INSTITUTO POLITÉCNICO DE BRAGANÇA
Escola Superior de Tecnologia e Gestão

STATISTICAL GAIT ANALYSIS IN PATIENTS AFTER TOTAL HIP ARTHROPLASTY

Susana Moreira Carneiro

Final Report of the Project / traineeship submitted to
Escola Superior de Tecnologia e Gestão
Instituto Politécnico de Bragança

To the fulfilment of the Master of Science degree in
Biomedical Technology

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I dedicate this work to my family.
I can honestly say that without your support,
I wouldn’t have gotten this far.

Abstract

Patient’s functional recovery after Total Hip Arthroplasty (THA) is often slow. Besides, patients tend to adjust gait patterns to avoid the pain, a condition referred to as antalgic gait. The aim of this work is to highlight changes in gait and muscle activation patterns of patients after total hip arthroplasty, by means of a statistical gait analysis.

The gait analysis was performed on 20 patients with unilateral hip prosthesis (3, 6 and 12 months post-operatively) and 20 controls, at self-selected and fast speed. The analysis was performed using the system Step 32 (DemItalia, Italy). Various statistical analyses were done to compare the outcomes of the two groups. Subjects were examined bilaterally by means of basographic sensors (foot switches), goniometric sensors (in the knee and hip), and surface electromyography of five leg muscles.

This study demonstrated that, for patients, the number of atypical strides is higher and the heel contact phase is extended in time, in both sides. Besides, on the operated leg, despite a significant increase in the hip dynamic range of motion, patients do not reach normal range of motion (ROM) values even one year after the intervention. Furthermore, the electromyographic results show that the number of simpler activations tends to increases and the number of complex activations decreases over the time for THA patients, suggesting a compensations strategy.

Keywords: Total hip arthroplasty, statistical gait analysis, basographic sensors, goniometric sensors, electromyography.

Resumo

A recuperação funcional do paciente após a artroplastia total da anca (PTA) é muitas vezes demorada. Além disso, os pacientes tendem a ajustar padrões de marcha de forma a evitar a dor, uma condição referida como marcha antálgica. O objetivo deste trabalho é destacar as alterações na marcha e padrões de ativação muscular dos pacientes após artroplastia total da anca, por meio de uma análise estatística da marcha.

A análise da marcha foi realizada em 20 pacientes com prótese unilateral da anca (3, 6 e 12 meses pós-operatório) e 20 controles, com velocidade auto-selecionada e rápida. A análise foi realizada através do sistema Step 32 (DemItalia, Itália). Várias análises estatísticas foram realizadas para comparar os resultados dos dois grupos. Os indivíduos foram examinados bilateralmente através de sensores basográficos (interruptores de pé), sensores goniométricos (no joelho e anca) e electromiografia de superfície em cinco músculos da perna.

Este estudo demonstrou que, para os pacientes, o número de passos atípicos é maior e a fase de contacto do calcanhar (H) é prolongada, em ambos os lados. Além disso, na perna operada, apesar de ocorrer um aumento significativo na amplitude dinâmica do movimento da anca, os pacientes não atingem valores de amplitude de movimento (ADM) normais, mesmo um ano após a intervenção. Além disso, os resultados electromiográficos mostram que o número de ativações mais simples tendem a aumentar e do número de ativações complexos a diminuir ao longo do tempo para os pacientes submetidos a artroplastia total da anca, sugerindo uma estratégia de compensação.

Palavras-chave: Artroplastia total da anca, análise estatística da marcha, sensores basográficos, sensores goniométricos, electromiografia.

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To my supervisor, Professor Marco Knaflitz, a great and recognized professional in the statistical gait analysis area. Thank you for accept me and guide me in all these months. I deeply acknowledge your support.

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Thanks to Mrs. Rosanna Evangelisti and Laura Rivella, respectively coordinator and former director of Struttura complessa Recupero e Rieducazione Funzionale, ASLTO4 Piemonte, Torino, Italy.

I would like to thank all patients who agreed to participate in a survey and wish them the best recovery. I sincerely hope that works like this can help them to promote their welfare.

To my father, for always pushing me to be the best I can be, both intellectually and emotionally. I hope someday to be able to repay twice in all that you did for me.

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Chapter 1. Introduction

1.1. Background

Osteoarthritis (OA) also known as degenerative arthritis or degenerative joint disease or osteoarthrosis is a progressive musculoskeletal disorder characterized by gradual loss of articular cartilage. It is the most common cause of long-term disability in most populations of people over 65 due to the aging process that reduces the ability of the cartilaginous tissue to withstand loads and stresses. The lower limbs should be strong enough to allow the process of locomotion, body support and posture. Because of this, when knees or hips are affected, it becomes one of the most debilitating ones, considerably reducing the patient's physical and psychosocial functions.

The surgery performed to relieve pain and restore range of motion by realigning or reconstructing a dysfunctional joint is called arthroplasty. Total hip arthroplasty (THA) is a surgical procedure performed in patients with osteoarthritis of the hip. Thanks to the continuing development of joint arthroplasties, physical therapy and psychosocial support, it is now possible to restore a near normal quality of life to patients. However, after surgery, many individuals still experience an antalgic gait pattern, or adapted walking pattern to avoid pain, during the post-operative recovery period (Illyés A, 2005; Beaulieu M, 2010). According to Loizeau *et al.* (Loizeau J, 1995), in subjects with locomotor disorders, either orthopaedic or neurological, asymmetries in the gait pattern are expected. This particular gait pattern is often non-ideal for fracture devices and can greatly reduce device lifespan and patient quality of life.

The importance of evaluation of arthroplasty outcome has long been recognized. Post-operative evaluations of THA are recognized as an important means for judging patient recovery. A large variety of scores and evaluation systems have been used to assess the outcome of hip and knee arthroplasties. Nevertheless, due to clinician subjectivity and the lack of a universal standard, quantifying surgical results and subsequent recovery progress can be difficult. Quantitative gait analysis is generally accepted as an objective measurement of surgical success. The clinical use of gait analysis systems is effective in determining functional outcomes of lower limb corrective surgeries by their abilities to quantify the spatio-temporal parameters of walking and provide an overall assessment of physical capability in recovering patients.

Therefore, the use of instruments that have a better sensitivity and specificity than traditional scoring systems is needed to evaluate the results of arthroplasty and enhance the surgeon’s ability to assess the overall outcome, allowing a more directed treatment.

1.2. Motivations and objectives

Statistical gait analysis is not a common concept in Portugal. The gait analysis techniques involve analysis systems with algorithms that can automatically detect the main gait events. Here, equipment and skilled personnel to acquire, process and interpret the biomechanics gait information are very limited and underdeveloped. The main motivation of this study is to introduce new concepts and new techniques that may somehow contribute to the development of gait analysis.

A total hip arthroplasty is the procedure used to treat patients suffering from osteoarthritis of the hip. But what are the consequences of THA in patients? Do operated patients walk again normally and with normal muscle activation patterns? If so, how long it takes to acquire a march within the range considered normal? What is the effect on the non-operated limb? With this particular study, we intend to realize what happens when patients are subjected to a THA.

Thus, in this thesis, it is suppose to obtain, by means of statistical gait analysis, the evaluation of the outcome of patients that underwent total hip arthroplasty. Gait signals are already available for patients at 3, 6 and 12 months after the intervention and for an age-matched control group.

1.3. State of art

A lot of studies have been done over the time. In this state of the art some of the research topics related to gait analysis are outlined, describing different methodologies and the main findings, in chronological order.

In 1989, Kadaba *et al.* (Kadaba M, 1989) present a marker system that can be easily implemented for routine clinical gait evaluations. Motion analysis was performed using a computer-aided video motion analysis system with five infrared cameras under the control of a computer. Foot contact patterns were recorded using pressure-sensitive foot switches attached to the heel, first and fifth metatarsals, and great toe of each foot. Gait parameters were calculated for each run using foot switch data. A five point window (Hanning) was used for smoothing raw three-dimensional marker trajectories before computing the joint angle motion. Data presented in this paper should be a useful reference for describing and comparing pathologic gait patterns.

In 1995, Schroeder *et al.* (Schroeder H, 1995) performed a study to assess gait parameters and patterns of patients with stroke, and the temporal changes of these parameters. A foot-switch gait analyzer was used to test 49 ambulatory patients with stroke and 24 controls. Gait was analyzed using a portable stride analyzer. The device consisted of insoles which contained four foot switches in the heel, first and fifth metatarsal, and great toe regions. The data were transferred to a personal computer for analysis of comprehensive and unilateral gait parameters. General gait parameters improved over time, with the largest changes occurring in the first 12 months. However, parameters which describe the asymmetrical pattern of gait did not change over time.

In the same year, Loizeau *et al.* (Loizeau J, 1995) carried out a study to determine whether the muscle powers and the mechanical energies developed during the push-off period of the gait cycle of patients having a total hip prosthesis were different from able-bodied subjects as well as the effect on the non-operated limb. The gait analysis was performed with reflective markers placed to identify the three-dimensional kinematics of the lower limbs and with the Expert Vision software of Motion Analysis system and a four-segment (pelvis, thigh, leg, and foot) chain link model was elaborated in the KINTRAK software from Motion Analysis Corporation. Gait analyses showed that not only the hips of the surgical group were affected but also the knees. The operated and the non-operated hip developed less energy than the able-bodied group. These results confirmed the presence of some mechanical dysfunction in the non-operated limb.

In 1999, Benedetti *et al.* (Benedetti M, 1999) report a single case study on a patient that underwent THR trying to show that quantitative gait analysis is essential to augment the understanding of the mechanisms underlying gait. Lower limb functional evaluation during gait was performed using the ELITE stereophotogrammetric system for the acquisition of kinematic variables. A Kistler platform was used to study foot-ground reaction forces, which were utilized to estimate joint moments. Kinematic data relative to the lower limb and to foot-ground reaction forces during stance were acquired and digitized with a sampling rate of 100 Hz, with the synchronization managed directly by the ELITE System. Reflective markers, were strapped to the pelvis, thigh, shank, and foot on the patient right and left sides. This study enabled clinicians to adapt the rehabilitation program to the specific patient.

In 2004, Vogt *et al.* (Vogt L, 2004) examined the hip abductor activation pattern of 14 hip replacement patients and 10 age-matched healthy controls by measuring surface electromyography (EMG) onset and cessation times. Stride characteristics, surface EMG from bilateral gluteus medius, and 3D pelvis kinematics were evaluated. EMG onset times were normalized with regard to the individual stride time for each gait cycle. Bipolar surface electrodes were used to sample EMG activity during treadmill ambulation. The EMG activity was recorded by a multichannel EMG datalogger system (BIOVISION) operating at 1000Hz. The different phases of the gait cycle were registered by four pressure-sensitive footswitches. The results indicated deficiencies in the hip abductor recruitment pattern of hip arthroplasty patients.

At the same time, Duhamel *et al.* (Duhamel A, 2004) performed a gait analysis study to design statistical tools for solving the principal problems encountered in the clinical practice of gait analysis. Kinematic gait parameters were recorded using a Vicon video system for motion analysis, using five infrared cameras. Thirteen spherical, retro reflective markers were used to define different segments of the pelvis and lower limbs. The three-dimensional trajectories in the frontal, sagittal and axial planes were recorded by the cameras placed in defined positions in a room. The pelvic tilt, pelvis obliquity, pelvic rotation, hip flexion/extension, hip abduction/adduction, hip rotation, knee flexion/extension, knee varus/valgus, foot alignment, foot rotation and ankle flexion/extension were analysed for each subject. Gait disturbances are mainly characterised by measurements of kinematic parameters, revealing a decrease in velocity, a stride length with a higher cadence rate than in control subjects and an increase in the time spent with double limb support to obtain better balance.

Madsen *et al.* (Madsen M, 2004) examined the effect of the surgical approach used in total hip arthroplasty (THA) on gait mechanics six months following surgery. Discriminant function analysis was to determine the distinction of the groups with respect to sagittal plane hip range of motion, index of symmetry, trunk inclination, pelvic drop, hip abduction, and foot progression angles. A six-camera motion analysis system was used to measure the positional data of the markers. **A** strain gauge force plate sampling at 1200 Hz was used to measure ground reaction forces. Ten successful trials, five plate contacts with the right leg and five plate contacts with the left leg, were collected. Data were analyzed using the Vicon Bodybuilder software and MATLAB programs. These results support the conclusion that six months following surgery, the gait of the majority of THA patients has not returned to normal.

Filially, also in 2004, Cho *et al.* (Cho S, 2004) evaluate the abnormal gait patterns and gait improvements after a total hip arthroplasty (THA) in patients with hip dysplasia and osteonecrosis of the femoral head (ONFH). The parameters measured were those for the temporal gait measurements, kinematics and kinetics. Gait analysis was performed using a three-dimensional computerized Vicon 370 motion analysis system. Results show that, there were less postoperative gait improvements in the patients with severe hip dysplasia than in those with ONFH who had a relatively normal anatomy.

In 2005, Bennett *et al.* (Bennett D, 2006) used a prospective blinded design to analyse early post-operative walking ability using gait analysis to compare gait kinematics in patients receiving minimally invasive and traditional hip replacement surgery. The three-dimensional gait analysis was carried out using a Vicon camera system and lower-body markers set. Data were processed using Vicon Plug-In-Gait. Contrary to previous studies, there was no improvement in early post-operative gait for those patients who received THR using the minimally invasive technique.

Also in 2005, Illyés *et al.* (Illyés A, 2005) performed a study to determine how selected gait parameters may change as a result of coxarthritis. Gait analysis was performed using an ultrasound-based Zebris system with a 19-point biomechanical model. The measuring head with three sensors was positioned behind the individual and the five ultrasound triplets with three active markers on each were placed on the sacrum, left and right thighs, and left and right calves. From the spatial coordinates of the investigated anthropometrical points, the kinematical data (step length, step width, knee, hip and pelvic angles) was calculated. The results indicate a generally poor functional outcome, even though asymmetrical loading was observed. Major limitations in physical function were detected.

In 2006, Illyés et al, (Illyés A, 2006) studied about how selected gait parameters may change as a result of total hip arthroplasty at a constant gait speed. The gait of 20 patients with unilateral hip disease, who underwent total hip arthroplasty (THA), was analyzed. The spatial-temporal and angular parameters were analysed. Spatial coordinates for the determination of kinematic data were collected using an ultrasound-based Zebris three-dimensional motion analysis system. The measuring head with three sensors was positioned behind the individual and the five ultrasound triplets with three active markers on each were placed on the sacrum, the left and right thighs, and the left and right calves. The data, obtained from the measuring system recording the active markers. The spatial coordinates were recorded at a frequency of 100 Hz. Simultaneously, the ground forces were measured at 1000 Hz. This study suggested that the THA could reverse the adverse influence on other joints prior to the symmetrical normalization of hip motion.

In 2007, Nankaku *et al.* (Nankaku M, 2007) examined the effects of lateral displacement on walking efficiency after THA. Gait analysis was performed using a three-dimensional motion analyzer composed of four charge-coupled device cameras and two floor reaction force platforms. The sampling frequency was 240 Hz for the floor reaction platform data and 60 Hz for the three-dimensional data. Reflective markers were attached to 11 points on the body surface of each subject. The results suggest that trunk compensation strategy for hip abductor weakness in patients soon after THA can lead to increased energy expenditure.

Foucher *et al.* (Foucher K, 2007) evaluated whether preoperative gait adaptations persist one year after Total Hip Replacement (THR) in the same set of subjects. Hip kinematics and kinetics were measured for 28 subjects before and one year after THR and compared to those of 25 subjects with radiographically normal hips. Motion of six passive retroreflective markers, placed at the iliac crest, greater trochanter, lateral knee joint line, lateral malleolus, lateral aspect of calcaneus, and the head of the fifth metatarsal, were tracked by four optoelectronic cameras. Ground reaction force data were collected with a multicomponent forceplate. The three-dimensional locations of each joint centre are known throughout gait, based on the measured marker trajectories and anthropometric measurements. Despite good to excellent clinical functional outcome, gait in THR patients does not return to normal by one year after surgery.

Also in 2007, Mont *et al.* (Mont M, 2007) evaluated temporal-spatial parameters, hip kinematics, and kinetics in hip resurfacing patients compared with patients with unilateral osteoarthritic hips and unilateral standard total hip arthroplasties. The gait analysis laboratory used 8 strategically located Falcon cameras and 2 centrally located force plates. Twenty-six reflective markers were placed on subjects, and the information obtained from them was later used to create a musculoskeletal model. The data were then used in further processing with OrthoTrak software (Motion Analysis Corporation). This study showed more normal hip kinematics and functionality in

resurfacing hip arthroplasty, which may be due to the large femoral head.

In 2009, Nantel *et al.* (Nantel J, 2009) made an observational study comparing gait patterns in patients with total hip arthroplasty (THA) and surface hip arthroplasty. The main outcomes measures were gait patterns (cadence, duration of single and double support phases, stride length, and walking speed), hip abductor muscle strength, clinical outcomes, and radiographic analyses were compared between groups. Nineteen 14-mm diameter reflective markers were used to define lower-limb body segments. The kinematic and kinetic data were collected at 60 Hz by using 8 optoelectronic cameras and at 120 Hz with 2 embedded force platforms, respectively. The abductor muscles' strength on both sides was assessed by using a handheld myometer. Kinematic and kinetic parameters were derived by using VICON Clinical Manager. This work allows to conclude that the surface hip arthroplasty characteristics could allow the return to a more normative gait pattern compared with THA.

In 2010, Lugade *et al.* (Lugade V, 2010) investigated pre and postsurgical changes in gait symmetry in patients receiving either an anterior or anterolateral hip replacement. Three-dimensional kinematic and kinetic gait analyses were performed on the patients while walking. Three-dimensional marker trajectories of 29 markers placed on bony landmarks were captured at 60 Hz using an 8-camera motion analysis system. Motion data were filtered using a fourth order low-pass Butterworth filter, with a cut-off frequency of 8 Hz. Ground reaction forces (GRF) of both feet were collected at 960 Hz with two force plates placed in series along the walkway. Findings of this study highlight the potential impact of surgical approaches on short-term changes in gait asymmetry.

Beaulieu *et al.* (Beaulieu M, 2010) studied the effect of THA on the pelvis, hip, knee and ankle joint kinematics, as well as the hip, knee and ankle kinetics of both the operated and non-operated limbs during walking. A nine-camera digital optical motion capture system was used to capture 45 spherical retro-reflective markers placed on various landmarks of the participants. Furthermore, a force platform was used to record, at 1000 Hz, ground reaction forces during the stance phase of the gait cycle. The raw three-dimensional marker trajectories were filtered using a Woltring filter, whereas a low pass Butterworth filter (cut-off frequency of 6 Hz) was applied to the ground reaction forces. From the filtered 3D marker trajectories, a kinematic model was previously described. THA patients displayed kinematic adaptations at the ankle joint of the operated limb and non-operated hip joint that may be leaving them at risk of developing other joint diseases.

Tanaka *et al.* (Tanaka R, 2010) investigated the factors influencing gait improvement in the patients who had undergone total hip arthroplasty (THA). All the patients were analyzed during free walking along a 5-meter walkway equipped with a ground-reaction force plate (Gait Scan 8000). Basic parameters such as the velocity, cadence, stride length, step length, and single-support and double support duration were directly displayed on the Gait Scan 8000 system. Statistical analysis was performed using SPSS version 12.0. Analysis of variance was performed to assess the mean values and standard deviations for the above parameters. The mean values of the spatiotemporal parameters of the patients showed considerable improvement by 12 months after surgery; however, they did not reach the same values as those observed in the healthy subjects.

Still in 2010, Agostini *et al.* (Agostini V, 2010) carried out a study with the objective to present a normative dataset of muscle activation patterns obtained from a large number of strides in a population of 100 healthy children aged 6–11 years. Signals were acquired by means of a multichannel recording system for statistical gait analysis (Step 32, DemItalia, Italy). Each subject was instrumented with foot-switches, knee goniometers, and SEMG probes. Three foot-switches were attached beneath the heel, the first and the fifth metatarsal heads of each foot. A goniometer was attached to the lateral side of each lower limb for measuring the knee joint angles in the sagittal plane. Surface EMG probes were attached over some leg muscles, bilaterally. EMG signals were further amplified and low-pass filtered by the recording system (450 Hz, 6 poles). The analysis allowed to obtain the most recurrent patterns of activation during gait, demonstrating that a subject uses a specific muscle with different activation modalities even in the same walk.

1.4. Thesis outline

Chapter 2 reviews the anatomy of the hip and the anatomical position of the principal muscles that allows the human locomotion. Reviews also one of the most common and painful problems on the hip (osteoarthritis of the hip joint) and the most common orthopaedic procedure performed in patients with this pathology. Finally, some different methodologies used to assess THA outcome. It discusses the requirements of a scoring questionnaire that must be valid, reliable and responsive; and explains the problems with the questionnaires that are subjective, restricted to a specific pathology, and their low sensitivity to change.

Chapter 3 introduces the gait analysis and explains the importance of this method of analysis in the study of human gait. Statistical gait analysis and its instrumentation (basographic, goniometric and electromyographic systems) are also commented as well as the kind of information we can obtain through this methods. Finally, are referred some applications of gait analysis to hip prosthesis.

Chapter 4 describes the materials and methods used, including the selection criteria applied to choose the patients and the controls and the experimental protocol, as well as the characteristics of the system Step 32, the procedure made to obtain the results and the signal processing inherent in this system.

