

École polytechnique de Louvain

Preliminary Design of an Active Hip Prosthesis for Disarticulated Amputees

Author: **Luis Felipe AYCARDI**
Supervisor: **Renaud RONSSE**
Readers: **Benoit HERMANS, Paul FISETTE**
Academic year 2019–2020
Master [120] in Biomedical Engineering

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Abstract

Lower limb amputations are events that affect human mobility and health at different levels, causing the person to partially or totally lose his/her independence and quality of life. When amputations are performed at lower levels, like the ankle joint, although traumatic, there are several technological advances that can fairly reestablish the patient's healthy biomechanics. However, as the amputation is done at higher levels, like the hip, the challenges to successfully replace the healthy system increase exponentially.

Hip disarticulation and hemipelvectomy are amputations at the level of the hip joint. Individuals who suffer this loss, present reduced levels of motor function compared to other amputees and their mobility is limited by the lack of available prostheses that can efficiently replicate biologically accurate movements. To be able to use a prosthesis, they require a great deal of rehabilitation training and even if they can walk using it, their walking speed and metabolic cost are out of the normal ranges of healthy individuals.

Given the low incidence rates compared to other types of amputations, few studies and technological improvements have been done to help these patients recover their mobility. Evidence in prosthetics have shown that active actuation during gait improves the biomechanics and metabolic costs of amputees, and yet for hip disarticulation no active joints have been released. In this thesis, the concept of active prosthetics is explored with the purpose of designing a prosthetic hip joint. By studying the mechanical advantages of the best passive prosthetic hip joint, together with the supporting evidence of the need of hip actuation, the preliminary design of a prosthetic hip joint was done. A polycentric and compliant electromechanical prosthetic hip joint, fixed anterior to the socket was chosen. The design proposed is based on (i) a timing belt with the proximal pulley fixed, (ii) a four-bar mechanism and (iii) a parallel spring, that reduces the torque and power demands of the joint; thus, increasing energy-efficiency and reducing the size of the motors. The selection of the main components in the catalog, the estimation of the stiffness for the parallel elastic actuator (PEA) and their optimal assembly in the virtual volume of the prosthetic joint were achieved.

Chapter 1

Introduction

This work focuses on the design of an active prosthetic hip joint of a complete-leg prosthesis for disarticulated amputees. The exploration of the design process is addressed throughout the different stages and the analysis of the hip joint is exhaustively presented. This chapter presents some foundations regarding the motivation of this project and existing technologies up to date. The structure of this document is also presented at the end of this chapter.

1.1 Motivation

Lower-limb loss is a strongly disabling condition that affects both the health and the quality of life of the person. Lower-limb amputation can be performed at the level of the foot, calf, thigh, or even at the hip joint where three main cases are presented, as shown in Figure 1.1: (i) a short transfemoral amputation, where the separation is done through the femur, (ii) a hip disarticulation, done by separating the ball (femur's head) from the socket of the hip joint and (iii) a hemipelvectomy, an amputation of the whole lower limb and ipsilateral hemi-pelvis.

