visually identify their patterns. In general, the signal can have these kind of characteristics because of a bad positioning of the sensors. Small deviations to the orientation is fixed during the pre-processing, but small differences in the positioning leads to considerable differences between tests. Also the tape used to place the sensors is not always completely glued to the skin, adding some noisy movements and rotations which can affect some acceleration components that are not expected to be recorded.

This problem was verified in the events detection method presented earlier, used to detect events on gait signals. This was the trickiest and most hardest part of the algorithm to accomplish. The toe off detection done by the determination of the minimums of the thigh angle and the
Results and Discussion

Figure 4.4: Thigh, shank and knee angle during gait

heel strike detection done by the determination of the maximums of the shank angle implies the elimination of some wrong peaks, maintaining only the real ones. The problem is that not only this process of elimination is not very reliable, but the whole method is basically a threshold-based method.

For example, on toe off detection, the determination of the minimums implies the determination of the maximums and then compare all the minimums detected between two maximums and chose the last one. This works fine on good signals, but as this system is used in more than one subject, and there is no guarantee that sensors are always placed in the same position, wrong detections happen.

So, perfect signals lead always to good detections of foot events. However, when signals presented different characteristics foot events detection was not correctly performed. This poses some questions, regarding the kind of signals that will be encountered when many different people are going to be tested. As more variability of gait is expected, it is also expected that signals with more variability are obtained, which poses some problems on the correct detection of foot events.

As observed on Figure 4.5 the foot events were correctly detected. However, on Figure 4.6 one heel strike was detected after the real moment.

Parameters estimation

As described in table 3.1 presented in the previous chapter, five parameters are extracted from the linear acceleration of the sensors placed on the shank of both limbs.

The step number, step length and velocity were significantly different for the two groups of the dataset (p=0.008, p=0.01 and p=0.01, respectively), as can be seen in figure 4.7. Statistical results tables are presented in the Appendix.

As expected, step number was higher in the test group (31.38 ± 4.47) than in control group (24.22 ± 4.52), which means that patients need more steps to walk the same distance. Conse-
4.2 Spatio-temporal data

![Figure 4.5: Example of toe off (green lines) and heel strike (red lines) well detected](image)

- Figure 4.5: Example of toe off (green lines) and heel strike (red lines) well detected

![Figure 4.6: Example of toe off (green lines) and heel strike (red lines) not well detected](image)

- Figure 4.6: Example of toe off (green lines) and heel strike (red lines) not well detected

Subsequently, this leads to a lower step length (0.41 ± 0.06m compared to 0.56 ± 0.11m) [55]. Although the step period (0.62 ± 0.09s in control and 0.72 ± 0.12s in test) and the cadence (98.29 ± 13.25steps/min in control and 85.57 ± 14.15steps/min in test) have no statistically significant differences, the speed was lower in the test group (0.61 ± 0.16m/s compared to 0.93 ± 0.27m/s), also as expected. Step number and step period are the variables which are apparently more reliable, since the other ones are dependent on these two and are also dependent of other external factors, such as the real distance walked by the subject.

The maximum knee angle was statistically equal for the control group (52.07 ± 8.98) and the test group (44.00 ± 7.95) with a $p = 0.08$. A difference was expected, since an individual that underwent a knee surgery has tendency to not recover the full range of motion [4]. The minimum
Results and Discussion

Figure 4.7: Step related variables for both groups

Knee angle also was statistically equal for the control group (−3.18 ± 3.61) and the test group (−5.89 ± 6.295) with a $p = 0.34$. When comparing the two limbs, the difference is not statistically significant between the control group (1.04 ± 0.21) and the test group (1.18 ± 0.19).

These variables are demonstrated in figure 4.8.

Figure 4.8: Maximum knee angle during gait and its ratio

The gait cycle time was statistically equal for the control group (1.20s ± 0.10) and the test group (1.38s ± 0.20) with a $p = 0.05$. A difference was expected, since an individual that underwent a knee surgery has tendency to walk more slowly [4]. When comparing the two limbs, the difference is not statistically significant between the control group (1.08 ± 0.20) and the test group (1.09 ± 0.17).

These variables are demonstrated in figure 4.9.

The stance/swing ratio was statistically equal for the control group (55.82% ± 35.02) and the test group (70.42% ± 39.88) with a $p = 0.43$. A difference was expected, since an individual that underwent a knee surgery has tendency to spend more time with the foot supported on the floor [4]. When comparing the two limbs, the difference is not statistically significant between the control group (53.01% ± 31.01) and the test group (63.18% ± 36.90).
4.2 Spatio-temporal data

group (1.48 ± 1.00) and the test group (3.28 ± 1.50). The double support ratio is also statistically equal for both groups, the control group (21.76% ± 19.64) and the test group (31.21% ± 32.41) with a $p = 0.47$.

These variables are demonstrated in figure 4.10.

The angle between the limbs at the heel strike was statistically different for the control group (20.99 ± 2.69) and the test group (13.92 ± 3.49) with a $p = 0.003$. This difference was expected, since an individual that underwent a knee surgery has tendency to loose some range of motion [4]. When comparing the two limbs, the difference is not statistically significant between the control group (1.47 ± 0.82) and the test group (1.21 ± 0.58).

These variables are demonstrated in figure 4.11.

**Limitations**

Detection of parameters from inertial signals could not be properly evaluated as desired, largely due to the small dataset. This weak evaluation was caused mainly by high standard deviations.

To effectively validate foot events and spatio-temporal parameters, the use of traditional methods of gait analysis, such as cameras and force-plates would be required in the future development of this work. Using these methods, the exact time of moments of interest can be determined, and reliable measures of some parameters could be determined.
4.3 EMG data

EMG signals obtained during gait were acquired and stored on the computer for further analysis. Figure 4.12 shows an EMG signal of all positions obtained during gait (already pre-processed), where muscular activations can be identified.