The basographic, goniometric and electromyographic results are presented and commented in the chapter 5, with enlightening graphs. These graphs show the evolution over the year, in patient’s case and the results from the control group in all the trials.

Chapter 6 reports the discussion of results obtained in this study.

Finally, chapter 7 summarizes the contribution of this thesis and outlines some perspectives of the proposed methods.

Chapter 2. Total Hip Arthroplasty

2.1. The hip joint

Locomotion is a very complex task, for which contribute the coordinated efforts of sensorial, muscle and skeletal systems. It results form a complicated process involving the brain, spinal cord, peripheral nerves, muscles, bones and joints which makes its assessment a very difficult task (Whittle M, 2007). Locomotion or alternated bipedal walking is a basic, key essential function as it allows humans to perform several other tasks. The process of locomotion, body support and posture is executed by the lower limbs.

The human leg is composed of a basal segment, the femur (thighbone), an intermediate segment, the tibia (shinbone) and the smaller fibula; and a distal segment, the foot, consisting of tarsals, metatarsals, and phalanges (toes).

Hip is the portion of the body joining the lower extremity to the trunk. It is designed for strength as well as mobility. Hence, it is where the bones are heavier, stronger, with their processes more marked and with muscles bigger and more powerful. It is often the place of injury and disease, the bones being fractured, the joint luxated, and sometimes affected by bone tuberculosis and other diseases.

The hip joint, or coxofemoral joint, is the articulation of the acetabulum of the pelvis and the head of femur (Figure 1). The hip joint is called a ball-and-socket joint because the spherical head of the femur rotates inside the cup-shaped hollow socket (acetabulum) of the pelvis. The head of the femur is closely fitted to the acetabulum for an area extending over nearly half a sphere, and at the margin of the bony cup it is still more closely embraced by the glenoidal labrum, so that the head of the femur is held in its place by that ligament even when the fibres of the capsule have been divided (Gray H, 1918). The normal hip joint is well designed to withstand the forces that act through and around it, assisted by the trabecular systems, cartilaginous coverings, muscles, and ligaments (Levangie P, 2005). To give the necessary support and security, the band-like ligaments joining the bones are strong and the extent of the movements is restricted.

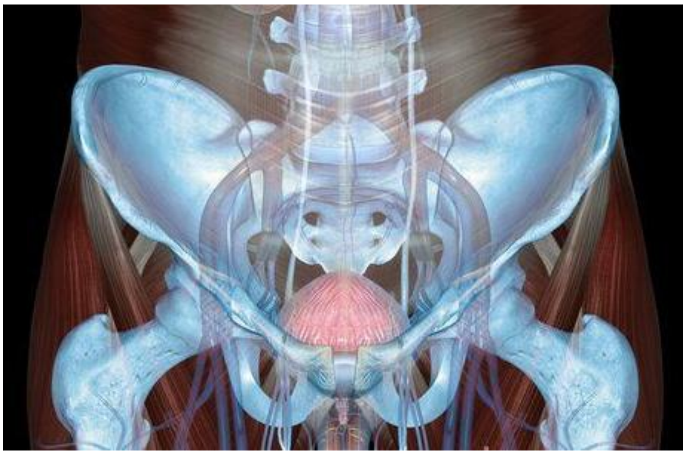

Figure 1. The hip joint, formed by the head of the femur and the acetabulum.

The length of the neck of the femur and its inclinations to the body of the bone has the effect of converting the angular movements of flexion, extension, adduction, and abduction partially into rotation movements in the joint (Gray H, 1918), Table 1. Thus, when the thigh is flexed or extended, the head of the femur, on account of the medial inclination of the neck, rotates within the acetabulum with only a slight amount of movement. The forward slope of the neck similarly affects the movements of adduction and abduction. Conversely rotation of the thigh which is permitted by the upward inclination of the neck, is not a simple rotation of the head of the femur in the acetabulum, but is accompanied by a certain amount of gliding (Gray H, 1918).

Table 1. The next table describes the major muscles surrounding the hip, categorized by function.

Function	Muscles
Flexion	Rectus femoris, iliopsoas, tensor fasciae latae and adductor longus
Extension	Gluteus maximus, hamstrings and adductor magnus
Abduction	Tensor fasciae latae, gluteus medius and gluteus minimus
Adduction	Adductors (magnus, longus, brevis) and gracilis
Internal Rotation	Piriformis, tensor fasciae latae, gluteus medius and minimus
External Rotation	Piriformis, gluteus maximus, medius and minimus

The hip joint is completely surrounded by muscles. Most of the muscles which move the hip joint originate on the pelvis. The important exception is the psoas muscle which originates from the front of the lumbar vertebrae. Iliacus originates on the inside of the pelvis. The two tendons combine to form the iliopsoas, inserted at the lesser trochanter of the femur; the main action of these two muscles is to flex the hip (Whittle M, 2007). Iliopsoas is opposed by gluteus maximus, the strongest extensor of the hip. Gluteus medius and gluteus minimus originate from the side of the pelvis and are inserted into the greater trochanter of the femur; they primarily abduct the hip (Whittle M, 2007).

The leg muscles allow us to stand, walk, run and jump. These muscles work individually, and in cooperation with the other muscles, to provide movement of the legs and stability of the upper body. In general, the leg muscles can be divided into two groups: the upper leg muscles and the lower leg muscles, that can be further divided into anterior (front) and posterior (back) muscles.

The primary front leg muscles, or thigh muscles, are the four muscles of the quadriceps femoris: vastus intermedius, vastus medialis, vastus lateralis, and rectus femoris. Rectus femoris originates from around the anterior inferior iliac spine of the pelvis and inserts into the quadriceps tendon; it flexes the hip, as well as being part of the quadriceps which extend the knee (Whittle M, 2007). The muscles at the back of the upper leg are often called the hamstrings and include the biceps femoris, semitendinosus and semimembranosus. Figure 2 shows some superficial muscles of the leg.

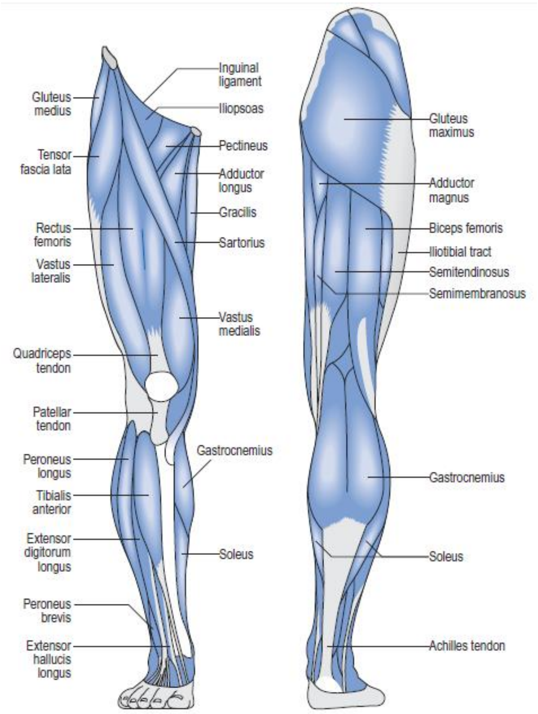

Figure 2. Superficial muscles of the leg (Whittle M, 2007).

In the lower leg, we find the shin muscles that are responsible for dorsiflexion of the foot, or bending the foot upwards at the ankle: tibialis anterior, extensor digitorum longus, extensor hallucus longus and peroneus tertius muscles. The outside lower leg contains the peroneus longus and peroneus brevis muscles that are responsible for sideways flexion and extension of the foot at the ankle and also to provide lateral stability to the foot. The back of the lower leg includes the calf muscles which are the gastrocnemius, soleus, and plantaris muscles. The calf muscles pull up the heel and extend the foot, during the "push-off" phase of walking and running.

2.1.1. Hip joint pathology

The very large active and passive forces crossing the hip joint makes the weakened components of the joint structures susceptible to wear and to failure. Small changes in the biomechanics of the femur or the acetabulum can result in increases in passive forces above normal levels or in weakness of the dynamic joint stabilizers. One of the most common and painful problems on the hip is related with the deterioration of the articular cartilage and to subsequent related changes in articular tissues, known as osteoarthritis.

2.1.1.1. Osteoarthritis of the hip joint

The term “arthritis” literally means inflammation of a joint, but is generally used to describe any condition in which there is damage to the cartilage. Osteoarthritis is the most common form of arthritis and is associated with degeneration of the joint cartilage and with changes in the bones underlying the joint. The cartilage becomes brittle and splits. Some pieces may break away and float around inside the synovial fluid within the joint that can lead to inflammation. Usually the pain early on is due to inflammation. In the later stages, when the cartilage is worn away, most of the pain comes from the mechanical friction of raw bones rubbing on each other. In Figure 3 is shown the aspect of a healthy bone comparing to a bone with osteoarthritis.

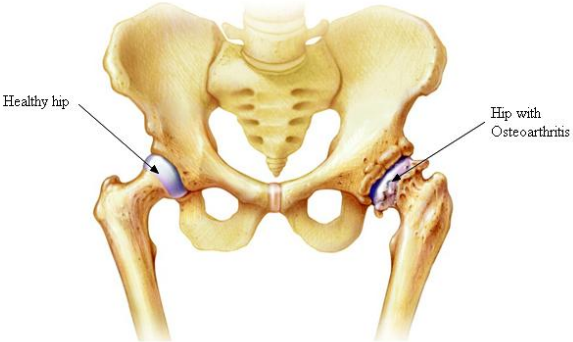

Figure 3. The hip joint normal (left) and deterioration of cartilage (right).

According to Levangie (Levangie P, 2005), many factors can increase the risk of developing osteoarthritis, such as obesity, muscle weakness, heredity, previous injury to the joint, childhood disorders, repeated overuse of the joint and aging. Once the disease is detected, must be treated immediately, otherwise may lead to other problems, e.g. “Limitation of hip extension as a consequence of osteoarthritis may lead to excessive lumbar spine movement to achieve adequate movement of the lower extremity during gait.”

Some treatment options may include weight loss, exercise and physical therapy, glucosamine and chondroitin supplements, and anti-inflammatory medications. However, if non-surgical treatment is unsuccessful, hip surgery is the best treatment option to help regain quality of life.

2.2. Total Hip Arthroplasty

Total Hip Arthroplasty (THA) is a common orthopaedic procedure performed in patients with hip problems. The most common condition for which THA is done is severe osteoarthritis of the hip, accounting for 70% of cases (Siopack J, 1995), which causes severe pain and the limitation in activities of daily life.

The first THA is thought to have been done in London by Phillip Wiles in 1938 (Siopack J, 1995). The procedure was further developed in the 1950s by pioneers such as McKee and Farrar. Later, in the late 1960s, Sir John Charnley approached the problem of artificial hip joint design by using the biomechanical principles of human hip joint function based on this previous work.

THA involves the surgical excision of the head and proximal neck of the femur and removal of the acetabular cartilage and subchondral bone. An artificial canal is created in the proximal medullary region of the femur, and a metal femoral prosthesis, composed of a stem and small-diameter head, is inserted into the femoral medullary canal (Siopack J, 1995). An acetabular component composed of a high-molecular weight polyethylene articulating surface is inserted proximally into the enlarged acetabular space (Siopack J, 1995). Figure 4 shows the aspect of the hip before and after a THA.

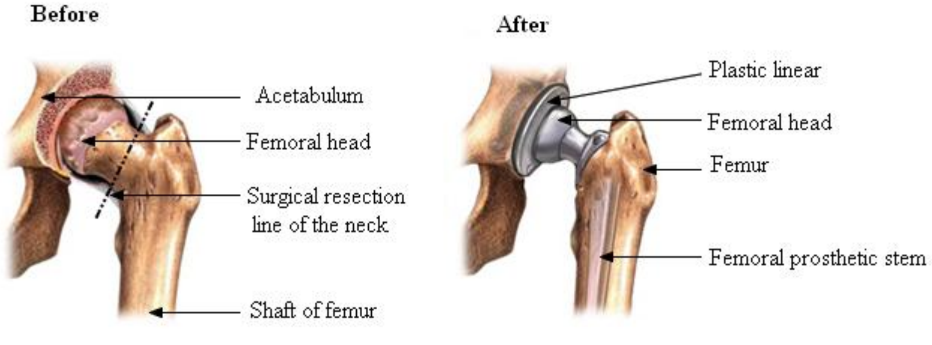

Figure 4. Anatomy of the hip before and after surgical THA.

Using metal alloys, high-grade plastics and polymeric materials, is possible to replace a painful, dysfunctional joint with a highly functional, long-lasting prosthesis. Over the past half-century, there have been many advances in the design of medical devices, construction, and implantation of artificial hip joints, resulting in a high percentage of successful long-term outcomes. To yield successful results, these THA components must be fixed firmly to the bone, either with polymethylmethacrylate cement or, in more recent uncemented designs, by bony ingrowths into a porous coating on the implant, resulting in "biologic" fixation (Siopack J, 1995). Hybrid prosthesis also exists, where only a femoral component is cemented.

A THA implant has three parts: the stem, which fits into the femur; the ball, which replaces the spherical head of the femur; and the cup or shell, which replaces the worn out hip socket. Each part comes in various sizes to accommodate different body sizes and types. In some designs, the stem and ball are one piece; other designs are modular, allowing for additional customization in fit. Figure 5 shows an example of prosthesis composed by stem, ball and socket shell.

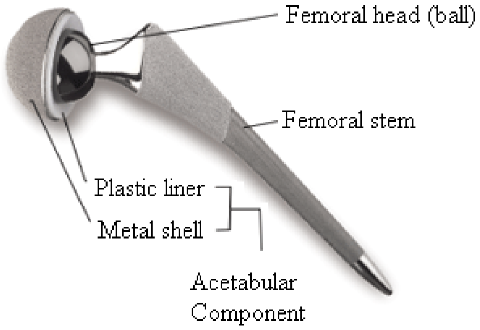

Figure 5. Three components comprise THA: stem, ball and socket shell.

There are some other conditions for which the procedure may be indicated and which predispose to the development of secondary osteoarthritis. This includes the developmental dysplasia of the hip, Paget's disease, trauma, fractures of the femoral neck and osteonecrosis of the femoral head. Patients with rheumatoid arthritis, other collagen diseases such as systemic lupus erythematosus, and ankylosing spondylitis may benefit as well (Siopack J, 1995).

Usually, patients presenting necrosis of the femoral head are aged 35-50 years; patients with arthritis are usually elderly (60–85 years) and patients with a femoral neck fracture who are elder than 70 years can benefit from a THA (Pfeil J, 2010).

To justify total hip replacement, pain must be refractory to conservative measures such as oral nonsteroidal anti-inflammatory medication, weight reduction, activity restriction, and the use of auxiliary supports such as a cane. It is generally preferred that THA is performed in patients older than 60 years because at this age, the physical demands on the prosthesis tend to be fewer and the longevity of the operation approaches the life expectancy of the patient (Siopack J, 1995).

The large number of operations performed each year reflects the fact that more than 90% of appropriately selected patients achieve complete pain relief and notable improvement in function (Siopack J, 1995). It is a well-established treatment and its benefits for physical functioning are sustained in the long term (Cushnaghan J, 2007). However, despite the success of the operation and according to some studies, patients after THA surgery can present some difficulties regaining a normal pattern of walking for several years (Nankaku M, 2007) (Madsen M, 2004). For example, the patients still report, some years post-surgery, problems particularly related to a difficulty in walking independently (Perron M, 2000); minor leg length discrepancies (Maloney W, 2004); slower gait speed and shorter stride length (Loizeau J, 1995).

In Perron’s article (Perron M, 2000) is possible to find a concrete example, obtained when is compared the gait patterns of 18 women with THA and 13 healthy women. Here, was found a major disability in the frontal plane: the peak abductor moment of force seen at the end of the weight acceptance period. This peak was 15% lower for the women with a THA than for the HLT subjects, as shows Figure 6. They concluded that, this problem is one of the causes of the persistence of abnormal gait patters one year after the THA.

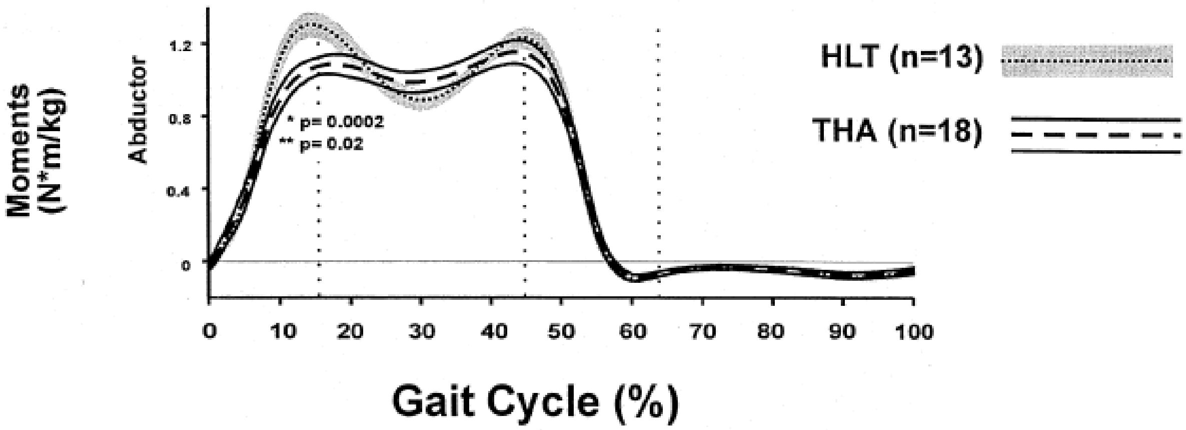

Figure 6. Moment of force at the hip in the frontal plane during the gait cycle normalized at 100% (Perron M, 2000).

Another problem related to THA is the leg length discrepancy. Leg length discrepancy after THA is common and difficult to avoid. The most frequent complications are limping, lumbar pain, neurological damage, patient dissatisfaction, and the need for contralateral shoe lifts for correction (Maloney W, 2004). Leg length discrepancy is also an important factor constraining gait recovery. The extent to which leg length discrepancy impairs motor activity is still controversial. A previous study of Benedetti *et al.* (Benedetti G, 2010) demonstrated that a leg length inequality up to 20 mm does not impair significantly gait and stairs negotiation of THA patients; and the study of Lai *et al.* (Lai K, 2010) showed that, with a discrepancy greater than 2 cm, there was a marked reduction in walking speed and in the length of the step for congenital hip dislocation.

2.3. Outcome measures: Harris hip score

Over the past years there have been changes in the outcomes used in the analysis of the effectiveness of medical treatments or surgical procedures in orthopaedics. Outcomes such as quality of life related to health, functional capacity, pain and satisfaction scales have been emphasized once they provide the analysis of the state of health and manifestations of disease in individuals' lives (Guimarães R, 2010). As a consequence, instruments, questionnaires and scales were developed to describe these kinds of variable. They can be classified as “generic” and “specific”. The generic variables quantify the patient's general state of health, while the specific ones target specific areas of the body and can measure the function with greater responsiveness than a scale that assesses the state of health as a whole (Guimarães R, 2010). Among the clinical scores developed to evaluate disorders of the hip, there are the Harris Hip Score, the Hip Disability and Osteoarthritis Outcome score (Nilsdotter A, 2011), the Short Form, and tests of walking speed and pain during walking (Hoeksma H, 2003).

The multidimensional Harris Hip Score is a specific evaluation tool that has frequently been used to measure outcome after THA. The original version was published in 1969. It presents a rating scale of 100 points with domains of pain, function, absence of deformity, and range of motion (Wamper K, 2010; Nilsdotter A, 2011). The pain domain (with 1 item, covering 0-44 points) measures pain severity and its effect on activities and need for pain medication. The function domain (7 items, 0-47 points) consists of daily activities (sitting, use public transportation, stairs use and managing shoes and socks) and gait (limb, support needed and walking distance). Absence of deformity (1 item, 4 points) takes into account hip flexion, abduction, internal rotation and length discrepancy. Range of motion (2 items, 5 points) measures the hip flexion, abduction, adduction, external and internal rotation (see Appendix A, Figure A1). A total score below 70 points is considered a poor result, 70-80 is considered fair, 80 to 90 is good and 90 to 100 is an excellent result (Nilsdotter A, 2011).

Several studies were performed and the results observed allows to the conclusion that Harris Hip Score has high validity and reliability and is a useful instrument that can be used by a physician or a physiotherapist to study the clinical outcome of hip replacement (Soderman P, 2001).

Chapter 3. Gait Analysis

3.1. Introduction to gait analysis

Gait analysis increased with the rapid development of computer technology during the past two decades, and is now widely used in the evaluation of the efficacy of hip replacement. The history of gait analysis has shown a constant progression from early descriptive studies, through increasingly sophisticated methods of measurement, to mathematical analysis and mathematical modelling (Whittle M, 2007). Initially was used by experienced observers. Later it was augmented by instrumentation: measuring body movements, body mechanics and activity of the muscles.

The human gait comprises a sequence of rapid and complex events giving to each individual a unique gait pattern. It is hard to analyse these phenomena by clinical observation, and to quantify the degree of departure from normality. Such limitations have led doctors, physiotherapists, biomedical engineers and researchers of the movement to develop gait analysis.