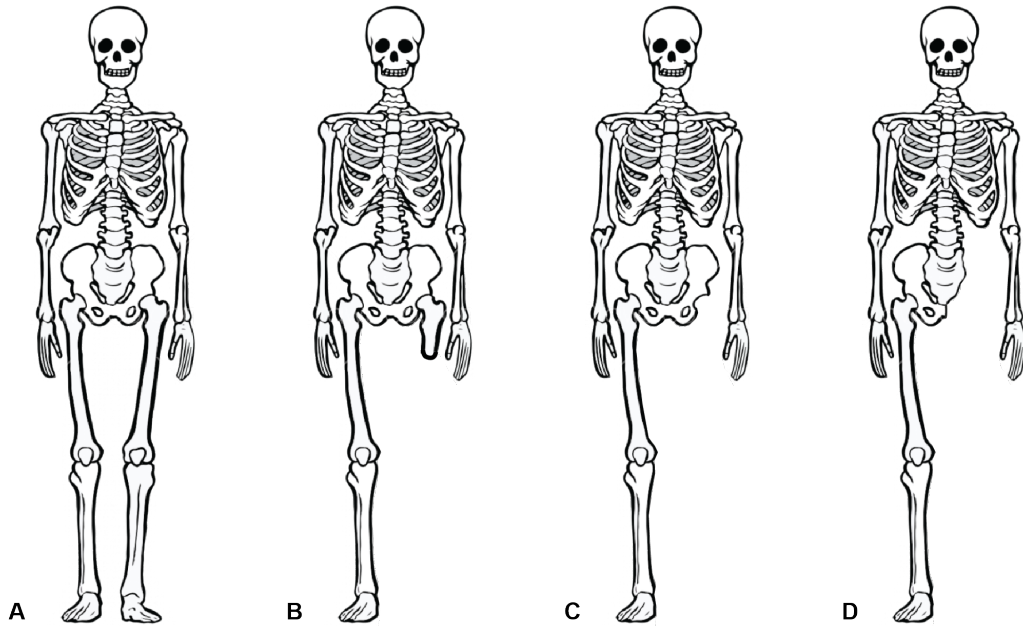


Figure 1.1: Amputations at the level of the hip joint. A. Healthy subject. B. Short transfemoral amputation. C. Hip disarticulation. D. Hemipelvectomy. Image adapted from [1].

Due to this big variation in the level of the amputation, each type of lower-limb amputee faces different challenges in his/hers quest for walking again. This variation is reflected in the fact that higher amputation levels require more complex prostheses and longer rehabilitation therapies, and is also captured in the associated gait biomechanics. As more joints and muscles of the leg are lost due to higher level amputations, the greater is the loss of the normal locomotor mechanisms for body propulsion and stabilization, and therefore, the greater are the energy cost and disability [2]. However, only around 0.5% of all lower-limb amputation surgeries correspond to hip disarticulation [3], which has represented a slow development of technologies to support this traumatic event. Until today, very few studies have reached to disarticulated amputees and even fewer products are available on the market for such conditions, in contrast with other lower level amputations.

The main objective of this thesis is to start the design of a functional prototype of an active hip joint prosthesis for disarticulated amputees, i.e. a joint prosthesis with the capacity to deliver mechanical energy, so that it can be further integrated into a complete-leg prosthesis.

1.2 State of the Art

Amputation at the level of the hip joint is evidently a highly mutilating intervention, which has both functional and psychological repercussions in a person. As it was previously stated, different amputations can be performed at the hip level. However, in this work the main population to consider are both hip disarticulation and hemipelvectomy, where there is no existence of a stump to fit a prosthesis on. While the majority of the lower limb amputations in the developed countries correspond to patients over 60 years of age (mainly due to vascular diseases), most hip disarticulations and hemipelvectomies are performed in the early decades of life [4]. Therefore, they condition the development of a lifetime for patients who suffer it.

1.2.1 Causes

The most common reason for performing these types of amputations is a bone tumor. Even though the incidence of primary malignant bone tumors is relatively low, the mortality related to them is disproportionately high, especially among teenagers and young adults. The global incidence of primary malignant bone tumors shows a specific age distribution pattern with two incidence peaks at 10- 20 yr and a steady increase from 40 yr to the age of 80 yr. In developed countries like the United Kingdom and United States (USA), bone tumors are the third leading tumors in cancer-related mortality in the age group of 15-24 yr and a rough estimation from *Global Incidence of Primary Malignant Bone Tumors* calculates around 1-1.5 cases per 100,000 person years worldwide [5]. Between the different types of tumor, one known as Chondrosarcoma, that occurs predominantly during adulthood, corresponds to 20% of all bone sarcomas and the pelvis is the most common location where it appears [6]. According to this, roughly speaking – in Belgium – about 10 cases of pelvis tumors are diagnosed every year and most likely will end up in some sort of amputation.

The second reason leading to hip level amputation is trauma. Cases are usually related to armed confrontation and some indexes have been released to the public. First, between 2002 and June 2007, 14 US soldiers had sustained a hip disarticulation (HD), with 8 having unilateral amputations and 6 having multiple limb involvement. And second, as of May the 3rd 2010 a total of 997 soldiers with amputations from Operation Iraqi Freedom and Operation Enduring Freedom were treated in US military treatment facilities; and 22 of them sustained hip disarticulations, which correspond to a 2.2% [7]. Finally, other reasons include vascular insufficiency, femoral osteomyelitis and seldomly decubitus, failed hip arthroplasty and congenital abnormality [4].

1.2.2 The Canadian Prosthesis

The technique of hip disarticulation has changed little since it was first described, even though the advent and development of anaesthesia, transfusion medicine, antibiotics and analgesia have caused a dramatic reduction of the perioperative morbidity and mortality [3]. Therefore, for those individuals where such procedures are deemed appropriate, advances in other areas have been in charge of compensating for the huge functional loss. In particular, pelvis prostheses have been developed since the 50's and the most frequently used is called the "Canadian prosthesis", shown in Figure 1.2, which provides the replacement of the whole lower limb [8].