Signals have a sampling rate of 1000Hz. As can be observed, signals present repetitive patterns, as is expected, since gait is a repetitive activity. However, the vertical scale between plots is sometimes very different. This was a problem when recording from the six channels at the same time, without any visible explanation.

Even applying the steps for signal pre-processing and segmentation, it was very difficult to extract relevant information from these signals. Figure 4.13 shows the segmented signals.
4.3 EMG data

Figure 4.13: Example of processed EMG signals during gait (Horizontal blue lines are the thresholds): first line is shank, second line is front thigh and third line is back thigh

The EMG results are presented in graphs that show the muscle activation patterns for all the muscles analysed. These graphs show, for each muscle group, the EMG activations timing as well as its voltage value.

Parameters estimation

As described in table 3.2 presented in the previous chapter, two parameters are extracted from the signals of electrodes placed on the both limbs: back of the shank, back of the thigh and front of the thigh.

The muscle activation time was statistically equal for all positions in the control group and the test group:

- Shank: $0.35 \pm 0.09$ for the control group and $0.33 \pm 0.05$ for the test group ($p = 0.59$)
- Thigh (front): $0.33 \pm 0.05$ for the control group and $0.36 \pm 0.07$ for the test group ($p = 0.32$)
- Thigh (back): $0.49 \pm 0.31$ for the control group and $0.37 \pm 0.09$ for the test group ($p = 0.31$)

However it seems that the control group has a higher value on the back of the thigh, namely the hamstrings muscular group.

These variables are demonstrated in figure 4.14.

The muscle activation was not statistically significant in the thigh positions for the control group and the test group:

- Shank: $0.051 \pm 0.028$ for the control group and $0.025 \pm 0.022$ for the test group ($p = 0.05$)
Results and Discussion

Figure 4.14: Muscle activation and correspondent time

- Thigh (front): $0.084 \pm 0.021$ for the control group and $0.035 \pm 0.039$ for the test group ($p = 0.005$)

- Thigh (back): $0.004 \pm 0.002$ for the control group and $0.022 \pm 0.004$ for the test group ($p = 0.001$)

Limitations

Detection of parameters from EMG signals could not be properly done as desired, not only due to the small dataset.

EMG recording in moving subjects proved to be difficult as the system is drastically affected by the movement. The small connectors, even when isolated, are a constant source of noise and artifacts. In some cases, a random channel suddenly disconnected while recording, becoming a real problem when performing the tests.

There are some considerations to take into account, regarding the operability of a future system. For example, the position of the electrodes can limit the reliability of tests as small positional deviations can lead to considerable differences in the recordings. Therefore, exploration of electrode placement would be needed to ensure more reliable results in future work.

To effectively validate EMG parameters, the use of traditional methods of gait analysis would be required in a future development of this work.
Chapter 5

Conclusions and Future Work

Human gait can be affected by several problems and diseases such as osteoarthritis. The process of ageing is responsible for the aggravation of this problem which is requiring more and more attention. The increasing prevalence of TKR highlights the need to appropriately assess post-operative outcome of this procedure since many individuals still experience an antalgic gait pattern. Therefore, there is a need to obtain a more complete analysis that overcomes the limitations of the current analysis methodologies, providing information about the recovery process after the surgery.

In this project, a study was done regarding the use of IMU and EMG monitoring system as a gait analysis tool, aiming to improve the current gait analysis techniques.

Based on the current research, inertial sensors were used to quantify some gait parameters already measured in previous studies that were related with knee functionality. Also, other important parameters were determined with an EMG monitoring system in order to access muscular activity.

The objective of this project was substantially achieved since the results from signals analysis suggest that they can be used to evaluate gait changes, based on significant differences in several gait parameters, as described in Sections 4.2 and 4.3.

The most challenging parts of the work was the detection of gait events such as foot contacts. The integration of IMU and EMG information was not possible because of the reliability of the EMG monitoring system and time constrictions.

The use of body fixed sensors for gait analysis purposes may be therefore a major strategy to evaluate gait changes in patients recovering from a TKR. Frequent assessment of gait changes can be done over time, providing an evolution description to health professionals who can modify the ineffective recovery strategies. Therefore, further development of the project would be of great value.

5.1 Future work

Considering the results obtained in this project, some future work was identified to potentially improve this gait analysis technique:
Conclusions and Future Work

- Bluetooth connection of the used IMU’s is quite sensitive. There is a need to improve the connection stability and to adapt the algorithm to different IMU’s with better specifications.

- Signal-to-Noise ratio in both signals are quite low, specially in EMG signals, caused by artifacts of the movement. A new approach should be tested to access muscular activity since BITalino is not a reliable system in these dynamic conditions.

- Specific position of IMU’s leads to variations between tests. Recording should be independent of sensors position and orientation.

- Validation of foot contacts detection and gait parameters estimation using the traditional methods (i.e. cameras and force-plates) would be important, instead of using a recording video to the effect.

- Development of an Android application capable of recording and processing both inertial and EMG data would be an added value asset, dismissing the use of a laptop and becoming more ergonomic.

- Robustness of algorithm should be improved since in some cases the gait parameters are not well determined due to differences in gait which compromises some events detection.

- Evaluation of gait changes should be performed in the same subject over time in order to reduce to the maximum the enveloping variables.

- Synchronization of both inertial and electromyographic from the IMU’s and EMG signals.
Bibliography


Appendix A

Statistical tables

### Teste-T

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Figure A.1: Student’s T-test for dataset samples
### Figure A.2: ANOVA for step related parameters

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### Figure A.3: ANOVA for angle related parameters

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