Therefore, gait analysis is the systematic study of human locomotion, more specifically it is the study of the human motion during a walking task using observational methods, augmented by instrumentation for measuring body movements, body mechanics, and the activity of the muscles. It is also used to assess, plan, and treat individuals with conditions affecting their ability to walk. It is commonly used in sports biomechanics to help athletes run more efficiently and to identify posture-related or movement-related problems in people with injuries.

The study encompasses quantification, that is, introduction and analysis of measurable parameters of gait, including evaluation of velocity, cadence, stride length, single-limb support (SLS or SS) and double-limb support (DLS or DS) time (percentage of gait cycle) (Kelly K, 1998), as well as interpretation, e.g. drawing conclusions about the subject's health status from his/her gait.

Nowadays, comprehensive gait analysis usually includes kinematics, kinetics and electromyography and this complex information can only be obtained in a specialized laboratory (Illyés A, 2005). Kinematics, kinetics, and electromyography are fundamental for the purpose of characterising gait patterns and their underlying mechanisms.

3.2. Statistical gait analysis and instrumentation

The purpose of statistical gait analysis is to describe gait functionally, analyzing several tens or hundreds of consecutive steps and it is intended to evaluate the patient during a "functional" walk, typical of the daily life. When traditional gait analysis is applied, generally only two or three consecutive steps may be analysed and this is not enough to assess a number of gait abnormalities.

Gait analysis studies involve the processing of continuous data signals measured over many gait cycles. Signal analysis and interpretation requires adequate statistical methods that include the statistical characterization of spatio-temporal parameters, joint angles curves and parameters derived from electromyographic (EMG) signals. These quantitative measures, in conjunction with observational, qualitative measures, can provide a quick and easy assessment that can be repeated while tracking the recovery or rehabilitation of a patient.

It is important to refer that, to obtain a high repeatability of the results, it is convenient to study gait cycles relative to steps executed while walking along a straight path. In order to record a sufficient number of gait cycles, the subject is asked to walk back and forth over the straight walkway. During acceleration, deceleration and changes of direction, the steps are different from those relative to "regime" walk (Agostini V, 2012). Thus, to obtain results highly repeatable and independent from the path length, it is appropriate to isolate these "regime" steps using automatic and user-independent methods.

Three different kinds of signals are generally considered in this analysis: a) foot-switch signals, b) signals coming from goniometers attached to the different lower limb joints, and c) myoelectric signals.

3.2.1. Gait cycle and basography

Walking uses a repetitious sequence of limb motion to move the body forward while simultaneously maintaining stance stability (Perry J, 1992). Initially, one limb acts as source of support while the other limb advances itself to a new support site and then the limbs reverse their roles. This series of events is repeated over and over again during a pathway. A single sequence of these functions by one limb is called a gait cycle (GC).

The gait cycle is defined as the time interval between two successive occurrences of one of the repetitive events of walking. In the normal gait cycle, limbs move in a symmetrical alternating relationship, which can be described by a phase lag of 0.5 (Shumway-Cook A, 2007). This means that one limb initiates its step cycle when the opposite limb reaches the midpoint of its own cycle.

3.2.1.1. Gait cycle and its phases

The gait cycle is split into two main phases: stance, which starts when the foot strikes the ground, and swing, which begins when the foot leaves the ground. Stance phase of gait begins with initial contact and is divided into four periods: loading response, midstance, terminal stance, and pre-swing. Swing phase begins as the foot is lifted from the floor (toe-off) and is divided into three periods: initial swing, midswing, and terminal swing (Kharb A, 2011).

The beginning and ending of each period are defined by specific events. The major events during the gait cycle are: initial contact, opposite toe off, heel rise, opposite initial contact, toe off, feet adjacent, tibia vertical and a new cycle starts once again with initial contact.

Although any event could be chosen to define the gait cycle, it is generally convenient to use the instant at which one foot contacts the ground, which is the initial contact. Thus, the complete cycle starts from the instant in which one foot touches the ground and ends when the same foot touches the ground again. In Figure 7, it is represented a gait cycle relative to the right leg.

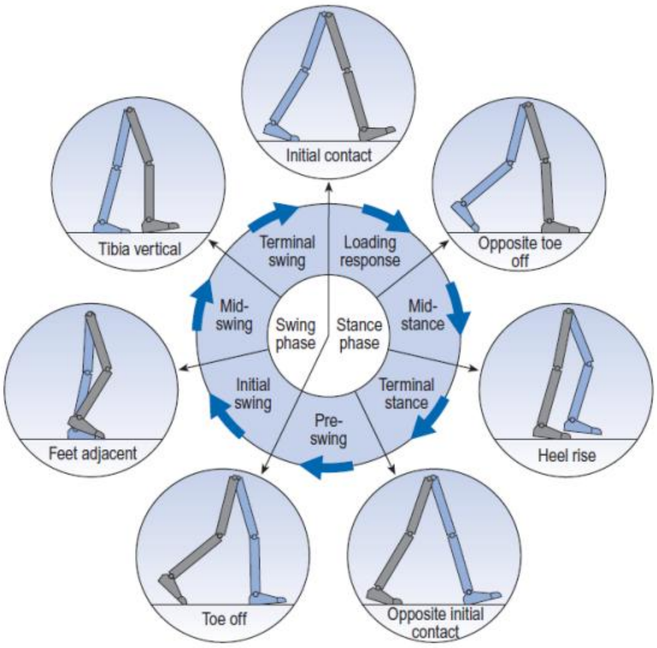

Figure 7. Positions of the legs during a single gait cycle by the right leg (Whittle M, 2007).

As we can see in the figure above, in stance phase, loading response begins with initial contact, in the instant the foot contacts the ground. Usually, the heel contacts the ground first but, in some cases, for example in patients who demonstrate pathological gait patterns, the entire foot or the toes contact the ground first. Loading response ends with opposite toe off, when the opposite extremity leaves the ground. Thus, loading response corresponds to the gait cycle's first period of double limb support. Midstance begins with opposite toe off and ends when the centre of gravity is directly over the reference foot. Terminal stance begins when the centre of gravity is over the supporting foot and ends when the contralateral foot contacts the ground. During terminal stance, the heel rises from the ground. The last phase, preswing, begins at opposite initial contact and ends at toe off, at around 60% of the gait cycle. Thus, preswing corresponds to the gait cycle's second period of double limb support.

In swing phase, the initial swing begins at toe off and continues until maximum knee flexion (60 degrees) occurs. Midswing is the period from maximum knee flexion until the tibia is vertical or perpendicular to the ground. Terminal swing begins when the tibia is vertical and ends at initial contact.

When running, a higher proportion of the cycle is swing phase, as the foot is in contact with the ground for a shorter period. Because of this there is no double stance phase, and instead there is a point where none of the feet are in contact with the ground: the flight phase (Kharb A, 2011). As running speed increases, stance phase becomes shorter and shorter.

3.2.1.2. Gait cycle time

During walking, there is a period when both feet are in contact with the ground, called double support, which represents approximately the first and the last 10% of the stance phase. Single support phase is the period when only one foot is in contact with the ground. This consists of the time when the opposite limb is in swing phase (Shumway-Cook A, 2007). The single support phase can be divided in right single support, when only the right foot is on the ground and ends with initial contact by the left foot; and left single support, that corresponds to the right swing phase and the cycle ends with the next initial contact on the right, as shown in Figure 8.

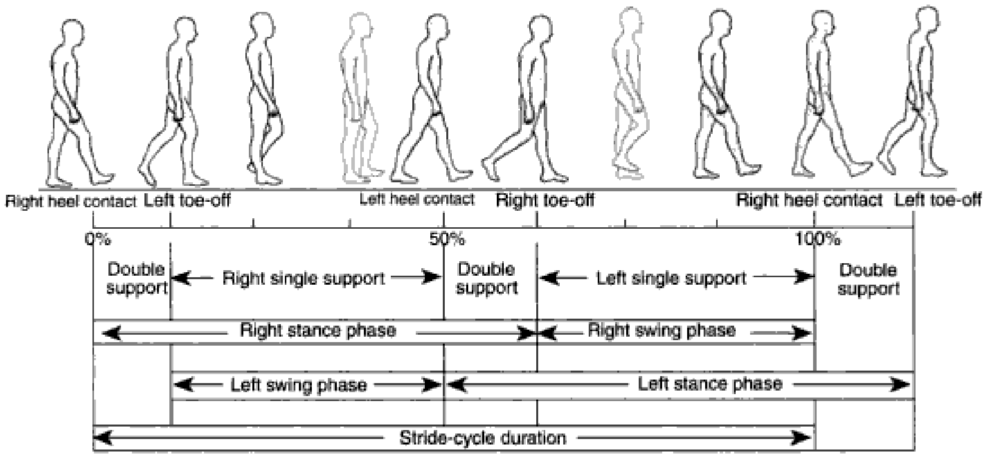

Figure 8. Temporal and distance dimensions of the gait cycle. Swing and stance phase characteristics (Shumway-Cook A, 2007).

In each gait cycle, there are thus two periods of double support and two periods of single support (Kharb A, 2011). At freely chosen walking speeds, adults typically spend approximately 60% of the cycle duration in stance phase, and the remaining 40% in swing and each period of double support lasts about 10% of the gait cycle.

3.2.1.3. Typical and atypical cycles

During gait, the basographic signals coming from both feet are collected continuously. To describe the contact of the foot on the floor and measure the corresponding temporal gait parameters, it can be used from 2 to 4 sensors. When using 3 sensors, as we used in this analysis, it is possible to distinguish among 8 different conditions of support. However, usually, in statistical gait analysis, the 8 level basography is simplified in order to obtain the correspondent 4 level basography. In Figure 9, it is possible to observe the differences between 8-level basography and 4-level basography of a single gait cycle.

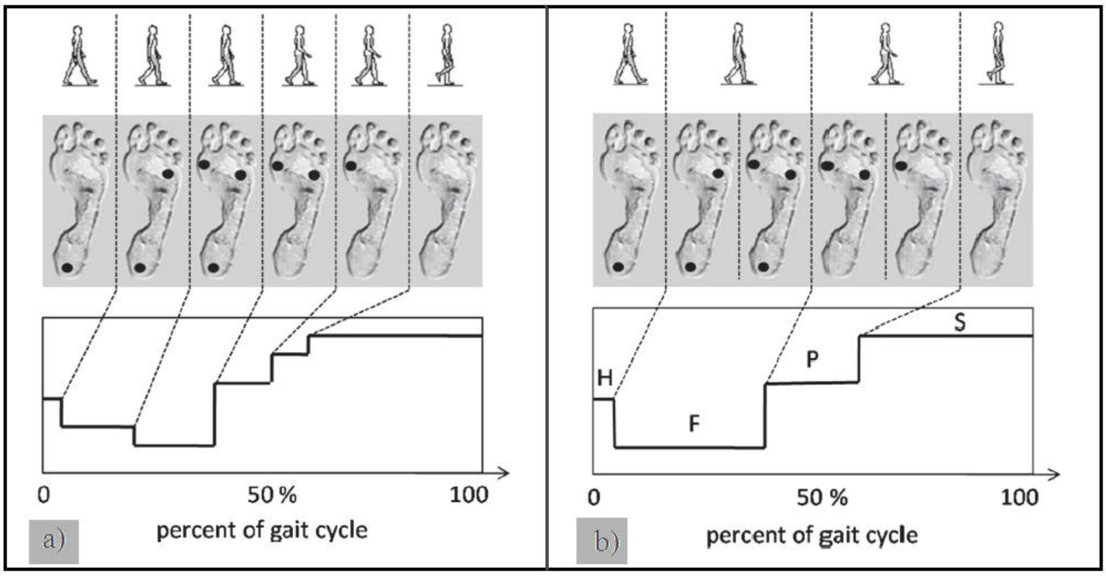

Figure 9. a) 8-level basography; b) 4-level basography of a single gait cycle (Agostini V, 2012).

Different basographic cycles may be observed during a walk. The most frequent basographic cycle in healthy subject is represented by a sequence of Heel contact (H), Flat foot contact (F), Push off (P) and Swing (S) and can be considered the “typical” foot-floor contact sequence. It is important to refer that even in healthy subjects, there is also a small percentage of “atypical” cycles (as PFPS and FPS, for example), usually less than 5%.

The basography analysis is fundamental since it gives a “time reference frame” in which to evaluate the behaviour of all the other signals (Agostini V, 2012). After each basographic cycle has been correctly classified, the mean, standard deviation, and standard error of each basographic phase are calculated, separately for each specific typology of gait cycle detected. In typical HFPS cycles it is usually interesting to obtain also mean values, standard deviations, and standard errors of the single and double supports.

3.2.1.4. Basography

Basographic systems consist of integrated sensors made in such a way as to give a signal while the subject is made to walk along a pathway. Foot-switches are particularly useful for the synchronization and the evaluation of the temporal parameters of gait. They make it possible to collect the temporal data relative to the foot-floor contact phase. In order to allow a complete analysis of the stride phase, the foot-switches are placed in 3 independent zones: heel, first and fifth metatarsal heads. The footswitches are usually connected through a wire to a computer.

The signal acquired while the subject is walking, is converted into a 4-level signal (HFPS) and is then segmented in strides. If switches are mounted on both feet, the single and double support times can also be measured. In particular, the acquisition of events such as heel-strike and toe-off of both feet makes it possible to identify in temporal terms the phases of initial contact, loading response, terminal stance, pre-swing and swing.

3.2.2. Joint angles and goniometry

Generally speaking, the hip angle is defined as the angle between the femur and acetabulum of the pelvis and the knee angle is the angle between the femur and the tibia. The ankle angle is usually defined as the angle between the tibia and an arbitrary line in the foot (Whittle M, 2007). Figure 10 shows in the left side the successive positions of the right leg at 40 ms intervals, measured over a single gait cycle and in the right side shows the corresponding sagittal plane angles (in degrees) at the hip (flexion positive), knee (flexion positive) and ankle joints (dorsiflexion positive).

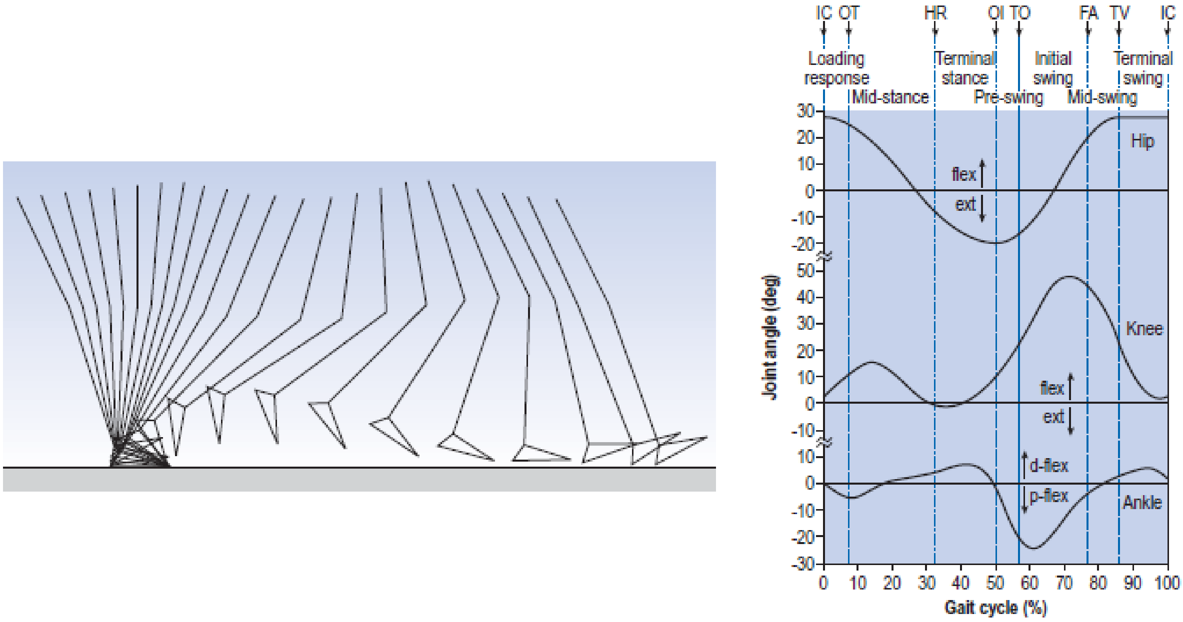

Figure 10. Left side: Position of the right leg in the sagittal plane at 40 ms intervals during a single gait cycle; Right side: corresponding sagittal plane angles at the hip, knee and ankle joints. IC = initial contact; OT = opposite toe off; HR = heel rise; OI = opposite initial contact; TO = toe off; FA = feet adjacent; TV = tibia vertical (Whittle M, 2007).

Range of motion is a description of how much movement exists at a joint. It refers to the distance and direction a joint can move between the flexed position and the extended position. Limited range of motion refers to a joint that has a reduction in its ability to move. The reduced motion may be a mechanical problem with the specific joint or it may be caused by diseases such as osteoarthritis, rheumatoid arthritis, or other types of arthritis. Pain, swelling, and stiffness associated with arthritis can limit the range of motion of a particular joint and impair function and the ability to perform usual daily activities.

Each specific joint has a normal range of motion that is expressed in degrees. Devices to measure range of motion in the joints of the body include the goniometer and inclinometer which use a stationary arm, protractor, fulcrum, and movement arm to measure angle from axis of the joint.

3.2.2.1. Goniometry

A goniometer is in general an instrument that allows to study the joint angles during the continuous movement. The electrogomometer we use in this analysis is a device consisting of two articulated parallelograms attached to two segments, which allows measuring joint angles in one or two planes. When fixed to the joint segment, it supplies a precise measurement of the relative instantaneous angles between the two segments. They offer a simple and affordable alternative to motion capture systems, allow the joint angle data to be collected and viewed instantaneously, and prove highly accurate (Zhao S, 2010). Due to their structure based on an articulated parallelogram, they do not require the alignment of the potentiometer shaft with the instantaneous centre of rotation of the joint.

During the gait, the goniometers signals coming from the joints are collected continuously for the two legs. The goniometric signals are then low-pass filtered by means of a digital low-pass filter with a cut-off frequency usually in the range 10-20 Hz (Agostini V, 2012).

It is easy to use, and can procedure a large amount of reliable and reproducible data with an accuracy of about 1 degree and repeatability higher than 0.5 degrees (Agostini V, 2012). The resulting data are available in real time and do not require long data reduction process. The costs are relatively low.

3.2.3. Muscle activity and electromyography

Muscle activity is typically studied using electromyography. EMG signals differ among individuals and for a single individual, depends on variables such as velocity. Furthermore, muscles show different activation patterns during walking, even if we consider a specific subject during a single walking session along a straight path.

Valentina *et al.* (Valentina A, 2010) demonstrated that there are various patterns of muscle activation during gait: each muscle usually shows 1 to 5 activations during the gait cycle.

When analyzing EMG signals, it is desirable to obtain, for each muscle, the different activation patterns and how frequently they are observed. In normal walking, muscles contract and relax in a precise and characteristic moment of the gait cycle, depending on the biomechanical task that it is dealt with. Some of them are active primarily in stance phase or primarily in swing phase.

The goal of stance phase is to prepare for weight bearing. At initial contact, a deceleration of the limb begins by simultaneously activating the knee extensor and flexor muscles to stabilize and position the knee in space before it accepts weight (Rose J, 2006). The hip extensors slow the forward movement of the leg with an eccentric contraction.

During loading response, the ankle dorsiflexors eccentrically contract as the foot reaches the ground. The knee extensors contract eccentrically as the knee bends, to accept the weight of the body, but as the knee extends the contraction changes to concentric (Rose J, 2006). The gluteus medius muscle isometrically contracts in order to stabilize the pelvis.

During midstance, gluteus medius acts as a hip abductor to stabilise the pelvis as the contralateral leg swings through, while the triceps surae prevents excessive dorsiflexion of the ankle and then prepares to drive the person forward (Vaughan C, 1992; Perry J, 1992). During midstance and midswing, most muscles (with the exception of gluteus medius and triceps surae during stance, and tibialis anterior during swing) are relatively quiescent (Perry J, 1992). This is interesting because it is during these two periods (midstance and midswing) that the greatest observable movement takes place.

At terminal stance the body accelerates forward and nearly all the muscle work is generated by a shortening contraction of the ankle plantar flexors. This burst of energy is responsible for most power generation that keeps the body moving forward in normal gait. There may be a small burst of iliopsoas activity to lead into the unloading response of preswing. Just before toe off, hip flexors concentrically contract in order to prepare the leg for swing phase and therefore unloading.

In preswing phase, the ankle plantar flexors are no longer active, and the hip flexors (iliopsoas and rectus femoris) begin to lift the limb and swing it forward, generally by concentric contraction. The energy consumed in preswing muscle activity is efficiently brief since the limb behaves like a passive pendulum for the most part of the swing phase.

During swing phase, most of the lower limb muscles are inactive and the leg swings freely like a pendulum. At the Initial swing, the ankle dorsiflexors contract concentrically to allow the foot to clear off the ground and remain contracted throughout the whole swing phase.

During midswing, the tibialis anterior provides active dorsiflexion and thus prevents the toes from dragging on the ground (Vaughan C, 1992). Midswing sees continuation of the passive pendulum action of the leg.

At terminal swing, the goal is to decelerate the leg and prepare it for weight acceptance and the hamstrings contract either isometrically or eccentrically in order to slow both hip flexion and knee extension. The contraction in the ankle dorsiflexors changes from concentric to isometric or eccentric.