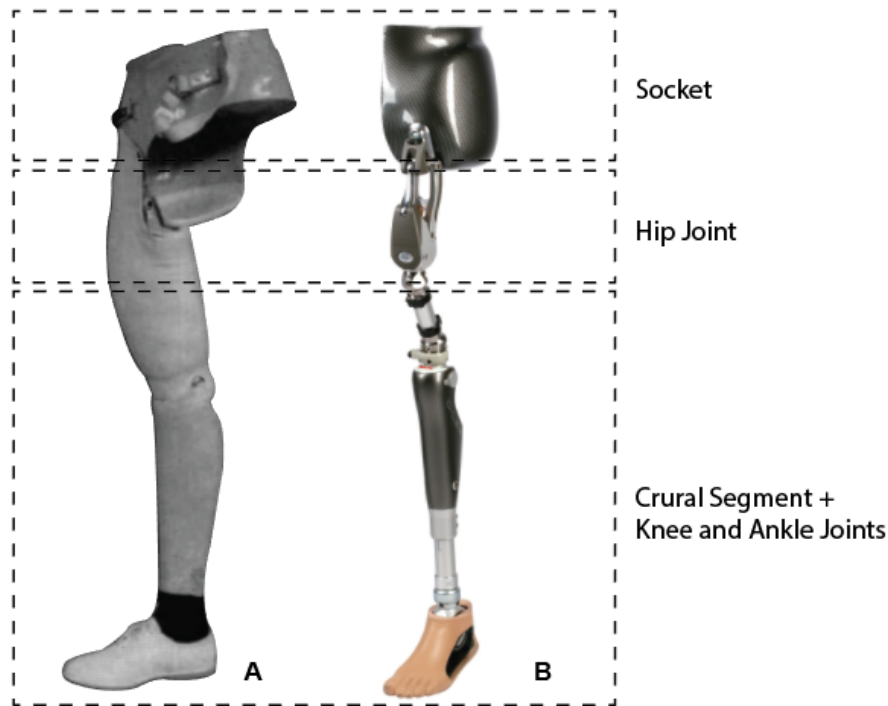


Figure 1.2: Example of Canadian prosthesis. A. Prosthesis in 1953 by the Prosthetics Research Group of the, then, Prosthetic Devices Research Project of the University of California, Berkeley [9]. B. Helix prosthetic system (Helix3D hip joint coupled with a C-leg microprocessor knee joint and a Triton feet) nowadays by Ottobock.

Although this prosthesis is relatively old, it is still the gold standard for disarticulated amputees. The main components of a typical Canadian prosthesis are, as presented in Figure 1.2:

- A socket, which is the primary and critical interface between the amputee's residual limb and the rest of the prosthesis. It is in charge of transmitting the body weight during the stance phase of walking and ensuring stability and control. It is custom-made for every patient from a cast of his/her pelvis stump and is usually made of resin with carbon fibers reinforcement;
- A hip joint itself, located just below the socket. It must be appropriately placed [8] to allow both the stability in extension during the stance phase, and the controlled flexion and foot clearance during the swing phase (i.e. when the leg does not contact the ground);
- A crural segment, i.e. a tube connecting the hip to the prosthetic knee and the rest of the prosthesis. The knee and ankle joint are standard components also used in transfemoral and transtibial prostheses.

Despite sophisticated and expensive, these prostheses do not sufficiently compensate the lower-limb loss, up to the point of restoring normal biomechanical functions. For example, their passive nature does not allow them to replicate the mechanical energy production of a healthy hip during ground-level walking [10]. Since disarticulated amputees rely on an increase and decrease in the lordosis of the lumbar spine ¹ for propulsion when walking with a prosthesis, the weight of the prosthesis is itself notable. Thus, the physical load imposed by walking with a prosthesis is clearly very large and previous experience has shown that these amputees walked about 40% slower and spent about 80 - 250% more energy than able-bodied persons [11].

Consequently, these and other reasons related to awkwardness and intolerance of the socket have led to only about 33% of potential users making daily use of their prostheses [12].

1.2.3 Prosthetic Joints Evolution

An extended review of all prosthetic joint modules (hip, knee and ankle) is out of the scope of the present document, however, a global trend can be outlined. The development of leg prostheses is switching nowadays from passive to active solutions (or bionic limbs) mainly due to the success of miniaturized technologies for energy storage, bio-compatible sensors, and compliant/soft actuators [13]. The initial passive prostheses are capable of storing and releasing mechanical energy in elastic elements to assist in forward propulsion, but they are unable to reproduce the positive mechanical work of a normal joint during level-ground walking [10].

In the case of the ankle joint, active ankle prostheses (also called “Bionic Feet”, see Figure 1.3) are devices equipped with a power actuator, and are thus capable of providing a net positive energy over the walking cycle. Many of these devices are still at research stages but have already shown great capabilities. For example, the ankle in Figure 1.3.B proved to normalize the gait pattern of transtibial amputees, both regarding kinematics and metabolic cost [14]. It’s scientific transfer to the market led to the development of the only commercially available ankle joint with powered propulsion: the "Empower" prosthesis by the German company Ottobock presented in Figure 1.3.C.

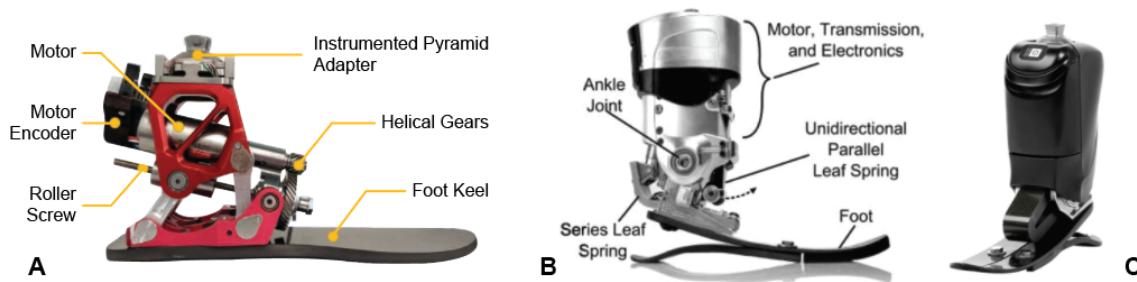


Figure 1.3: Example of active ankle prosthesis. A. Romeo Foot prosthesis by the University of Utah [15]. B. Bionic Foot from the MIT group of Herr [14]. C. Empower prosthesis by Ottobock.

In contrast to the ankle and hip, the knee biomechanics reveal a global energy dissipation over a walking cycle [10], [13]. Therefore, a power actuator is not needed to reproduce natural kinematics, at least for level-ground walking. The simplest knee prosthesis developed were hinge joints being locked during walking and unlocked only for sitting, which produced un-aesthetical gaits. However, recent developments combine springs and hydraulic/pneumatic dampers, that are servo-controlled during both stance and swing phases as the one presented in Figure 1.4.

¹the normal inward curvature of the lumbar and cervical regions of the human spine



Figure 1.4: Example of hydraulic/pneumatic controlled knee prosthesis. A. Orion3 prosthesis by Blatchford. B. Rheo Knee prosthesis by Ossur. C. Genium prosthesis by Ottobock.

Finally, and coming back to the hip joint, the development of active hip modules has not been part of the research and company development agendas. Likely due to the intrinsic challenges of the whole-limb replacement, and the small incidence of hip disarticulations with respect to transfemoral and transtibial amputations [8]. Therefore, the joints available today correspond to passive prostheses as presented in Figure 1.5. Up to recently, the hip was replaced by a hinge and most of the joints corresponded to monocentric configurations, which could include at most an hydraulic control unit. However, in 2008 Ottobock (HealthCare GmbH, D) developed the Helix3D hip joint (see Figure 1.5.C) to improve the gait pattern of amputees through the use of an hydraulic control, a polycentric joint design and flexion assist springs [16]. The axis geometry is based on a 4-bar mechanism, which makes it possible to link hip flexion and extension to hip joint rotation. It consists of two ball joints and two single-axis connections, with the rear axis tilted in relation to the structure. In comparison with other conventional joints, this one claims to (i) produce less sudden pelvic movements during weight transfer, (ii) support for swing phase initiation, (iii) control hip movements during the swing phase and (iv) allow three-dimensional movements in terms of the relationship between hip joint extension/flexion and transversal pelvic rotation [17].



Figure 1.5: Example of passive hip prosthesis by Ottobock. A. 7E7 modular monocentric prosthesis. B. 7E9 monocentric prosthesis with hydraulic control unit. C. 7E10 Helix3D prosthesis.

Even though biomechanical advantages and characteristics of this gold-standard hip joint will be analyzed further (see Section 3.2), some studies have shown its functional benefits over

other conventional hip joints. These enhancements include [18]:

- Improved hip extension control and polycentric design offer increased security during stance phase with respect to uncontrollable hip motion;
- Longer, more natural period of hip extension during stance phase;
- More constant hip flexion movement during swing phase;
- Increased knee flexion during stance phase;
- Increased knee flexion during swing phase;
- Significant reduction in the range of pelvic tilt.