3.2.3.1. Electromyography

Electromyography (EMG) is an experimental technique concerned with the acquisition, recording and analysis of myoelectric signals. Myoelectric signals are formed by physiological variations in the state of muscle fibre membranes (Konrad P, 2005). The electric signal coming from muscles activity can be measured by electrodes and represents a highly complex wave form whose shape depends on the type and location of the electrode, the number of motor unit action potentials detected, the spatial geometry of the motor unit itself, and filtering characteristics of muscle tissue (Rose J, 2006).

Electromyographic records give information about the timing of muscles activity and the relativity intensity of muscle activity during a movement, the action potentials of the motor units. By processing the raw EMG signal electrically and mathematically, information about force generation, motor unit recruitment, and muscle fatigue may be extracted (Zhao S, 2010).

This technique can be used to detect abnormal gait behaviour and assess neuromuscular control. In addition, the frequency content of the EMG signal can be analysed to identify neural injury, denervated muscle, or primary pathologic processes.

3.2.3.1.1. Electrodes

Basically, an electrode is a transducer, a device that converts one form of energy into another, in this case ionic flow into electron flow (Vaughan C, 1992).

The EMG signal is based upon action potentials at the muscle fibre membrane resulting from depolarization and repolarization processes. The extent of this depolarization zone (Figure 11) is described in the literature as approximately 1-3mm^2 (Konrad P, 2005). After initial excitation this zone travels along the muscle fibre at a velocity of 2-6m/s and passes the electrode side:

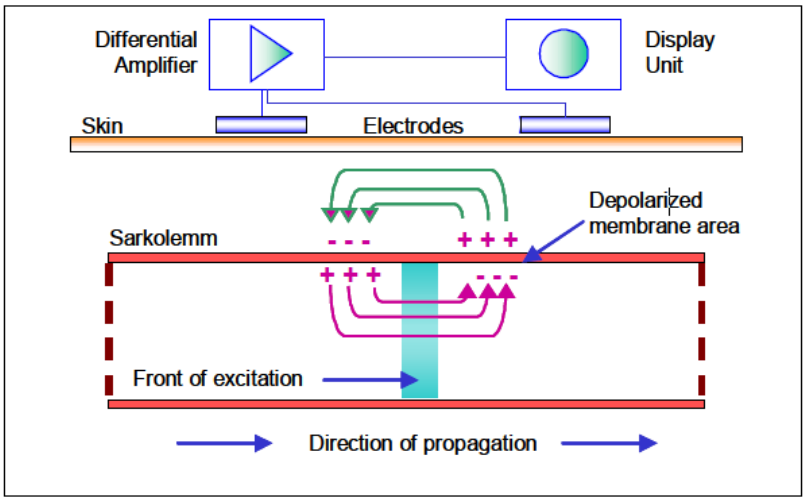

Figure 11. The despolarisation zone on muscle fibre membranes (Konrad P, 2005).

Both fine wire electrodes and surface electrodes are used for EMG analysis. Usually, the use of invasive techniques is reserved for the study of deep muscles, namely muscles that cannot be directly accessed from the skin. Due to their non-invasive character in most cases surface electrodes are used in kinesiological studies. Besides, the data that they provide are more repeatable than wire electrode data, but show less discrete phases of muscles action. Despite the benefit of easy handling, their main limitation is that only surface muscles can be detected (Konrad P, 2005).

In general, surface probes consist of 2 or 3 detection surfaces made of a conductor material and, in active probes, an amplifier stage is positioned very close to the electrodes (Agostini V, 2012). Nowadays, active probes are preferred to passive ones due to their better performance and ease of use.

The EMG signal can be influenced by several external factors altering its shape and characteristics. The most common when using surface probes is crosstalk, which occurs when neighbouring muscles produce a significant amount of EMG signal that is detected by the local electrode site, even if the muscle under study is not active. Typically this phenomenon does not exceed 10%-15% of the overall signal contents (Konrad P, 2005). To avoid this problem it is advisable to use probes with an electrode distance slightly higher than the thickness of the tissue interposed between the surface of the muscle to be observed and skin. Computer analyses quantify muscle activity.

3.2.4. Other gait parameters

The cyclic nature of human gait is a very useful feature for reporting different parameters. There are literally hundreds of parameters that can be expressed in terms of the percentage cycle. The general gait parameters usually include cadence, velocity, stride length and stride time. These quantitative measures, in conjunction with observational, qualitative measures, can provide a quick and easy assessment that can be repeated while tracking recovery or rehabilitation (Dugan S, 2005). General parameters specific to gait activity such as time-distance parameters are potentially measurable from images by computer vision techniques.

The cadence is the number of steps taken in a given time. There are two steps in a single gait cycle, and the cadence is a measure of half-cycles. However, in this thesis, the cadence is calculated in strides per minute, which means one gait cycle (right step+ left step). A person's cadence can be calculated using the formula.

(Equation 3.1)

Velocity is the distance covered in a given time and is calculated as follows:

(Equation 3.2)

Stride length is the distance (in meters) measured between the initial heel contact of a gait cycle and the heel contact of the subsequent cycle. It can be determined in two ways: by direct measurement, or calculating it from velocity and cadence. To determine the calculated stride length, measure cadence and velocity, and then use the following formula:

(Equation 3.3)

Finally, stride time, also known as the "cycle time", in seconds, is:

(Equation 3.4)

All these parameters provide the simplest form of objective gait evaluation. Cycle time, stride length and speed tend to change together in most locomotor disabilities (Whittle M, 2007), so that a subject with a long cycle time will usually also have a short stride length and a low speed (speed being stride length divided by cycle time). Variations in time-distance values often are pathology-specific.

Thus, these general gait parameters give a guide to the walking ability of a subject, but little specific information. They should always be interpreted in terms of the expected values for the subject's age and sex. There are in literature some normative tabled values related to these gait parameters which can be used for comparison.

3.3. Applications of gait analysis to hip prosthesis

Gait abnormalities may result from neurological, orthopaedic and also systemic disorders. According to Whittle *et al.* (Whittle M, 2007), a large number of diseases affect the neuromuscular and musculoskeletal systems and may thus lead to disorders of gait. Among the most common are cerebral palsy, Parkinson, muscular dystrophy, osteoarthritis, rheumatoid arthritis, lower limb amputation, stroke, head injury, spinal cord injury, myelodysplasia and multiple sclerosis.

Gait analysis is widely used in clinics to study gait abnormalities for surgery planning, definition of rehabilitation protocols, and objective evaluation of clinical outcomes (Agostini V, 2012). In this section will be presented some examples application of gait analysis in the evaluation of hip prosthesis.

There are studies concerning the contribution that gait analysis can give to decide among different kinds of hip intervention. For example, Lavigne *et al.* (Lavigne M, 2008) carried out a study comparing different replacement types: RHA (resurfacing hip arthroplasty), standard THA, and THA using a large diameter femoral head. They found better gait measurements in patients with RHA or with THA using large diameter heads than in those with standard THA. Also Mont *et al.* (Mont M, 2007) found improved gait parameters (speed of walking, abduction moments) after RHA when compared to standard hip arthroplasty.

In another study, Loizeau *et al.* (Loizeau J, 1995) discovered that a group of patients subjected to a total hip prosthesis had a slower gait speed, shorter stride length and spend more time in stance than the able-bodied group. Furthermore, the non-operated knee and hip displayed lower energies than the able-bodied subjects confirming the presence of some mechanical dysfunction, indicating that eventually orthopaedic problems may occur at the contralateral hip.

Perron *et al.* (Perron M, 2000) studied three-dimension gait analysis in women that underwent a THA and they discovered a decrease in gait speed and the persistence of abnormal gait patterns one year after the total hip arthroplasty. These facts were associated respectively with a decrease in the hip extensor moment of force and with a decrease in the range of hip extension (sagittal plane) or in the hip abductor moment of force (frontal plane).

Walking efficiency and the lateral displacement of the trunk in patients in early stages after total hip arthroplasty was studied by Nankaku *et al.* (Nankaku M, 2007) and the results obtained suggested that exists a trunk compensation strategy for hip abductor weakness in patients soon after THA that can lead to increased energy expenditure. This occurs because the THA patients need more energy to progress their body forward in a gait cycle that causes a reduced walking efficiency.

Chapter 4. Materials and methods

4.1. Subjects

In this study we analysed a population of 20 patients and 20 healthy controls matched for gender and age. Able-bodied subjects had to be free of any past or present condition that could affect walking and THA patients with coxoarthrosis also in the contralateral limb were excluded from the study.

Patients underwent unilateral THA surgery with posterior-lateral incision and the indication for the surgery was hip coxoarthrosis. After surgery, they were submitted to a muscular rehabilitation program and instructed to use first two crutches, then removing one crutch and finally without crutches to restore the load of the operated limb in a gradual manner.

Patients have been evaluated at 3, 6, and 12 months after surgery with clinical examination using the Harris Hip Score and with an instrumented gait analysis. The results of the longitudinal evaluation of the Harris Hip Score for the THA patients are presented in Table 2. In Appendix B, in Table B1 are the details of the anthropometric data, and in Table B2 is reported the leg length discrepancy and Harris Hip Score results.

Table 2. Results of the clinical examination using Harris Hip Score. *

	3 Months	**6 Months**	**12 Months**
Harris Hip Score	90,0±7,9 (73-100)	96,6±5,0 (83-100)	98,4±2,8 (89-100)

*Data were presented as mean ± standard deviation (range).

Before the gait analysis test, patients and controls underwent a physical examination and anthropometric data were collected for each subject. The mean values of age, height and weight for the two populations are reported in Table 3, as well as the leg length discrepancy after surgery and body mass index (BMI).

Table 3. Anthropometric characteristics of the THA patients and control group. *

	THA patients (N=20)		**Controls (N=20)**	
Gender (M/F)	9 Males	11 Females	11 Males	9 Females
Age (yr)	65,2±7,4 (55-79)	66,8±7,3 (49-74)	65,1±5,0 (57-74)	65,8±5,4 (58-74)
Weight (kg)	80,2±10,7 (60-92)	74,3±15,0 (59-100)	76,1±11,1 (60-96)	60,3±6,6 (51-69)
Height (cm)	175,1±7,7 (165-185)	163,4±9,6 (150-179)	175,8±7,7 (166-193)	162,4±5,1 (155-170)
BMI (kg/m^2)	26,1±2,1 (22,0-27,8)	27,8±4,7 (20,9-34,5)	24,6±2,7 (20,5-29,0)	22,9±2,4 (19,1-26,3)
Leg length discrepancy (cm)	0,3±0,4 (0-1)	0,8±0,5 (0-1,5)	-	-

*Data were presented as mean ± standard deviation (range).

Patients were recruited from the Rehabilitation and Functional Recovery Unit at the Ivrea Hospital (Torino, Italy). The experimental protocol was approved by the local ethical committee and all participants gave their written informed consent to be included in the study.

4.2. Experimental protocol and set-up

Patients have been evaluated at 3, 6 and 12 months after surgery, with an instrumented examination based on statistical gait analysis using the system Step 32 (DemItalia, Italy). The contralateral leg makes part of this study to investigate if the healthy leg suffers changes in gait strategy to compensate for what happens in the operated leg.

In each session, during approximately 2h, patients were equipped in both legs with: a) foot-switches attached beneath the heel, the first and the fifth metatarsal heads, b) knee and hip goniometers positioned in the sagittal plane, c) surface EMG electrodes positioned over tibialis anterior (TA), gastrocnemius lateralis (LGS), rectus femoris (RF), lateral hamstrings (LH) and gluteus medius (GMD). EMG probes were placed oriented longitudinally over the muscle fibres, according to the guidelines suggested by

Winter (Winter D, 1991). In Figure 12 it is shown the configuration and location of the probes used during a data acquisition section, in one patient.

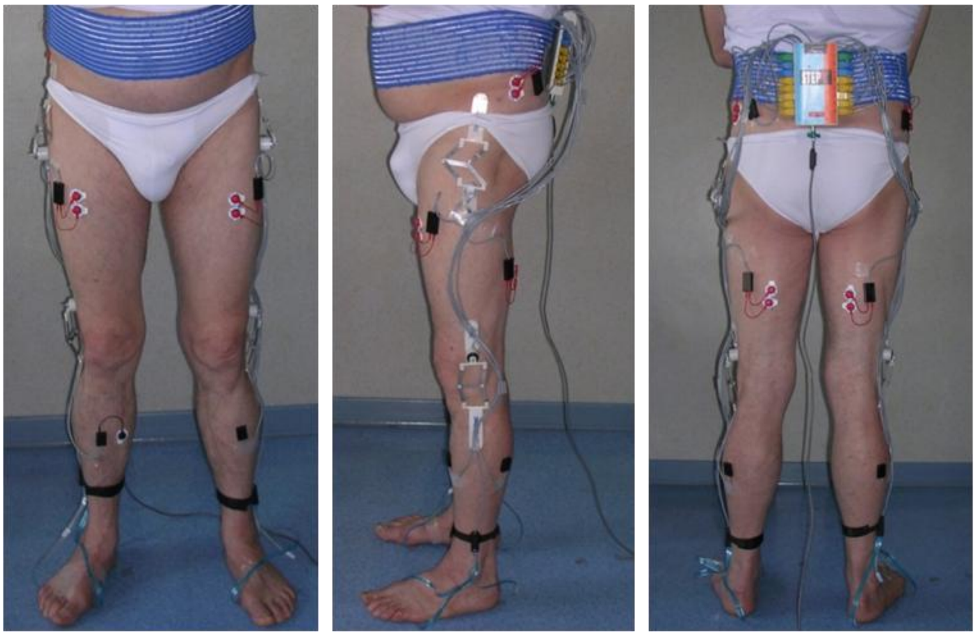

Figure 12. Probes positioning in one of the patients (system Step 32).

After the probes positioning, patients were asked to walk forward and backward over 10-m straight track at self-selected speed, and as fast as they could still feeling save. Each acquisition lasted 150 s, with un acquisition frequency of 2kHz, and the recorded file is then saved in the computer.

4.3. Step 32

Styles of gait analysis systems vary. At present they mainly include motion capture systems, force plates, electromyography (EMG), and sensors, including accelerometers, electrogoniometers, gyroscopes and pressure sensors, which are small and portable (Zhao S, 2010). In the present work was used the medical system Step 32, created for statistical gait analysis (producer: DemItalia, Italy). This system allows the study of signals coming from foot switches, goniometers, and surface EMG probes without any user interaction working with hundreds of steps, in very realistic situations.

The basic configuration of STEP 32 consists of: a workstation based on a personal computer running on Windows XP™ operating system; a video camera that allows to acquire video recordings synchronized with the gait signals; a proprietary data acquisition board; a thin cable 12-meter long that connects the patient unit, usually fixed to the patient's waist, to the workstation; a set of different sensors (foot switches, goniometers, active probes for surface or indwelling EMG electrodes); the patient unit itself; and, finally, the STEP 32-DV software package. Figure 13 shows the basic configuration of STEP 32.

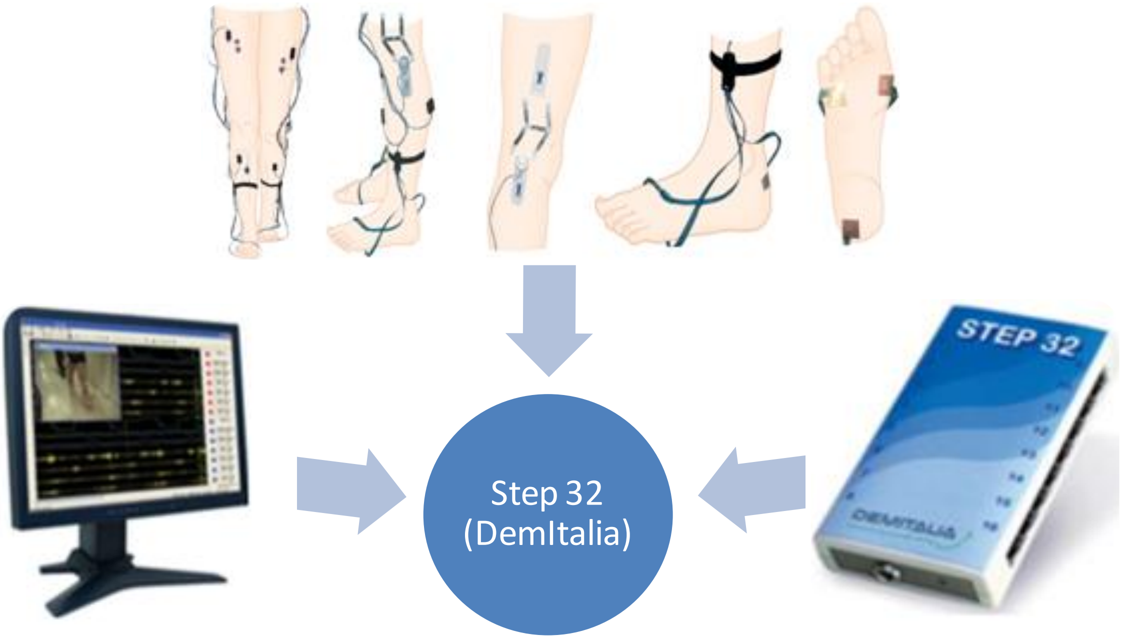

Figure 13. Basic configuration of STEP 32. Adapted from (DemItalia, 2012)

Due to the large number of strides required for a statistical analysis of gait, it is important to execute the gait segmentations and classification automatically and in a user-independent way. Thanks to its proprietary processing algorithms, it analyses in a few seconds hundreds of steps, thus allowing for evaluating the patient’s performances in realistic situations. Results are reliable and repeatable, independently from the user expertise. (DemItalia, 2012)

The acquired data were offline statistically processed by the system software. The statistical gait analysis system automatically excludes non-regime strides like those recorded during turns and acceleration-deceleration phases. Traditional gait analysis usually divides the gait cycle only in stance and swing. However, step 32 is able to detect sub-phases of stance, which are the Heel contact (H), the Flat foot contact (FFC) and the Push off (P) phase; the swing phase; double support; velocity and cadence. For each gait parameter the mean value of right and left sides was calculated and used for statistical analysis.

4.3.1. Step software and data analysis

The procedure to access the results in the system Step 32 is given by a sequence of steps, shown in the diagram below, in Figure 14. First it is necessary to open the program and then (1) find the patient to analyse and press "Gait Analysis"; (2) select the exam (the acquisition at 3, 6 or 12 months post-operative) and press "view exam"; (3) select the acquisition (self-selected speed or fast speed) and run the data analysis; (4) create a new assistant results by clicking in "New Ass. Result"; (5) select reference gaits on both sides and press "OK" button (usually HFPS-HFPS); (6) set the results name; (7) select the file and press "View Results". After that, is possible to see the gait cycle results (8), the goniometric results (9), and EMG results (10) with EMG activations (11). After this, the results are exported to an excel file to be read by the software MATLAB version 7.11.0 (R2010b). This sequence is repeated to analyse the 5 muscles of the leg, at both sides, at both trials (self-selected speed and fast speed), for the data at 3, 6 and 12 months after surgery (in patient's case).

This procedure was performed for 20 patients (in 5 muscles of the operated leg and 5 of the sound leg, at self-selected speed and fast speeds, 3, 6 and 12 months after the operation) and 20 controls (in 5 muscles of the right leg and 5 of the left leg, at self-selected speed and the fast speed), resulting in a total of 1600 EMG signals analysed. In Appendix C, Table C1, is reported one representative excel file containing the numerical values resulting from the EMG analysis.

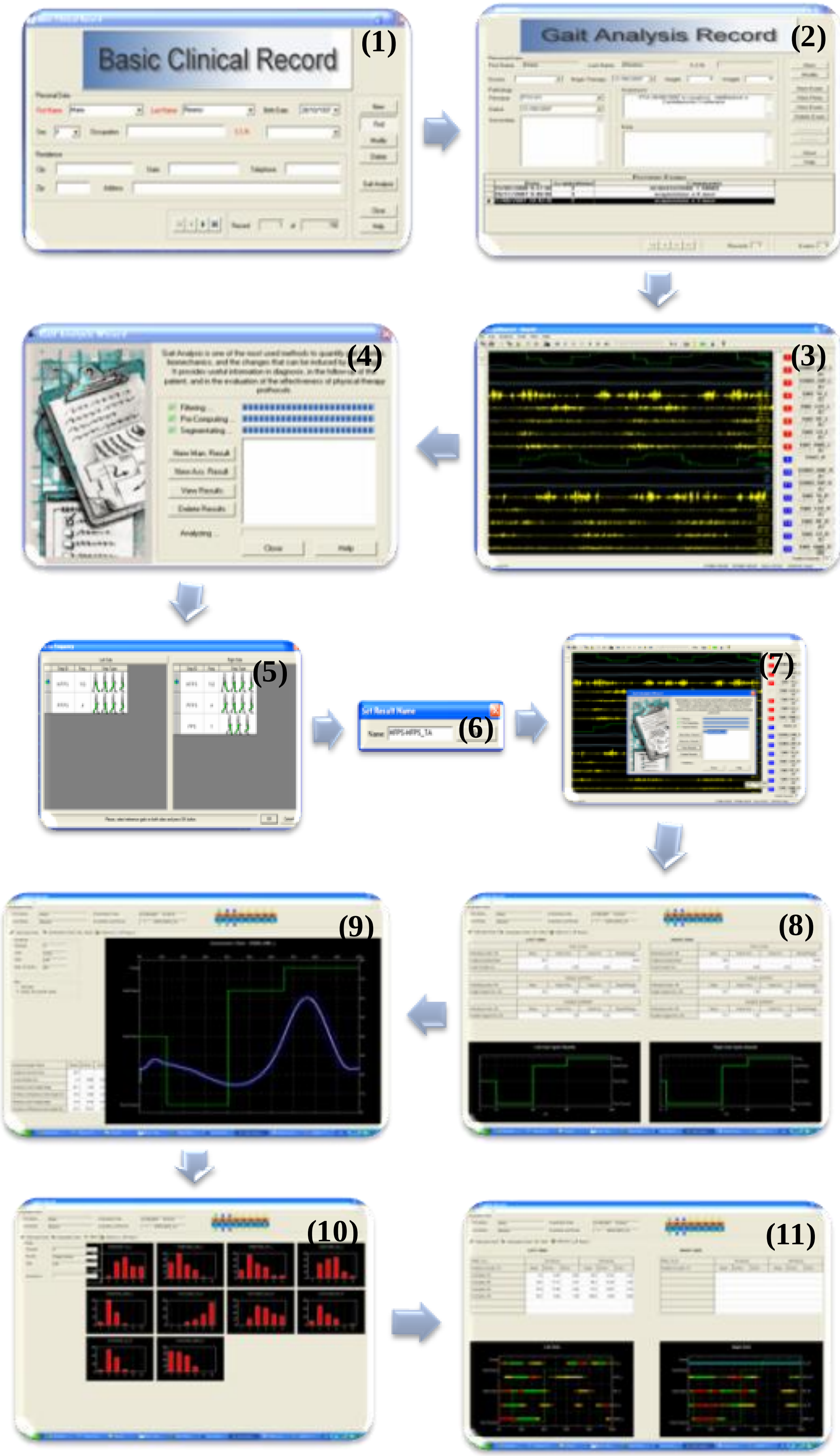

Figure 14. Diagram showing the sequence of interfaces of the system Step 32.