Nevertheless, the complex nature of the human hip joint has not been properly addressed by active prosthetic joints and the benefits of such a module could even overcome the ones reached so far by passive devices. Consequently, the design of a functional prototype of an active hip joint prosthesis for disarticulated amputees, that can be further integrated into a complete-leg prosthesis, will be addressed throughout this project.

1.3 Overview

This Master Thesis report is structured as follows.

Chapter 2 presents the biomechanical definitions and nomenclatures useful to understand and characterize the hip joint. The anatomy and classification of the hip is studied to then analyze its kinematic and kinetics during locomotion. Chapter 3 addresses the literature review concerning the degradation of the locomotion physiology in hip disarticulation. The biomechanics of walking of hip disarticulated using the Helix3D prosthetic joint are analyzed. This chapter also presents the few active prototypes of prosthetic hip joints. Chapter 4 begins with the formulation of the mechanical design following the Cross model designing technique. The requirements and mechanisms to achieve them are evaluated and a final configuration goal is proposed. Chapter 5 addresses the mechanical implementation of the main components. It makes some remarks over the compliance selection, the motor and gearbox specifications and then concludes with a conceptual design of the hip joint. Finally, Chapter 6 presents the conclusions and some recommendations for future works.

Chapter 2

Human Biomechanics

2.1 Introduction

The biomechanics of human movement can be defined as the interdiscipline that describes, analyzes, and assesses human movement. There is a wide variety of physical movements involved from the gait of the physically handicapped to the performance of a superior athlete to daily activities. In all cases, the physical and biological principles that apply are the same and what changes are the specific movement tasks and the level of detail that is being asked about the performance of each movement [10]. Some of the basic biomechanical definitions and nomenclature useful for further descriptions and characterisation will be addressed in the beginning of the chapter. Afterwards, the anatomy and biological elements of the hip joint with its normal kinematics will be studied. Then, the biomechanics of the hip joint during locomotion, including the control mechanisms and normal kinematic and kinetics, will be presented.

2.2 Principles and Definitions

In order to understand the structure and nature of the hip joint, a reminder of some basic definitions and concepts must be done.

2.2.1 Directional Terms

Anatomists and health care providers use terminology that can be bewildering to the uninitiated. Anatomical terms derive from ancient Greek and Latin words. Because these languages are no longer used in everyday conversation, the meaning of their words does not change. Certain directional anatomical terms will appear throughout this document. These terms are essential for describing the relative locations of different body structures [19], as presented in Figure 2.1.A.

- *Anterior/ventral* describes the front or direction toward the front of the body;
- *Posterior/dorsal* describes the back or direction toward the back of the body;
- *Superior/cranial* describes a position above or higher than another part of the body;
- *Inferior/caudal* describes a position below or lower than another part of the body proper; near or toward the coccyx, or lowest part of the spinal column;
- *Lateral* describes the side or direction toward the side of the body;
- *Medial* describes the middle or direction toward the middle of the body;
- *Proximal* describes a position in a limb nearer to the point of attachment to the body;
- *Distal* describes a position in a limb that is farther from the point of attachment or the trunk of the body;

- *Superficial* describes a position closer to the surface of the body;
- *Deep* describes a position farther from the surface of the body.

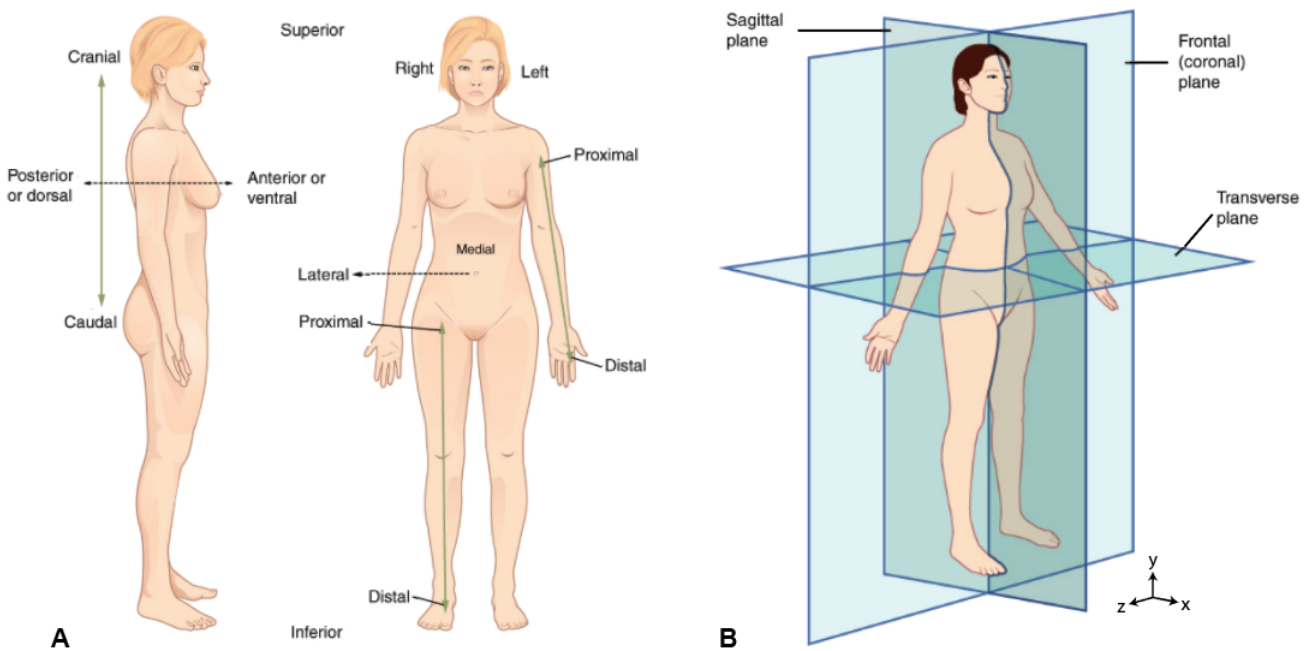


Figure 2.1: Nomenclature related to the human body. A. Directional terms applied to the human body. B. Planes of the body, the three planes most commonly used in anatomical and medical imaging are the sagittal, frontal (or coronal), and transverse plane. Images from [19].

2.2.2 Body Planes

An anatomic plane is an imaginary two-dimensional surface that passes through the body. There are three planes commonly referred to in anatomy and medicine, as illustrated in Figure 2.1.B: (i) the sagittal plane that divides the body vertically into right and left sides. If this vertical plane runs directly down the middle of the body, it is called the midsagittal or median plane. If it divides the body into unequal right and left sides, it is called a parasagittal plane or, less commonly, a “longitudinal section”, (ii) the frontal plane that divides the body into an anterior (front) portion and a posterior (rear) portion. The frontal plane is often referred to as a coronal plane (“corona” is Latin for “crown”) and (iii) the transverse plane that divides the body horizontally into upper and lower portions. Transverse planes produce images referred to as “cross sections” [19].

2.2.3 The Musculoskeletal System

The primary functions of this system include supporting the body, allowing motion, and protecting vital organs. However, to understand how exactly these are achieved a reminder of the fundamental tissues that are throughout the human body is needed. These are [19]:

- The epithelial tissue that covers all internal and external body surfaces and includes structures such as the skin and the inner lining of the blood vessels;
- The nervous tissue that provides a two-way communication system between the central nervous system (brain and spinal cord) and muscles, sensory organs, and various systems;

- The Connective Tissue (CT), which includes bone, cartilage, tendons, ligaments, and blood tissue, provides structural and metabolic support for other tissues and organs of the body;
- The muscle tissue, classified functionally as either voluntary or involuntary, and structurally as either smooth, striated/skeletal, or cardiac. There are approximately 430 skeletal muscles in the body, each of which can be considered anatomically as a separate organ. Of these, about 75 pairs provide the majority of body movements and postures [20].

The musculoskeletal system is composed of the last 2 tissues (connective and muscle) and, in short, describes how bones are connected to other bones and muscle fibers via connective tissues such as tendons and ligaments. The bones provide stability to the body. Different bones are connected by joints to allow motion and cartilage prevents the bone ends from rubbing directly onto each other. The muscles keep bones in place and also play a role in the movement of bones, as they contract to move the bone attached to the joint.

2.2.4 Joints

In terms of biomechanics, the structural characteristics condition the type of movement that a joint can perform and therefore it would be the focus of this section. The structural classification of joints refers to whether the articulating surfaces of the adjacent bones are directly connected by connective tissues, or whether the articulating surfaces contact each other within a fluid-filled joint cavity. These differences serve to divide the joints of the body into three structural classifications: (i) fibrous joints, where the adjacent bones are united by fibrous connective tissues, (ii) cartilaginous joints, where the bones are joined by hyaline cartilage or fibrocartilage and (iii) synovial joints, where the articulating surfaces of the bones are not directly connected, but instead connect each other within a joint cavity that is filled with a lubricating fluid. These last ones allow free movement between the bones and are the most common joints of the body. Likewise, synovial joints are subdivided based on the shapes of the articulating surfaces of the bones that form each joint. The six types of synovial joints are pivot, hinge, condyloid, saddle, plane, and ball-and-socket joints as shown in Figure 2.2 [19].