4.3.2. Signal processing

Signals are recorded using the Step 32 system that acquires and records the surface electromyographic bilateral data coming from the muscles together with foot-switch and goniometric signals.

Foot-switch signal is debounced and converted to a 4-level signal (HFPS) and is then segmented in strides and the different stride typologies performed by the subject during the walk are classified (Agostini V, 2012). The percentage frequency of typical and also the atypical strides are calculated for each gait analysis test.

Goniometric signal is low-pass filtered (FIR filter, 100 taps, cut-off frequency of 15 Hz) and the delay introduced by the filter compensated. The goniometric signal and the duration of foot-contact gait phases are then used by a multivariate statistical filter to discard outlier strides, i.e., strides with the proper sequence of gait phases (HFPS) but with abnormal timing, like those relative to deceleration, reversing, and acceleration.

EMG signal is high-pass filtered (FIR filter, 100 taps, cut-off frequency of 20 Hz) and the delay introduced by the filter compensated. The signal is then processed by a double-threshold statistical detector of muscle activation (Bonato P, 1998) to obtain, in a user-independent way, the muscle activation intervals. This detector operates on the raw myoelectric signal and, hence, it does not require any envelope detection.

4.4. Statistical analysis

For each lower limb of a single subject, we consider 145±25 strides (mean ± SD) collected during the same walk. This allows to adopt a “statistical gait analysis” approach (Agostini V, 2012), ensuring repeatable and accurate results.

For each subject and each test condition (self-selected and fast speed) we calculate the percentage frequency of atypical strides. Then, only “normal” HFPS strides are averaged to calculate spatio-temporal, kinematics and EMG parameters. For each parameter, we first average the values obtained for a single subject in a single trial. Then, the average over the population is calculated.

The temporal-spatial variables analysed were velocity, cadence, double support, single support (for both legs) and phases of basographic cycle of the affected and unaffected legs.

The kinematic variables analysed were the range of motion on the hip and knee.

During a walk, a subject shows different muscle’s activation patterns. Hence, for each walk and each muscle, we calculated the relative frequency of strides showing from one to five activations (Agostini V, 2010).

Chapter 5. Results

5.1. Basographic results

The mean and standard deviation of velocity is presented in Table 4 for patients and controls, in both trials.

Table 4. Velocity (m/s) for patients and controls, in both trials.

	THA patients			Controls
	3 Months	**6 Months**	**12 Months**	
Self-selected speed	0.78 (0.10)	0.92 (0.18)	1.00 (0.22)	0.99 (0.17)
Fast speed	1.15 (0.16)	1.24 (0.22)	1.30 (0.32)	1.37 (0.13)

Both at self-selected speed and fast speed, the velocity of THA patients increases in the postsurgical follow-up, as expected. One year after the operation, THA patients reach normal values of velocity.

For the self-selected speed, the percentage of atypical strides in THA patients and controls is shown in Figure 15. In Figure 16 shows the same graph for fast speed. In Appendix D, it is possible to find the details of the typical and atypical strides for each patients (3, 6 and 12 months after surgery) and controls, for both trials, from where these figures were obtained.

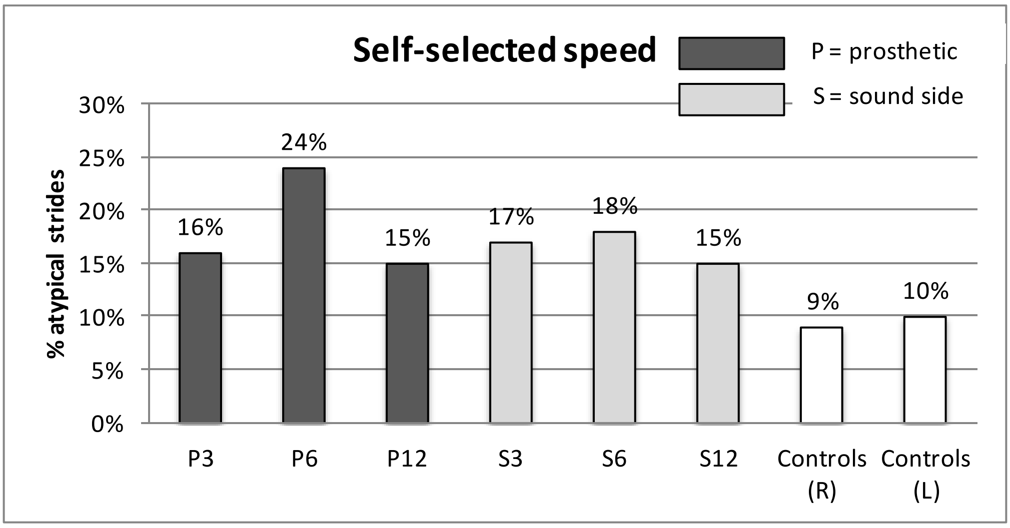

Figure 15. Percentage of atypical strides in THA patients and controls at self-selected speed (represented by the mean and standard error over the population). P3, P6 and P12 represents the prosthetic side and S3, S6 and S12 the sound side at 3, 6 and 12 months after surgery.

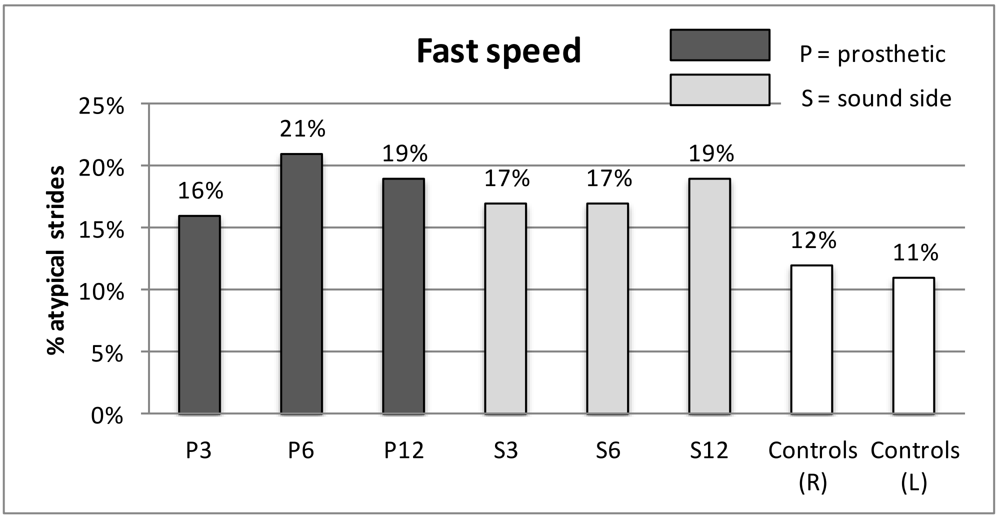

Figure 16. Percentage of atypical strides in THA patients and controls at fast speed (represented by the mean and standard error over the population). P3, P6 and P12 represents the prosthetic side and S3, S6 and S12 the sound side at 3, 6 and 12 months after surgery.

The percentage of atypical strides is greater for both the prosthetic and the contralateral side with respect to the controls, even 12 months after surgery. Notice that in the 6 month post-surgery assessment of the affected side, the values were more distant from normality, both at self-selected speed and fast speed. This happens also for the sound side, but only at self-selected speed.

For the cadence, double support, single support and the basographic cycle (HFPS), the results are presented in a boxplot that shows the smallest observation (sample minimum), lower quartile (Q1), median (Q2), upper quartile (Q3), and largest observation (sample maximum), see Figure 17. A boxplot may also indicate which observations, if any, might be considered outliers.

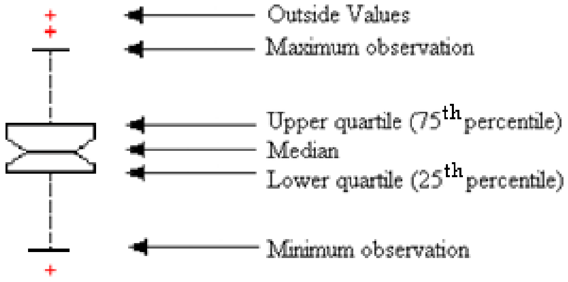

Figure 17. Values shown in the boxplot.

Regarding the cadence, shown in Figure 18, there is an expected improvement for the THA patients at self-selected speed and also at fast speed. The cadence of the control group is slightly lower but this difference is acceptable, it still within the acceptable pattern.

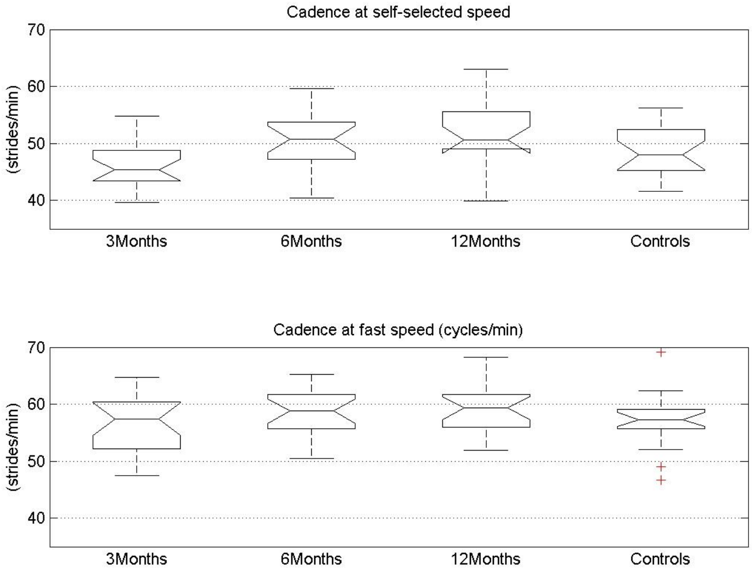

Figure 18. Cadence (in strides/min) for the prosthetic group, at 3, 6 and 12 months after surgery and for controls, in both trials.

The double support, shown in Figure 19, is decreasing at self-selected speed and also at fast speed. Besides, it can be noticed that the double support is significantly higher in patients 3 months after surgery with respect to controls, both at self-selected speed and fast speed.

A year after the intervention, in both trials there is still a huge difference between patients and control group. In particular, we highlight the fact that 12 months after the operation, 50% of cases (from Q1 to Q3) are widely dispersed. Furthermore, also the minimum and maximum are far apart (there are values too high and too low) when compared to controls which have a very small range of variation.

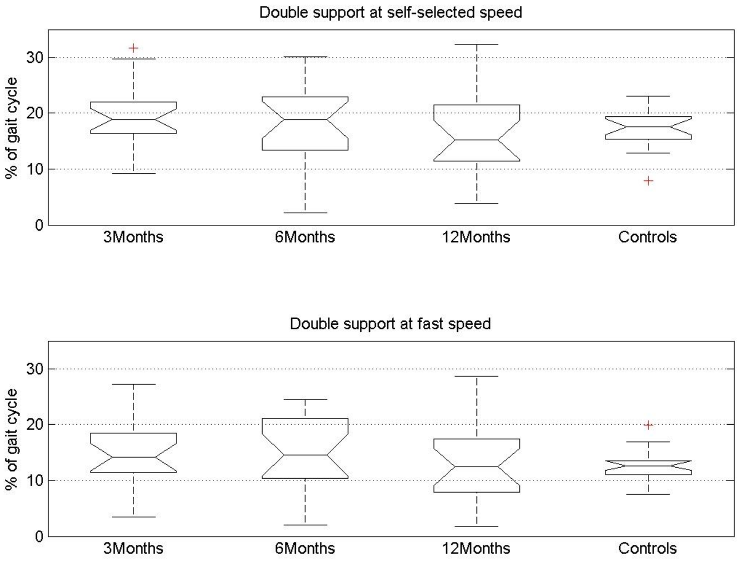

Figure 19. Double support (in percentage of GC) for the prosthetic group, at 3, 6 and 12 months after surgery and for controls, in both trials.

These results are also expected because when the cadence is low, the double support phase increases. For the operated patients, as the cadence increases, the double support becomes smaller. For controls this value should be smaller in both cases, but it is still within acceptable values.

Relatively to the single support for affected and contralateral sides, the improvement is more evident at self-selected speed than at fast speed, as shows Figure 20. In general, while the double support decreases, the single support increases.

Also in single support it is possible to observe a large range of variation, even 1 year after the intervention, when compared with controls, where the values are confined to a small range. It is important to mention that, in these plots, the right and left side were averaged for the control group.

Figure 20. Single support (SS, in percentage of GC) for the prosthetic group, at 3, 6 and 12 months after surgery and for controls, in both trials.

The sequence of the basographic cycle (HFPS) at a self-selected speed for the affected side is shown in Figure 21. In this case, the affected side of the patients is compared with the right side of the controls and the contralateral side with the left side of the controls. This was chosen because for 14 patients out of 20, the affected side is the right one.

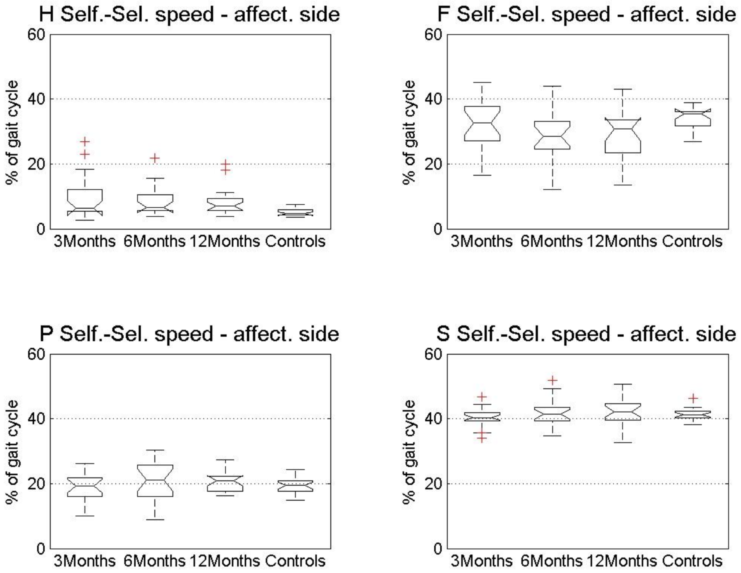

Figure 21. Basographic cycle (in percentage of GC) for the affected side, at 3, 6 and 12 months after surgery and for controls, at self-selected speed.

In the heel contact (H), we can observe a slight decrease in the operated patients over the time, nearly 5% of GC, with tendency to approach the values obtained by the control group. However, the patient’s performance is still not as good as that of controls. Besides, comparing the prosthetic and the sound sides, it is visible that this phase is significantly extended in time with respect to controls, both for self-selected speed and for fast speed. As a consequence of the prolonged heel contact (H), flat foot contact (F) is shortened for patients with respect to controls. The flat foot contact phase (F), in general, is decreasing for the patients during the follow-up. The push off phase (P) is slightly improving, up to more 20% of GC, value similar to that of controls. However, the changes of this parameter during the follow-up are very small. The swing phase (S) is also slightly increasing for the patients group, up to more 40% of GC, reaching values similar to those found for the control group.

The contralateral side, in Figure 22, at self-selected speed, follows the same behaviour as the affected side.

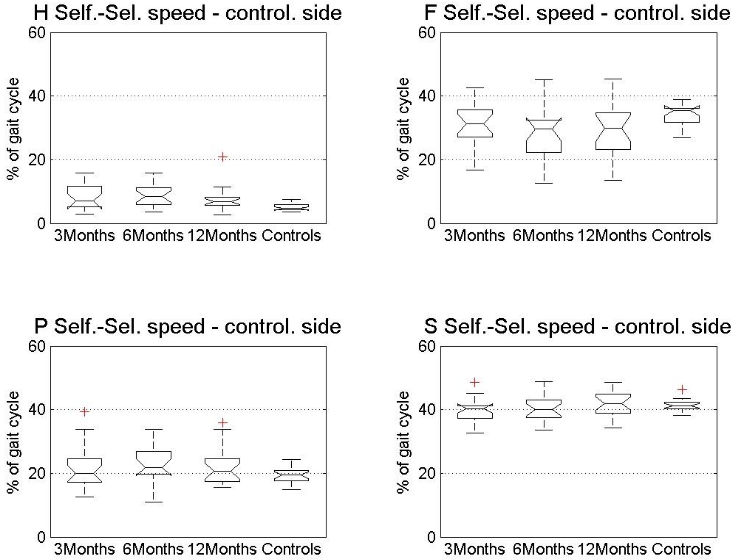

Figure 22. Basographic cycle (in percentage of GC) for the contralateral side, at 3, 6 and 12 months after surgery and for controls, at self-selected speed.

The same behaviour can be observed for the trial at fast speed, both for the affected side, in Figure 23, and for the contralateral side, in Figure 24.

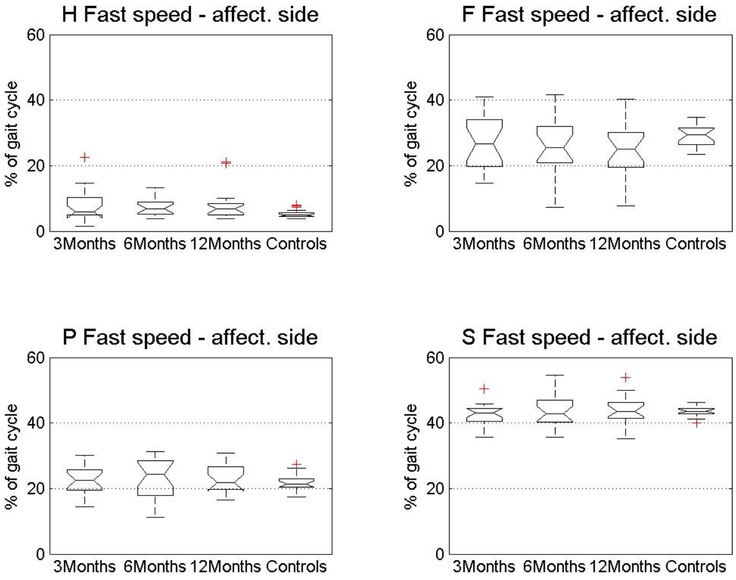

Figure 23. Basographic cycle (in percentage of GC) for the affected side, at 3, 6 and 12 months after surgery and for controls, at fast speed.

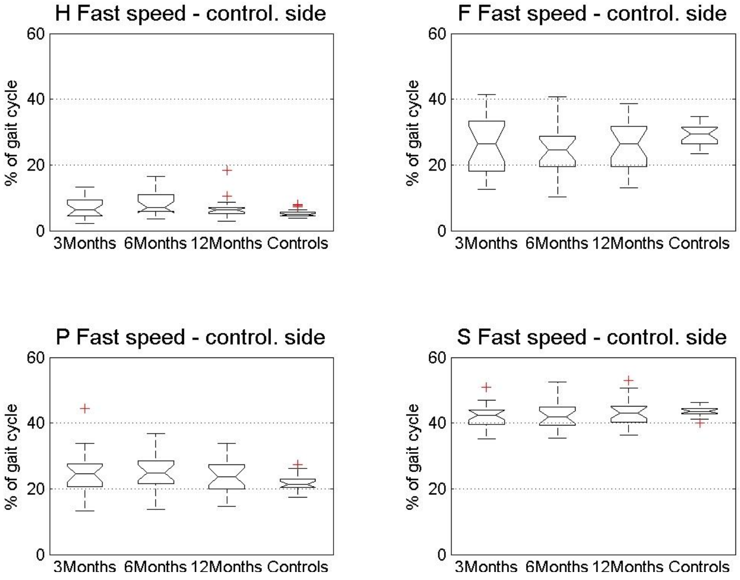

Figure 24. Basographic cycle (in percentage of GC) for the contralateral side, at 3, 6 and 12 months after surgery and for controls, at fast speed.

5.2. Goniometric results

The range of motion on the hip, in Figure 25, is improving in all the cases, tending to reach the values of the control group. Especially for the affected side, the increase of the hip ROM over the year is evident. However, even after 12 months post-operatively, it does not reach normality, in both trials. For the contralateral side, after 12 months is approximately the same for patients and for controls.

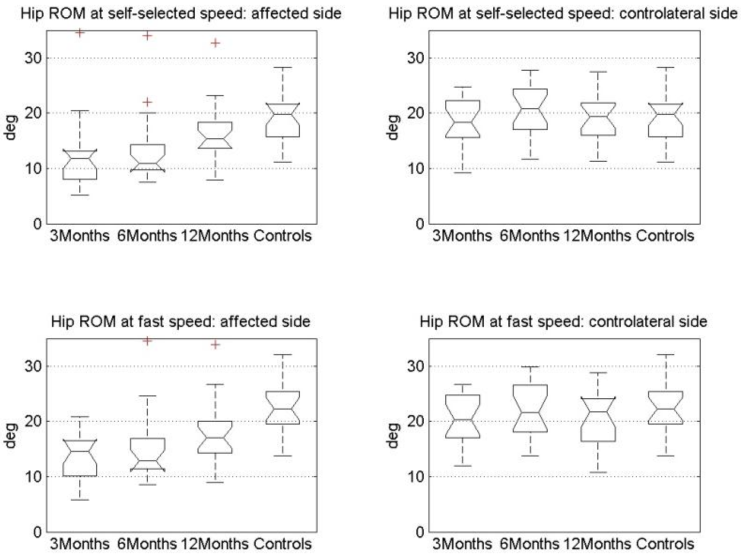

Figure 25. Range of motion of the hip (in degrees) for both sides, at 3, 6 and 12 months after surgery and for controls, in both trials.

At the knee, for the affected side, the results show that there is a significant improvement from 3 to 6 months and then, from 6 to 12 months, the results maintains constant. It is also glaring that, for the affected side, in both trials, 3 months after the intervention, the ROM is significantly smaller compared to controls.

At the same time, another phenomenon is happening. When the ROM of the knee in the affected side is increasing, in the opposite leg the ROM is decreasing, for self-selected speed and also for fast speed.

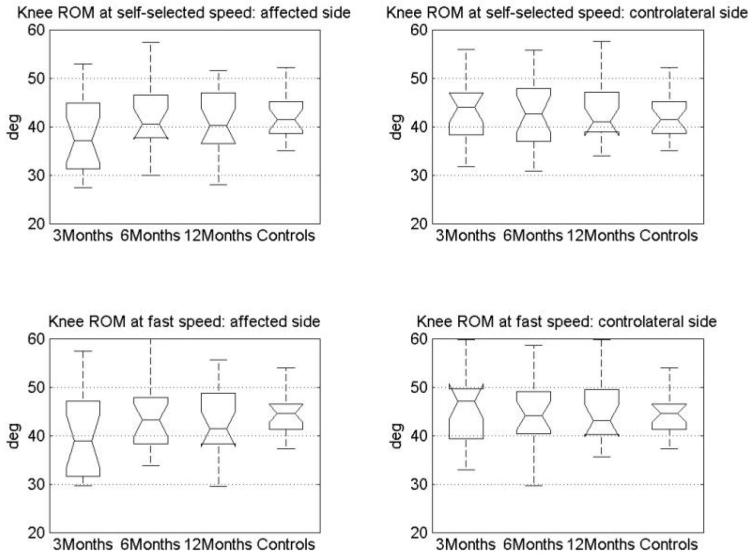

Figure 26. Range of motion of the knee (in degrees) for both sides, at 3, 6 and 12 months after surgery and for controls, in both trials.

The values resulting from the analysis of range of motion at the hip and knee are very scattered, not only 1 year after surgery, in the case of patients but also, to a minor extent, in the case of controls, in all the cases analysed.

In this analysis it is interesting to notice that, in general, the behaviour at 6 months is inconstant i.e. when the tendency is to increase over the time, at 6 months the values decrease and increase again at 12 months after the operation.

5.3. EMG results

The EMG results are presented in some graphs that show the muscle activation patterns for all the muscles analysed at self-selected speed and at fast speed. These graphs show, for each muscle, the EMG activations timing (expressed as % GC) in the different activation patterns observed, i.e. showing 1, 2, 3, 4 and 5 muscle activations during the gait cycle. The relative frequency of each activation pattern is displayed (expressed in %) in the right side of each sub-plot. Horizontal bars are grey-level coded – at each percentage of the gait cycle – according to the number of subjects in which a certain condition is observed. When the bar is filled in black it means that the entire population had the muscle contracted. On the contrary, when the bar is white it means that none of the subjects had the muscle active in that specific percentage of gait cycle. Data are reported for the prosthetic side (at 3, 6 and 12 months after the operation) and the sound side (also at 3, 6 and 12 months) of THA patients and controls (right side and left side), both at self-selected speed and fast speed.

For tibialis anterior at self-selected speed, Figure 27, in 26% of the strides performed by the THA group, in the prosthetic side, TA was activated twice along the GC, in 39% of the strides there were 3 activations, and in 34% of the strides 4 activations. Therefore, in this case, the 2 most representative activation patterns correspond to 2 and 3 activations occurring in the gait cycle. However, for the control group, at self-selected speed, the most probable activations are 3 and 4, and at fast speed are 2 and 3 activations.

Besides, in the prosthetic side we observe that the number of 2 and 3 activations is increasing during the year, and the number of 4 activations is decreasing. In the sound side, this effect is not visible, the values at 3, 6 and 12 months does not change much. Comparing with the controls, THA patients in the prosthetic side is acquiring a new strategy using the most simple activation of TA.

The same happens at fast speed, shown in Figure 28. For the prosthetic side, the number of simpler activations increases and the number of complex activations decreases over the time, also when compared with controls.

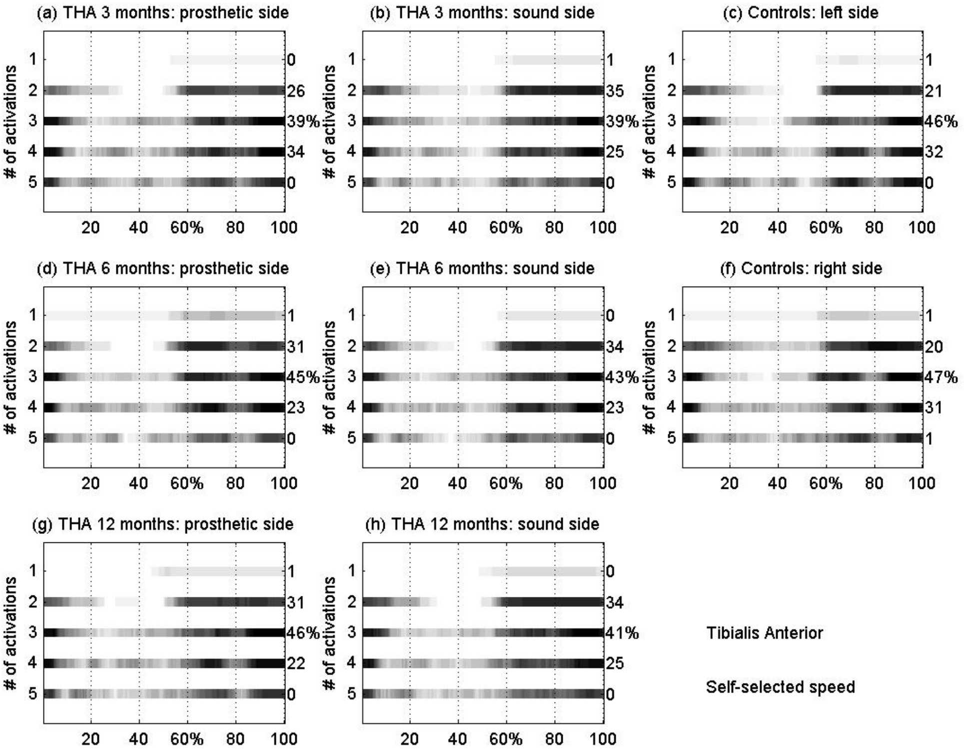

Figure 27. EMG activation timing (% GC) for tibialis anterior at self-selected speed.

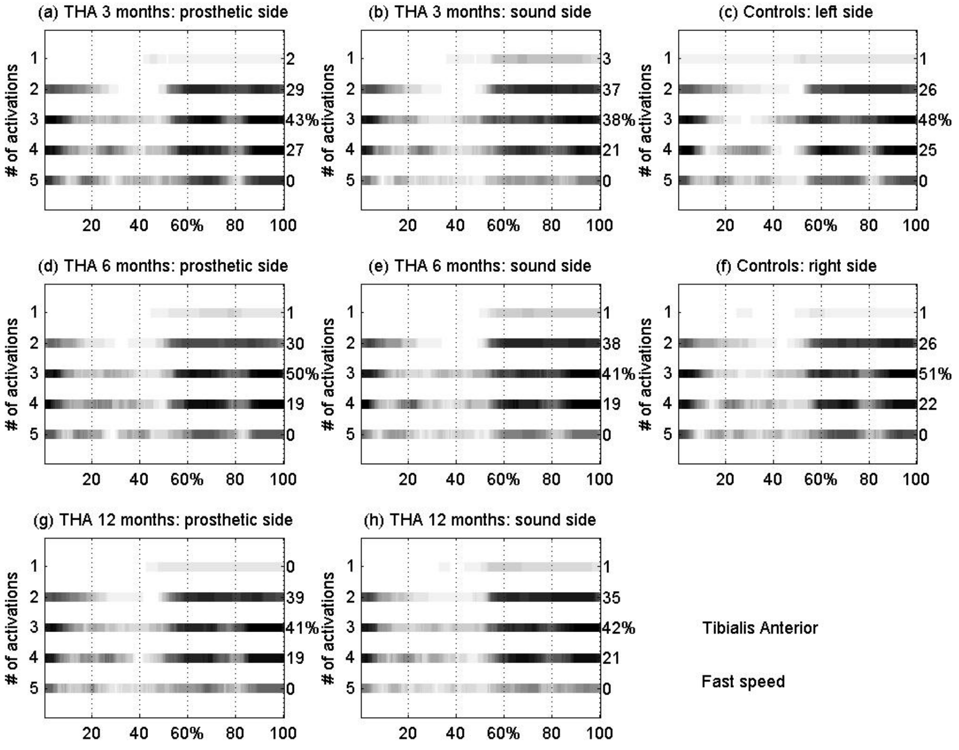

Figure 28. EMG activation timing (% GC) for tibialis anterior at fast speed.

Figure 29 shows the activations for gastrocnemius lateralis at self-selected speed and Figure 30 at fast speed. In both cases, the most representative activation patterns occurs with 1 and 2 activations, for prosthetic and sound sides and also for the control group.

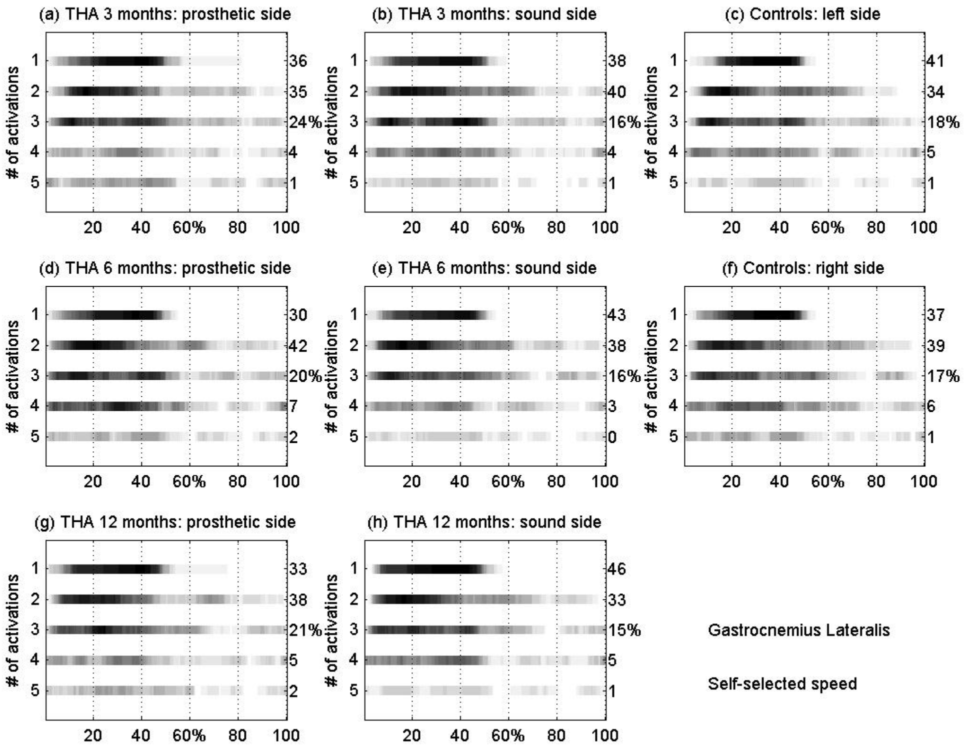

Figure 29. EMG activation timing (% GC) for gastrocnemius lateralis at self-selected speed.

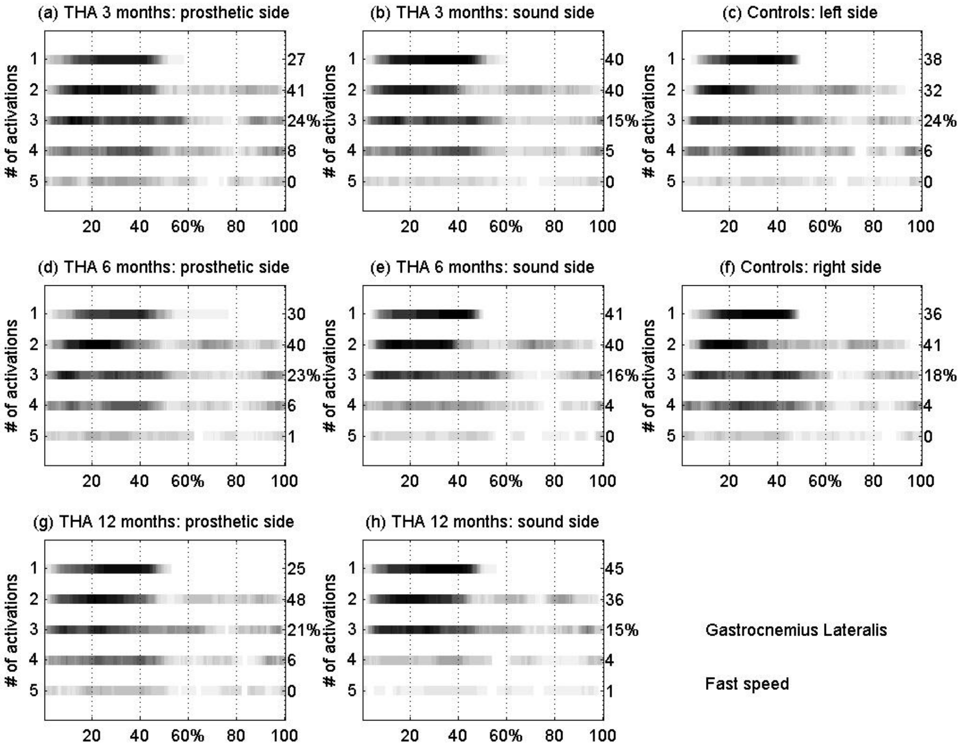

Figure 30. EMG activation timing (% GC) for gastrocnemius lateralis at fast speed.

Analysing the rectus femoris, Figure 31, when comparing the prosthetic side with the sound side and also with both sides of the controls, we can see that there are no significant differences over the time at self-selected speed. To the rectus femoris muscle, the activations more likely to occur are, in general, 2, 3 and 4 activations. In controls, at self-selected speed the most common are 2 and 3 activations and at fast speed are 3 and 4 activations. In both trials for the sound side, the major percentage of activation occoured for 3 and 4 activations; and for the prosthetic side switches between 2, 3 and 4 activations.

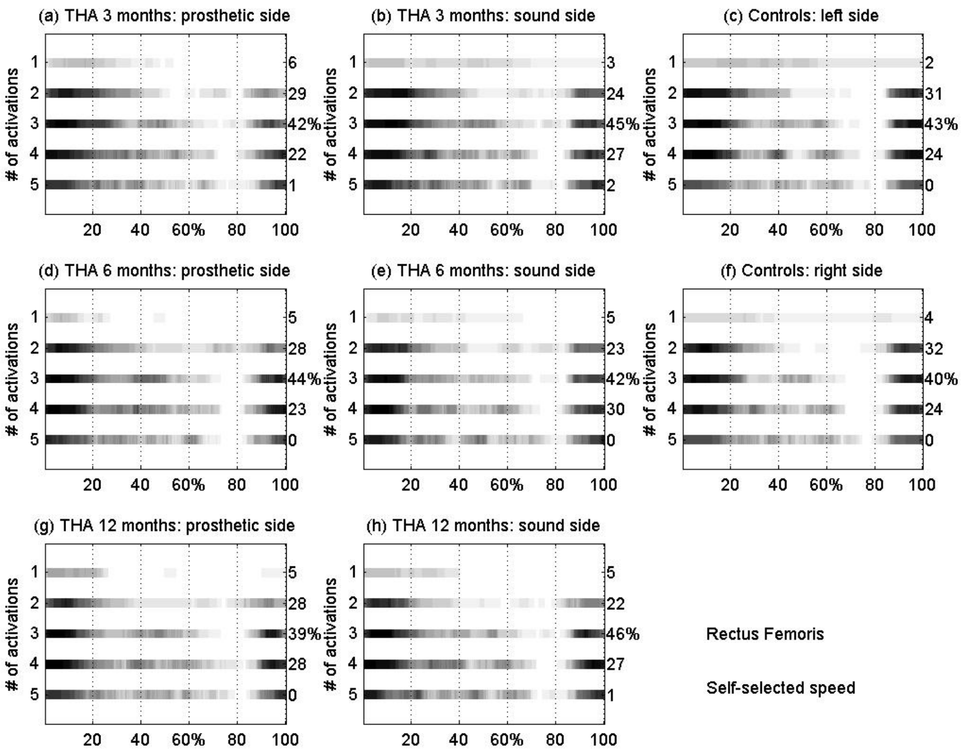

Figure 31. EMG activation timing (% GC) for rectus femoris at self-selected speed.

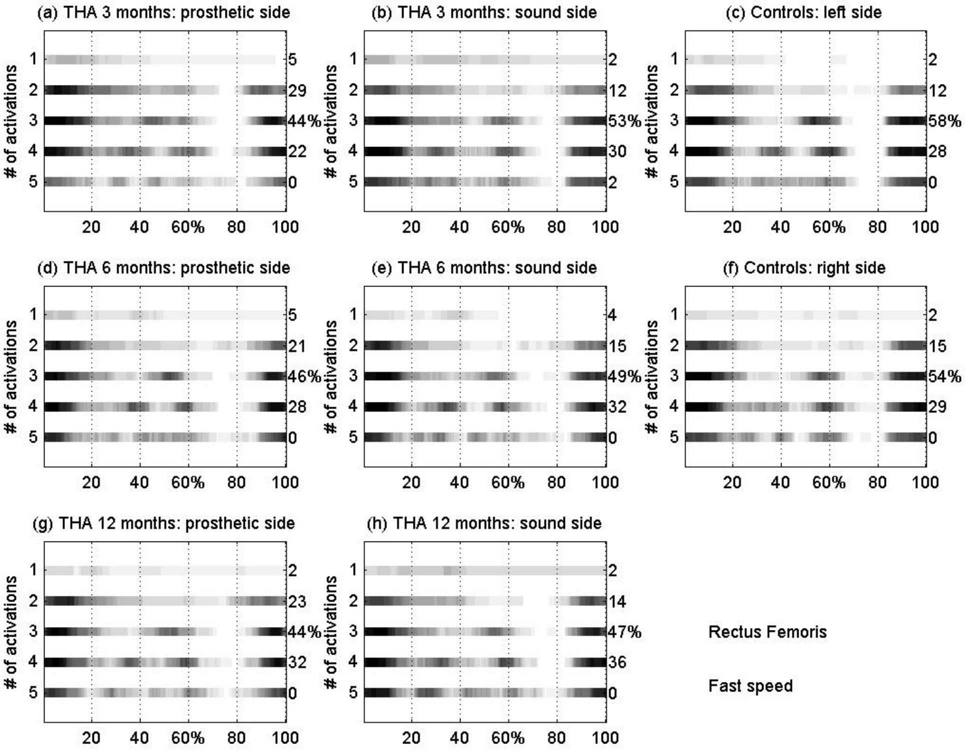

Figure 32. EMG activation timing (% GC) for rectus femoris at fast speed.

For the lateral hamstring, one year after the operation, the results are quite different when comparing the operated side with the sound side and with controls. The reduction of the third and fourth activation and the increment of the second activation over the time for the prosthetic side was also evident. Again, the acquisition of a new modality of walking involving the simpler activations, occurred.