The number of independent modes of motion at a joint is called the degrees of freedom (DoF). A joint can have up to 3 degrees of angular freedom, corresponding to the three dimensions of space. If a joint can swing in one direction or can only spin, it is said to have 1 DoF (i.e. the hinge joint in Figure 2.2.b). If a joint can spin and swing in one way only or it can swing in two completely distinct ways, but not spin, it is said to have 2 DoF (i.e. the saddle joint in Figure 2.2.c). If the bone can spin and also swing in two distinct directions then it is said to have 3 DoF. In the human body, only the shoulder and hip (ball-and-socket joints) have 3 DoF [21].

2.3 The Hip Joint

The hip joint, main focus of this project, is the articulation between the large rounded head of the femur and the acetabulum of the pelvis. Because it serves as the base of the lower extremity, the hip must provide a wide range of movement in all three planes. Generally, hip movement can occur (1) by rotating the femur relative to a stationary or otherwise fixed pelvis, as when lifting the leg up a step, or (2) by rotating the pelvis (and often the trunk) relative to a fixed femur. The second situation occurs during many common activities such as walking, running, and transitioning from sitting to standing or picking up objects off the ground [22].

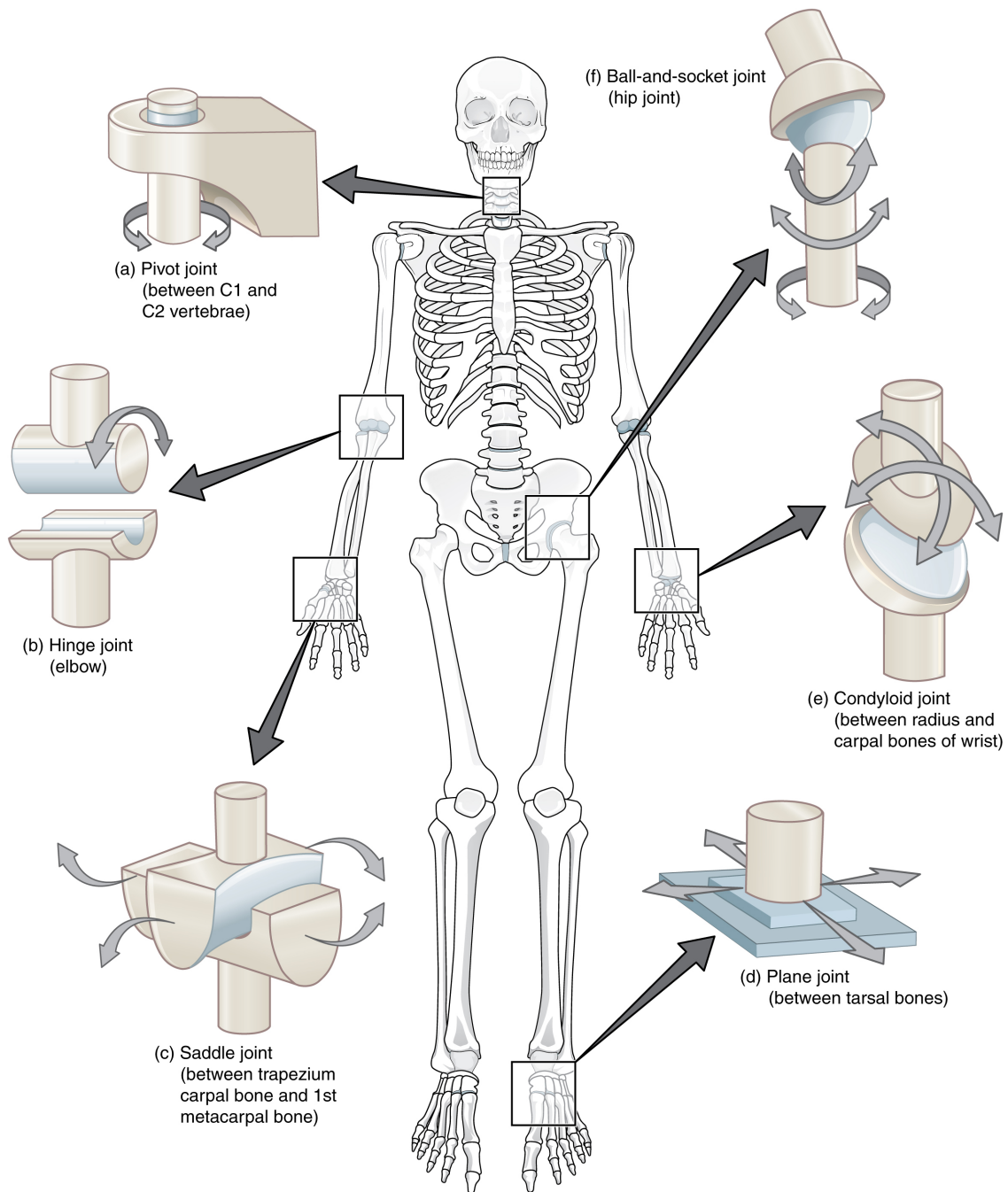


Figure 2.2: The six types of synovial joints allow the body to move in a variety of ways. (a) Pivot joints allow for rotation around an axis. (b) The hinge joint of the elbow works like a door hinge. (c) The articulation between the trapezium carpal bone and the first metacarpal bone at the base of the thumb is a saddle joint. (d) Plane joints, such as those between the tarsal bones of the foot. (e) The radiocarpal joint of the wrist is a condyloid joint. (f) The hip and shoulder joints are the only ball-and-socket joints of the body. Image from [19].

In order to provide stability during these activities the hip has many anatomic features. First, the femoral head is stabilized by a deep socket that is surrounded by an extensive set of capsular ligaments. Moreover, numerous muscles provide additional stability, while others supply the large forces required for a wide variety of functional activities. A brief discussion of the structure, kinematics, and muscles of the hip joint with an emphasis on prosthetic requirements is now presented.

2.3.1 Bone Anatomy

The pelvis (hip bone), also called “os coxa” or “innominate bone” (meaning “name- less” in Latin) is formed by the union of three bones: the ilium, the ischium, and the pubis as illustrated in Figure 2.3.B.

- The ilium is the wing-shaped superior portion of the innominate (in pink in the Figure). The iliac crest is a long, palpable ridge of bone that marks the superior border of the ilium. The height of the right and left iliac crests are often compared by clinicians to determine pelvic symmetry;