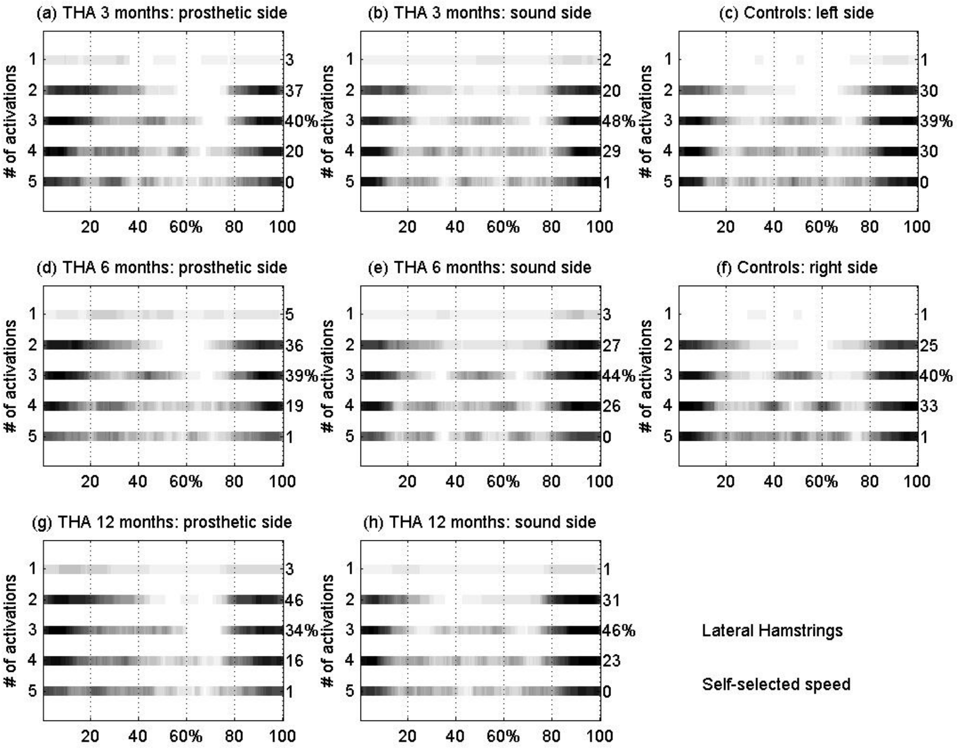

Figure 33. EMG activation timing (% GC) for lateral hamstrings at self-selected speed.

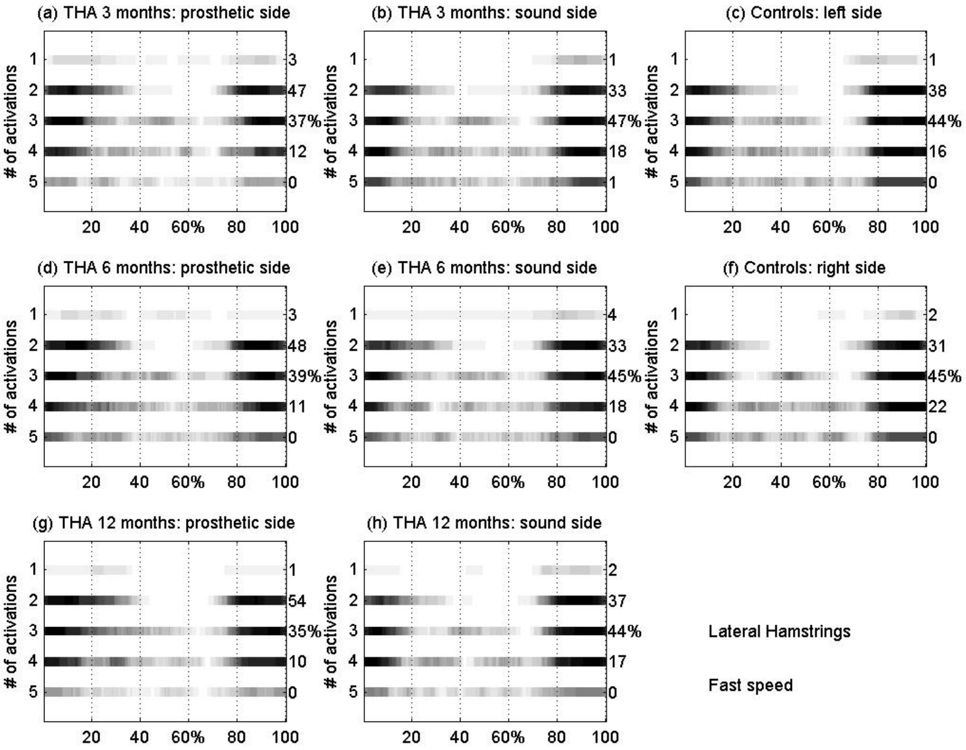

Figure 34. EMG activation timing (% GC) for lateral hamstrings at fast speed.

In the gluteus medius, Figure 34, it can be observed a slight increase of the simpler modality. The most probable activation patterns are, in all the cases, those with 2 and 3 activations, in both trials. Besides, during the follow-up, the percentage of 2 activations for the prosthetic side tends to rise and the percentage of 3 activations tends to decrease. Comparing to the controls, the percentage of activations after 1 year is quite different.

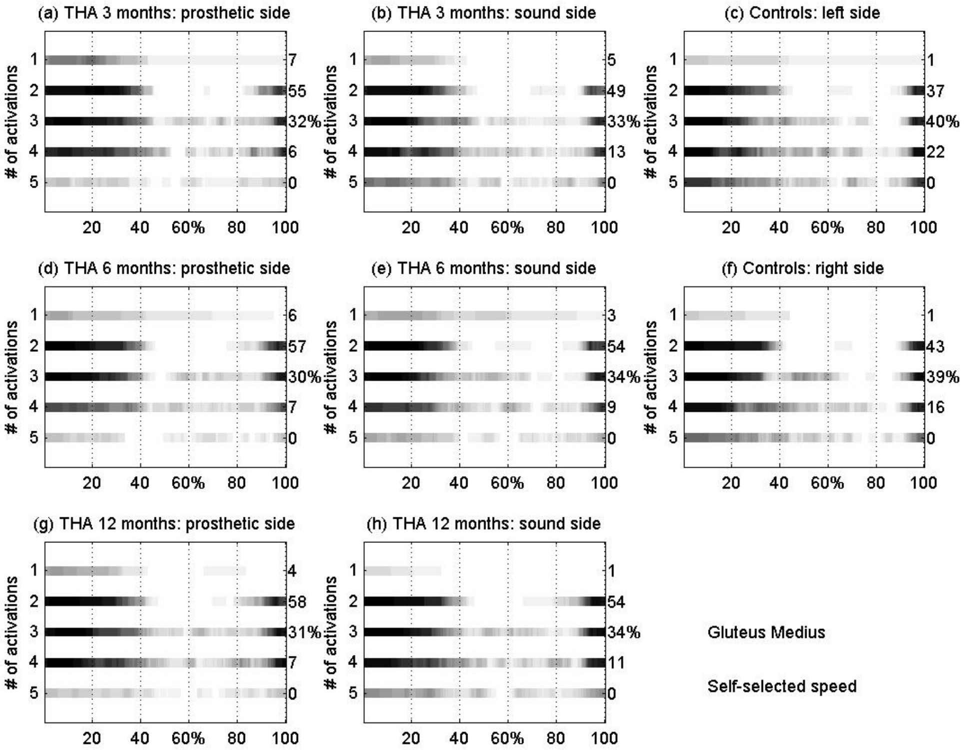

Figure 35. EMG activation timing (% GC) for gluteus medius at self-selected speed.

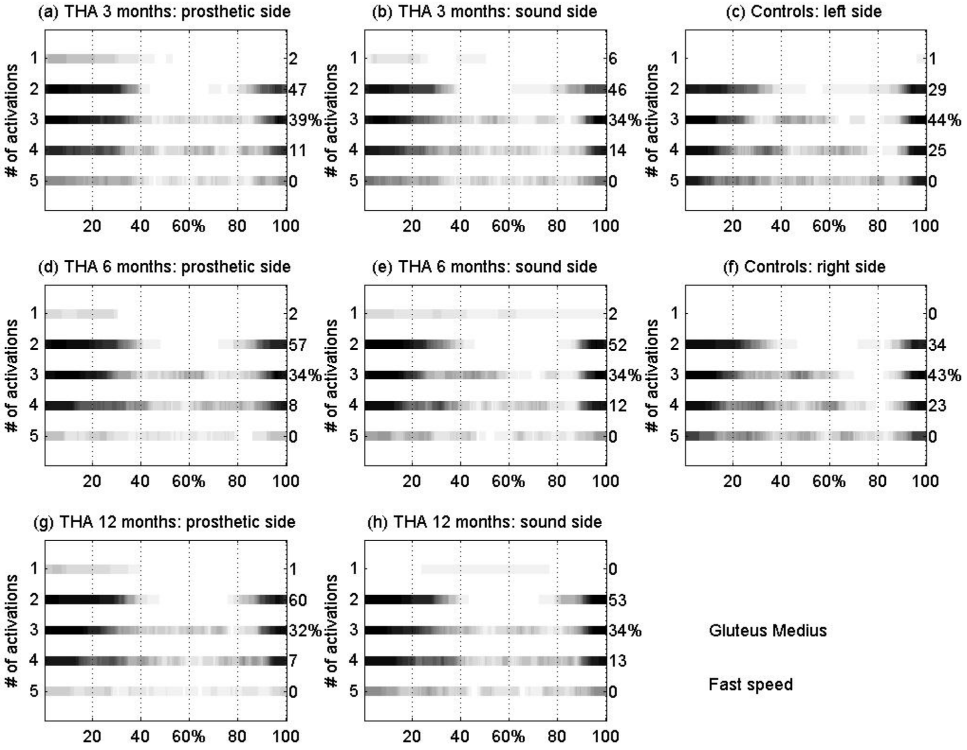

Figure 36. EMG activation timing (% GC) for gluteus medius at fast speed.

Chapter 6. Discussion of results

6.1. Basographic and gait cycle parameters

After analysing the results of this study, it is possible to say that, THA patients walk with more atypical strides than controls. The number of atypical cycles does not improve during follow-up and involves both legs. Besides, in all the cases, except for the sound side at 6 months and at fast speed, the percentage of atypical cycles is much greater at 6 months after surgery. This study did not allow understanding exactly the causes of these two events but, to explain this occurrence, two possible causes were found:

- Due to the leg length discrepancy after surgery (approximately 6mm, on average);
- Due to a diminished proprioception after THA as a consequence of loss of the joint capsule and capsule ligaments, and a partial loss of extra-capsular mechanoreceptors, such as stretch receptors in the adjacent tendons and muscles, that is involved in proprioception and in joint-position sense.

Velocity and cadence are improving during the follow-up for THA patients. Furthermore, one year post-operatively, THA patients walk with the same velocity as controls but with a slightly higher cadence. This work did not allow concluding about this event but it might be due to a smaller step length.

As literature reports, double support represents 20% GC and single support represents nearly 40% GC (in each leg). The double support in both trials is decreasing, up to 15%; and the single support is improving, reaching more than 40% GC.

The most glaring result concerning the gait phases happens with the heel contact (H). This phase is substantially prolonged in THA patients with respect to controls, also in the contralateral side, for all the trials. This phenomenon supports the hypothesis that loading response is a critical gait phase even 1 year after surgery. As a consequence of the extended heel contact (H), flat foot contact (F) is shortened for patients with respect to controls. In the remaining phases of the gait cycle, no considerable differences were detected. Nevertheless, even one year after surgery, patient’s performance is still not as good as that of controls.

Interestingly, some of the analysed gait parameters move away from normality in correspondence of the 6-month assessment, i.e. are worse in the 6-month assessment than in the 3-month assessment and then improve again 12 months post-surgery. In particular this behaviour is observed for: single support of the affected side and contralateral side at fast speed, flat foot contact (F) of the affected side at self-selected speed, heel contact (H), flat foot contact (F) and push off (P) of the sound side at self-selected speed, flat foot contact (F), push off (P) and swing (S) of the affected side at fast speed, flat foot contact (F) of the sound side at fast speed, percentage of atypical strides on both sides at self-selected speed and percentage of atypical strides on the affected side at fast speed. A possible explanation for this finding is that patients reorganize their walking strategy and establish possible compensative mechanisms around six months post-operatively.

6.2. Goniometric and range of motion

The sagittal-plane range of motion of the operated side improves considerably one year after surgery, both at the hip and knee, but it does not reach normality. Another interesting discovery was that, in all the trials, when the ROM in the affected side is increasing, in the opposite leg the ROM is decreasing. One of the possible explanations for this event is a possible compensation strategy of the non-affected limb possibly to improve the gait symmetry.

Furthermore, the analysis of the knee ROM shows that not only the hips of the surgical group were affected but also the knees, as showed by Loizeau *et al.* (Loizeau J, 1995).

6.3. EMG and muscle activations

One of the most notable findings concerning the EMG results is that, in general, the number of simpler activations increases and the number of complex activations decreases over the time for THA patients. Hence, the most frequent activation pattern 12 months after surgery for the prosthetic side of the patients is a 3-activation pattern for tibialis anterior and rectus femoris and 2-activation pattern for gastrocnemius lateralis, lateral hamstrings and gluteus medius. A possible explanation for this fact is that THA patients adopt a simplified muscle control strategy with respect to controls, as patterns with a small number of activations are favoured to the detriment of those with a high number of activations. This behaviour is more remarkable in lateral hamstrings and gluteus medius of the affected side and becomes more evident during the follow-up. It can also be observed in the contralateral side, and it is even more glaring for gluteus medius, possibly indicating an arising compensative strategy of the unaffected side aimed at improving gait symmetry. Also here, it can be hypothesized that these simplified motor control strategies are related to the proprioceptive loss consequent to the hip replacement and to the search of an effective walking scheme.

Likewise, tibialis anterior presents the same behaviour in both sides of patients, with a abnormal muscle activation timing, while gastrocnemius lateralis and rectus femoris seems to be slightly affected by the prosthesis.

The fast speed trial substantially confirms the results obtained with gait at the self-selected speed. However, walking at a higher cadence is a more demanding task for patients. As a consequence, in some cases the significance of the differences between patients and controls increase.

Chapter 7. Conclusions and future research

This analysis supports the conclusion that patients who underwent a THA, one year after the intervention continue showing gait abnormalities relative to controls. The most remarkable abnormalities found were:

- The percentage of atypical cycles, which does not improve during follow-up and involves both legs;
- Prolonged heel contact (H), supporting the theory that loading response is a critical gait phase even 1 year after surgery;
- The hip and knee ROM that is improving on the affected side (even if it does not reach normality) and becoming worst on the healthy side;
- The analysis of the knee ROM proves that not only the hips of the surgical group were affected but also the knees.
- Six months after surgery is the period in which the results are more distant from the normal pattern, in almost all the analysed parameters, because patients reorganize their walking strategy and establish possible compensative mechanisms;
- Patients adopt a simplified muscle control strategy, as patterns with a small number of activations are favoured to the detriment of those with a high number of activations.

In this work, we are faced with the fact those six months after surgery is the most critical time, where the results are worst, leading us to conclude that the rehabilitation protocols should not only focus on the first few months after surgery, but prosecute in a long-term effort to normalize gait by muscle strengthening and motor relearning.

In summary, statistical gait analysis allowed to evidence subtle differences in the muscular activation timing of THA patients with respect to controls. Despite improvements in the hip kinematics, the muscular engagement of patients remains higher with respect to controls and does not substantially change along the year after the hip implant, as conclude Foucher et al (Foucher K, 2007). This problem is very worrying because, according to Beaulues *et al.* (Beaulieu M, 2010), when the gait parameters does not reach the normality and the patients develop gait adaptations, they race the risk of developing other joint diseases.

It is important to refer that some of these findings were never related in literature before and that a paper describing them have been submitted to a peer-reviewed journal for publication.

With continuing advancement in biomechanics and information processing, it is expected that gait analysis system will become more productive, affordable, and important in hip arthroplasty in the near future. Further investigation is needed to confirm the reasons why THA patients' gait mechanics do not return to normal following surgery to develop better surgical techniques and/or rehabilitation programs.

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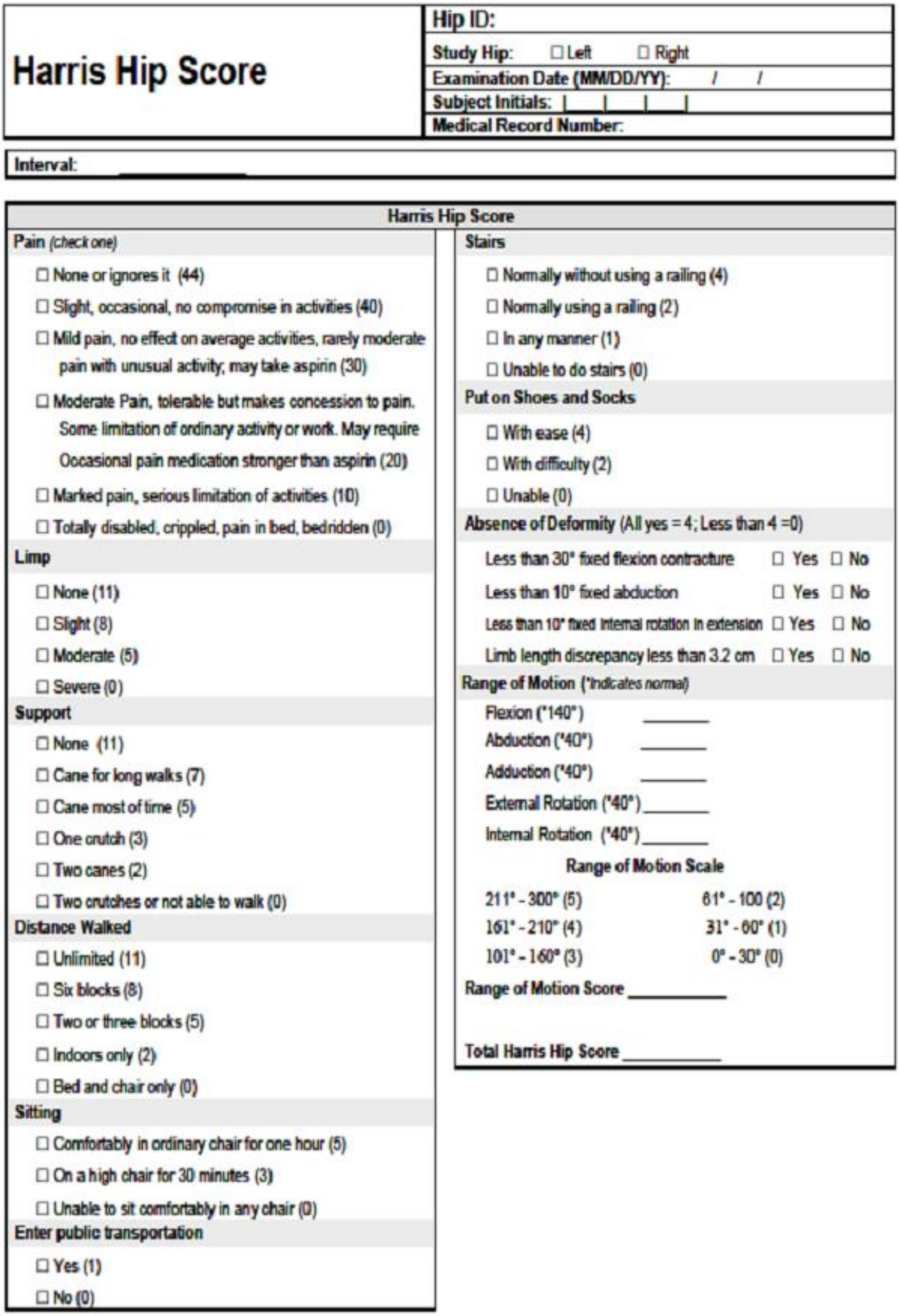
Appendix A



Harris Hip Score

Hip ID:
Study Hip: ☐ Left ☐ Right
Examination Date (MM/DD/YY): / /
Subject Initials: |__|__|__|
Medical Record Number:

Interval: ______________

Harris Hip Score

Pain *(check one)*

- ☐ None or ignores it (44)
- ☐ Slight, occasional, no compromise in activities (40)
- ☐ Mild pain, no effect on average activities, rarely moderate pain with unusual activity; may take aspirin (30)
- ☐ Moderate Pain, tolerable but makes concession to pain. Some limitation of ordinary activity or work. May require Occasional pain medication stronger than aspirin (20)
- ☐ Marked pain, serious limitation of activities (10)
- ☐ Totally disabled, crippled, pain in bed, bedridden (0)

Limp

- ☐ None (11)
- ☐ Slight (8)
- ☐ Moderate (5)
- ☐ Severe (0)

Support

- ☐ None (11)
- ☐ Cane for long walks (7)
- ☐ Cane most of time (5)
- ☐ One crutch (3)
- ☐ Two canes (2)
- ☐ Two crutches or not able to walk (0)

Distance Walked

- ☐ Unlimited (11)
- ☐ Six blocks (8)
- ☐ Two or three blocks (5)
- ☐ Indoors only (2)
- ☐ Bed and chair only (0)

Sitting

- ☐ Comfortably in ordinary chair for one hour (5)
- ☐ On a high chair for 30 minutes (3)
- ☐ Unable to sit comfortably in any chair (0)

Enter public transportation

- ☐ Yes (1)
- ☐ No (0)

Stairs

- ☐ Normally without using a railing (4)
- ☐ Normally using a railing (2)
- ☐ In any manner (1)
- ☐ Unable to do stairs (0)

Put on Shoes and Socks

- ☐ With ease (4)
- ☐ With difficulty (2)
- ☐ Unable (0)

Absence of Deformity (All yes = 4; Less than 4 = 0)

Less than 30° fixed flexion contracture ☐ Yes ☐ No
Less than 10° fixed abduction ☐ Yes ☐ No
Less than 10° fixed internal rotation in extension ☐ Yes ☐ No
Limb length discrepancy less than 3.2 cm ☐ Yes ☐ No

Range of Motion (**indicates normal*)

Flexion (*140°) _______
Abduction (*40°) _______
Adduction (*40°) _______
External Rotation (*40°) _______
Internal Rotation (*40°) _______

Range of Motion Scale

211° - 300° (5)
161° - 210° (4)
101° - 160° (3)
61° - 100 (2)
31° - 60° (1)
0° - 30° (0)

Range of Motion Score __________

Total Harris Hip Score __________

Figure A1. Harris Hip Score exam.

Appendix B

Table B1. Antropometric Data

	ID	**Age**	**Height**	**Weight**	**BMI**	**Gender**
Patients	35	70	150	70	31,1	F
	32	61	165	57	20,9	F
	36	59	183	92	27,5	M
	47	69	167	72	25,8	M
	51	67	150	72	32,0	F
	62	55	177	75	23,9	M
	66	69	165	75	27,5	M
	60	69	175	88	28,7	M
	71	65	180	90	27,8	M
	111	70	157	85	34,5	F
	115	57	185	90	26,3	M
	117	74	172	93	31,4	F
	118	65	179	80	25,0	M
	123	73	162	64	24,4	F
	124	79	165	60	22,0	M
	133	71	170	85	29,4	F
	134	49	164	59	21,9	F
	137	68	155	56	23,3	F
	146	71	173	76	25,4	F
	149	61	179	100	31,2	F
Controls	205	74	159	65	25,7	F
	202	64	166	73	26,5	M
	209	57	193	96	25,8	M
	210	58	160	52	20,3	F
	211	74	175	70	22,9	M
	213	62	173	69	23,1	M
	212	62	160	60	23,4	F
	215	62	180	72	22,2	M
	217	70	155	51	21,2	F
	220	59	170	63	21,8	F
	223	66	160	62	24,2	F
	224	68	171	60	20,5	M
	229	69	168	54	19,1	F
	225	69	182	75	22,6	M
	228	70	162	69	26,3	F
	231	64	168	67	23,7	F
	230	64	175	75	24,5	M
	235	68	180	94	29,0	M
	234	59	172	85	28,7	M
	239	69	167	68	24,4	M

Table B2. Leg length discrepancy and Harris Hip Score results for patients.

	ID	**Leg length discrepancy**	**Harris Hip Score**		
			3 mesi	**6 mesi**	**12mesi**
Patients	35	1,5	81	86	99
	32	0,75	89	100	100
	36	1	86	100	100
	47	1	77	91	97
	51	1,5	86	99	96
	62	0	96	100	100
	66	0,5	82	94	94
	60	0	94	100	100
	71	0	91	98	100
	111	1	88	98	98
	115	0	100	100	100
	117	0,5	88	96	100
	118	0	100	100	100
	123	0,5	94	97	97
	124	0	97	98	100
	133	0,5	73	83	89
	134	0,5	100	100	100
	137	0	87	92	98
	146	1	100	100	100
	149	1,5	90	100	100

Appendix C

Table C1. Numerical EMG results obtained at 3 months after surgery, for one muscle (Rectus Femoris) in the prosthetic side for the THA patients. The table is segmented due to their long length.

GENERAL DATA					**One activation**				
	ID_step	**Name**	**Type of cycle**	**Side (contro)**	**start**	**sd**	**end**	**sd**	**N°steps**
1	35	Rimerici M	HFPS	L	20,5	32,7	52,2	38,3	9
2	32	Gaviglio L	HFPS	R	26,0	12,0	30,2	13,9	3
3	36	Vironda G	HFPS	L	0,0	0,0	0,0	0,0	
4	47	Franceschi F	HFPS	L	0,0	0,0	0,0	0,0	
5	51	Giannese A	HFPS	R	0,0	0,0	17,1	0,0	1
6	62	Trione L	HFPS	L	0,0	0,0	0,0	0,0	
7	66	Chiartano P	HFPS	L	0,0	0,0	0,0	0,0	
8	60	Anrò M	HFPS	L	50,9	47,0	61,9	52,2	5
9	71	Vaira M	HFPS	R	0,0	0,0	0,0	0,0	
10	111	Ricca M	HFPS	R	0,0	0,0	19,6	16,9	2
11	115	Quaccia M	HFPS	L	0,0	0,0	0,0	0,0	
12	117	Nicola B	HFPS	L	0,0	0,0	69,4	53,1	1
13	118	Piran R	HFPS	R	0,0	0,0	0,0	0,0	
14	123	Gallo F	HFPS	L	0,0	0,0	0,0	0,0	
15	124	Vigna D	HFPS	L	0,0	0,0	100,0	0,0	1
16	133	Conta M	HFPS	L	43,2	27,9	49,6	25,1	6
17	134	Actis C E	HFPS	R	0,0	0,0	0,0	0,0	
18	137	Bernardi L	HFPS	L	0,0	0,0	0,0	0,0	
19	146	Maga ST	HFPS	L	2,2	2,5	16,7	8,2	7
20	149	Mesnil MD	HFPS	L	0,0	0,0	0,0	0,0	

Table C1. (continued).

Two activations								
first activation				**second activation**				**N°steps**
0,7	1,5	41,8	19,6	72,0	29,5	78,8	30,4	26
4,2	8,6	22,7	16,5	59,6	30,8	67,7	33,9	12
0,0	0,0	47,0	4,6	87,0	3,0	100,0	0,0	6
0,0	0,0	71,6	1,9	87,8	2,0	100,0	0,0	19
0,0	0,0	21,4	6,0	88,8	3,3	100,0	0,0	33
0,0	0,0	34,4	25,8	84,1	1,0	100,0	0,0	3
0,0	0,0	30,2	15,3	85,7	1,8	100,0	0,0	10
0,0	0,0	21,2	15,1	89,1	3,7	99,8	0,8	10
0,0	0,0	40,3	9,7	83,6	20,8	92,3	21,8	8
0,8	2,1	18,3	6,2	91,6	16,6	93,8	16,0	17
0,0	0,0	20,1	6,7	86,1	2,1	100,0	0,0	35
0,0	0,0	43,3	23,2	78,1	24,0	92,2	22,3	17
0,0	0,0	29,5	5,1	87,9	2,0	100,0	0,0	7
0,0	0,0	33,0	32,9	92,4	8,6	100,0	0,0	3
0,0	0,0	25,2	8,9	79,9	10,7	100,0	0,0	4
11,9	11,4	39,5	7,8	64,5	5,9	70,2	5,1	35
0,0	0,0	18,5	12,4	87,6	3,4	100,0	0,0	33
0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	
1,4	1,8	26,3	8,1	72,7	30,2	75,1	30,3	15
0,0	0,0	47,9	3,0	86,3	1,7	100,0	0,0	7

Table C1. (continued).

Three activations												
first activation				second activation				third activation				N°steps
0,4	1,1	40,3	16,7	49,5	22,9	57,6	18,6	90,5	15,4	96,9	12,6	16
1,5	5,8	28,1	17,3	49,6	26,4	56,1	26,6	73,2	27,4	83,6	28,2	31
0,0	0,0	45,7	4,2	61,0	5,7	64,7	5,4	86,8	3,0	100,0	0,0	25
0,0	0,0	46,9	10,3	51,3	10,9	72,2	2,6	88,1	2,3	100,0	0,0	25
0,0	0,0	21,1	7,2	47,4	22,5	52,5	23,4	88,9	2,1	100,0	0,0	11
0,0	0,0	18,8	5,2	43,3	5,5	55,1	5,1	83,8	1,7	100,0	0,0	23
0,0	0,0	24,2	4,8	40,7	8,0	49,8	7,9	86,5	2,1	100,0	0,0	39
0,0	0,0	17,3	11,8	48,1	20,1	54,7	21,6	86,5	12,4	98,6	7,1	27
0,1	0,3	42,7	8,5	53,4	9,9	61,2	10,8	86,3	4,2	100,0	0,0	41
0,2	0,7	18,0	9,1	39,5	21,6	45,6	20,8	94,3	12,6	96,9	11,8	29
0,0	0,0	20,1	6,4	33,4	10,6	38,3	10,5	85,9	2,5	100,0	0,0	23
0,1	0,3	32,9	13,4	43,8	16,9	59,2	20,5	84,2	10,6	96,7	9,7	28
0,0	0,0	29,3	8,1	48,1	9,9	55,4	9,8	87,7	1,5	100,0	0,0	23
0,0	0,3	28,5	13,0	56,6	7,9	66,3	7,6	87,9	6,5	100,0	0,0	42
0,0	0,0	19,2	6,6	45,3	11,9	51,6	12,2	85,4	3,3	100,0	0,0	52
9,1	7,3	25,3	12,9	34,9	18,5	48,9	13,2	74,2	14,4	78,4	12,6	30
0,0	0,0	14,2	8,7	28,9	17,9	35,7	17,9	87,2	5,0	100,0	0,0	30
0,3	0,8	33,3	10,8	41,5	15,1	54,5	14,1	75,0	20,0	82,0	17,3	15
1,0	2,2	30,9	12,5	42,0	19,2	48,1	20,4	83,7	21,4	91,5	20,7	13
0,0	0,0	49,4	3,5	64,3	3,5	71,1	4,5	85,8	1,8	100,0	0,0	43

Table C1. (continued).

Four activations																
first activation				second activation				third activation				fourth activation				N°steps
0,0	0,0	30,5	15,1	35,2	15,5	48,0	11,2	61,0	20,3	65,3	18,5	90,6	14,9	94,7	14,9	12
0,5	2,2	27,7	14,3	44,5	22,3	53,2	20,8	66,6	20,6	72,1	21,0	91,7	7,1	100,0	0,0	18
0,0	0,0	44,6	5,0	50,9	7,2	55,1	5,9	63,3	8,9	67,0	7,9	86,4	3,3	100,0	0,0	18
0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	
0,0	0,0	17,8	7,3	25,2	7,8	33,1	8,7	61,1	7,9	66,2	8,7	87,3	2,3	100,0	0,0	10
0,0	0,0	18,9	6,6	40,9	4,9	48,2	5,3	54,2	3,7	59,9	2,8	83,5	2,1	100,0	0,0	23
0,0	0,0	20,8	5,4	32,2	7,2	38,6	7,2	46,2	7,8	55,6	6,6	85,9	2,6	100,0	0,0	35
0,0	0,0	23,2	14,2	40,9	16,1	47,1	16,4	60,6	15,4	65,6	15,1	83,4	15,2	100,0	0,0	15
0,0	0,0	40,5	10,1	46,0	11,3	51,5	11,3	56,7	11,6	64,1	10,5	82,6	9,3	97,5	11,7	22
0,4	1,1	15,3	7,7	22,9	10,9	30,1	9,5	51,3	18,9	57,4	18,1	96,3	3,8	100,0	0,0	25
0,0	0,0	17,6	5,4	25,6	9,9	32,0	8,8	46,6	18,1	52,6	17,0	86,1	3,6	100,0	0,0	12
0,1	0,4	20,3	11,2	25,3	12,4	35,0	12,1	51,9	15,4	58,2	16,8	91,7	7,4	99,4	2,6	21
0,0	0,0	31,1	8,8	43,6	8,0	47,8	7,3	58,2	4,1	63,2	4,8	87,5	1,6	100,0	0,0	13
0,0	0,0	21,8	13,2	27,0	12,8	34,0	11,0	57,9	2,2	67,9	1,6	87,6	6,3	100,0	0,0	17
0,0	0,0	14,7	4,6	23,9	9,3	30,4	9,1	50,7	4,6	56,4	4,4	85,5	2,6	100,0	0,0	21
3,1	3,2	16,1	15,7	24,8	20,1	42,9	18,4	56,9	23,1	65,7	19,7	85,7	17,3	88,6	15,0	6
0,0	0,0	12,2	6,1	20,7	7,7	27,6	7,3	40,3	17,2	45,3	17,3	87,4	4,8	100,0	0,0	16
1,6	4,9	32,2	10,8	37,4	11,0	46,6	9,9	61,4	13,2	68,9	13,2	92,4	11,8	94,8	10,9	20
0,3	0,9	29,0	7,7	37,0	12,1	42,2	12,0	64,9	16,4	69,0	17,6	93,4	8,5	97,8	7,9	19
0,0	0,0	45,1	3,2	52,1	7,5	56,3	6,0	66,2	6,2	72,7	5,7	85,7	2,8	100,0	0,0	12

Table C1. (continued).

Five activations																				
first activation				**second activation**				**third activation**				**fourth activation**				**fifth activation**				**N°steps**
0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	
0,6	1,4	12,6	16,4	22,7	23,8	29,3	25,4	33,9	26,6	40,3	25,3	57,3	27,0	63,4	24,2	89,9	9,3	99,4	1,3	5
0,0	0,0	41,5	1,9	45,6	3,3	52,9	2,6	60,5	5,7	63,4	4,2	77,7	11,0	80,4	11,2	87,9	2,6	100,0	0,0	3
0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	0,0	
0,0	0,0	11,5	0,0	13,3	0,0	23,5	0,0	34,1	0,0	42,9	0,0	64,5	0,0	78,4	0,0	84,9	0,0	100,0	0,0	1
0,0	0,0	16,0	4,2	27,5	12,6	32,4	13,3	44,0	6,4	51,7	4,6	56,6	3,6	60,5	3,5	83,8	1,8	100,0	0,0	5
0,0	0,0	22,8	4,2	31,8	5,4	36,0	5,4	41,7	4,2	46,3	5,0	54,8	12,7	60,3	11,6	87,9	2,7	100,0	0,0	11
0,0	0,0	16,7	17,5	26,7	8,9	31,9	13,2	49,0	13,6	51,2	14,1	61,2	2,5	69,3	1,8	88,4	5,1	100,0	0,0	2
0,0	0,0	43,4	0,0	45,8	0,0	49,0	0,0	55,9	0,0	59,9	0,0	62,3	0,0	69,9	0,0	82,8	0,0	100,0	0,0	1
0,0	0,0	20,0	9,3	24,9	12,4	31,3	8,3	43,3	9,4	47,3	10,3	64,9	17,8	70,0	18,8	94,9	4,4	100,0	0,0	3
0,0	0,0	16,5	0,0	18,9	0,0	32,3	0,0	38,8	0,0	43,3	0,0	47,0	0,0	53,9	0,0	83,6	0,0	100,0	0,0	1
0,3	0,9	11,8	8,8	16,5	9,7	26,0	9,1	37,9	17,1	42,2	17,5	64,6	20,7	68,9	21,1	95,0	5,4	99,8	0,7	14
0,0	0,0	23,1	7,8	28,5	11,2	33,9	7,5	48,1	6,1	50,7	6,4	58,5	4,0	62,6	3,5	88,2	1,7	100,0	0,0	4
0,0	0,0	19,6	12,7	24,9	11,5	29,2	8,6	36,6	12,4	38,8	12,5	57,2	1,7	66,1	1,2	89,3	7,3	100,0	0,0	3
0,0	0,0	14,5	3,0	21,4	1,8	27,4	6,0	41,0	13,4	48,2	14,6	62,2	16,1	67,2	14,0	84,6	1,9	100,0	0,0	3
3,6	5,1	11,8	0,6	18,6	3,8	21,7	3,5	27,4	0,8	37,7	5,3	52,2	21,4	56,9	20,0	84,7	20,4	88,0	17,0	2
0,0	0,0	9,3	0,7	15,0	0,3	22,6	1,5	30,5	5,7	33,5	5,3	47,5	21,2	51,2	21,1	84,9	1,7	100,0	0,0	2
2,0	3,1	23,5	7,3	26,8	7,4	35,0	8,4	40,0	10,6	49,1	9,7	62,7	13,5	69,2	13,7	94,1	9,0	97,3	7,9	23
0,1	0,4	24,2	11,9	27,8	12,3	36,0	9,1	42,5	12,6	48,2	13,9	64,7	17,1	68,3	17,1	94,9	3,7	100,0	0,0	12
0,0	0,0	47,8	0,0	50,2	0,0	53,3	0,0	61,3	0,0	65,1	0,0	66,8	0,0	69,9	0,0	86,1	0,0	100,0	0,0	1

Note: It is important to refer that this is only one demonstrative case. The author is able to provide the remaining data, if asked.

Appendix D

Table D1- Typical and atypical strides at 3 months post-operative.

			Normal Gait				Fast Gait			
N	**ID**	**Side**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**
1	35	R	113	4	112	5	100	7	103	4
2	32	L	130	9	131	7	162	13	161	14
3	36	R	91	11	90	12	87	9	82	16
4	47	D	85	14	75	23	82	12	78	20
5	51	L	100	9	78	30	69	12	64	19
6	62	R	101	69	140	19	67	91	128	128
7	66	R	139	18	74	83	140	23	141	22
8	60	R	96	37	112	16	104	29	115	10
9	71	L	110	55	128	13	122	38	128	16
10	111	L	121	27	116	47	127	12	109	48
11	115	R	122	23	122	27	114	24	109	33
12	117	R	123	13	126	11	126	20	133	6
13	118	L	115	18	107	40	126	17	118	41
14	123	R	118	45	139	17	128	35	137	18
15	124	R	138	4	138	3	121	5	124	2
16	133	R	121	14	115	20	108	13	97	32
17	134	L	140	15	151	4	132	19	140	11
18	137	R	109	27	110	20	101	30	105	15
19	146	R	109	24	96	53	124	20	101	54
20	149	R	91	48	111	13	92	16	93	18

Table D2- Typical and atypical strides at 6 months post-operative.

			Normal Gait				Fast Gait			
N	**ID**	**Side**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**
1	35	R	110	11	114	7	114	20	120	9
2	32	L	103	75	124	38	150	17	137	25
3	36	R	123	29	136	6	115	18	112	12
4	47	D	134	29	121	51	131	26	111	61
5	51	L	40	39	51	10	45	26	45	15
6	62	R	140	15	136	14	163	14	151	25
7	66	R	133	41	127	54	123	36	114	57
8	60	R	125	34	109	55	127	13	113	35
9	71	L	135	21	143	12	127	21	135	16
10	111	L	144	17	127	50	111	8	107	17
11	115	R	100	47	125	24	119	22	116	26
12	117	R	107	33	126	10	124	21	124	11
13	118	L	103	52	116	18	101	33	105	20
14	123	R	150	48	143	56	134	41	135	42
15	124	R	142	41	90	98	127	33	50	111
16	133	R	133	20	86	117	92	31	99	28
17	134	L	152	17	148	25	126	25	128	25
18	137	R	119	33	136	5	115	26	116	22
19	146	R	119	17	116	26	128	21	118	36
20	149	R	117	16	113	23	106	22	114	16

Table D3- Typical and atypical strides at 6 months post-operative.

			Normal Gait				Fast Gait			
N	**ID**	**Side**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**
1	35	R	98	18	105	12	129	18	135	11
2	32	L	131	26	121	42	81	100	97	72
3	36	R	111	31	119	15	118	27	116	23
4	47	D	132	11	136	11	123	14	124	16
5	51	L	79	14	78	18	28	9	29	7
6	62	R	140	14	136	18	167	9	152	23
7	66	R	127	24	129	15	129	25	127	20
8	60	R	135	30	96	105	111	22	94	50
9	71	L	129	52	150	14	121	27	133	9
10	111	L	144	23	140	40	146	13	145	16
11	115	R	115	44	124	27	84	82	114	19
12	117	R	145	8	143	11	136	15	130	21
13	118	L	112	29	117	23	109	27	108	26
14	123	R	152	44	154	37	132	43	136	34
15	124	R	156	9	154	13	157	5	156	7
16	133	R	123	27	128	22	111	21	115	18
17	134	L	174	4	177	2	149	17	159	10
18	137	R	140	20	141	16	130	14	128	16
19	146	R	121	37	136	13	124	28	70	96
20	149	R	127	8	125	11	32	100	117	15

Table D4- Typical and atypical strides for controls.

		Normal Gait				Fast Gait			
N	**ID**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**	**Num HFPS Left**	**Num Atypical Left**	**Num HFPS Right**	**Num Atypical Right**
1	205	97	10	67	60	95	20	42	136
2	202	103	30	107	19	92	41	100	19
3	209	130	0	129	1	124	3	119	9
4	210	123	8	121	15	116	9	118	10
5	211	138	1	135	4	150	5	139	19
6	213	138	0	134	3	146	3	138	13
7	212	125	0	124	3	153	1	153	1
8	215	120	14	122	14	126	26	130	19
9	217	109	7	112	3	129	8	128	9
10	220	137	14	139	12	130	17	130	18
11	223	104	67	160	9	152	21	157	16
12	224	127	14	116	35	136	18	122	43
13	229	125	17	130	11	124	22	124	26
14	225	128	11	130	9	125	14	127	7
15	228	149	14	153	11	166	15	171	10
16	231	138	20	147	15	134	15	135	16
17	230	146	21	145	25	123	20	125	18
18	235	134	10	127	19	131	11	126	15
19	234	135	13	140	5	125	25	128	19
20	239	110	23	106	25	108	29	101	36

Note: The values contained in these tables were extracted from the software Step32.