- The ischium is located on the posterior-inferior aspect of the innominate (in green in the Figure). The ischial tuberosity is a bumpy projection to the posterior-inferior aspect of the ischium that serves as the proximal attachment for three of the four hamstring muscles. In a seated position, people literally sit on their ischial tuberosities;
- The pubis (in purple in the Figure) is the smallest of the three bones and consists of a body and inferior and superior arms, or “rami”. This relatively immobile joint completes the anterior “ring” of the pelvis;
- The acetabulum is the deep, cup-shaped structure that encloses the head of the femur at the hip joint (see Figure 2.3.B). It is interesting to note that the acetabulum is formed by a combination of all three bones of the pelvis previously mentioned: the ilium, pubis, and ischium.

The femur is the strongest and the longest bone in the body. The proximal end of the femur consists of a head, a neck, and a greater and lesser trochanter (see Figure 2.3.C). Of the many sections of this bone, some of the most important to mention are: (a) The greater trochanter which is the large, palpable projection of bone that extends laterally from near the junction of the femoral neck and shaft. It serves as an important distal attachment for numerous muscles of the hip. (b) The trochanteric fossa which is a small pit on the posterior-medial side of the greater trochanter. Many of the short external rotators of the hip attach near or on the trochanteric fossa. (c) More distally is the linea aspera, a line of slightly raised bone that courses along much of the posterior side of the femur. This bony ridge serves as the distal attachment for many of the adductor muscles and as the proximal attachment for two of the quadriceps muscles [22].

2.3.2 Arthrology

The hip is anatomically designed to withstand the large and potentially dislocating forces that can routinely occur during walking and other, more rigorous, activities. For example, during activities as walking and running, the hip supports around 1.3–5.8 and 4.5–8 times the body weight, respectively [24]. Additionally, stair climbing and descending increase the loads on the hip by approximately 10% and 20%, respectively [25]. Ligaments play an important role in stabilizing the hip and the external surface of the capsule of the joint is reinforced by a set of thick and important ligaments. But why is their role highly important? The answer is in the way most persons can stand for long periods of time using only minimal amounts of muscular energy about the hip. This nearly “free lunch” condition is attained primarily because of the relationship between the ligaments of the hip and the line of gravity in the human body. While standing in full upright posture, the line of gravity normally travels just posterior to the medial-lateral axis of rotation of the hips. Therefore, gravity provides a passive extension torque at the hips (causing a backward bending of the pelvis over the femur), but because the ligaments of the hip are stretched in extension, they generate a passive rubber band-like flexion torque that offsets this “gravity torque” [22].

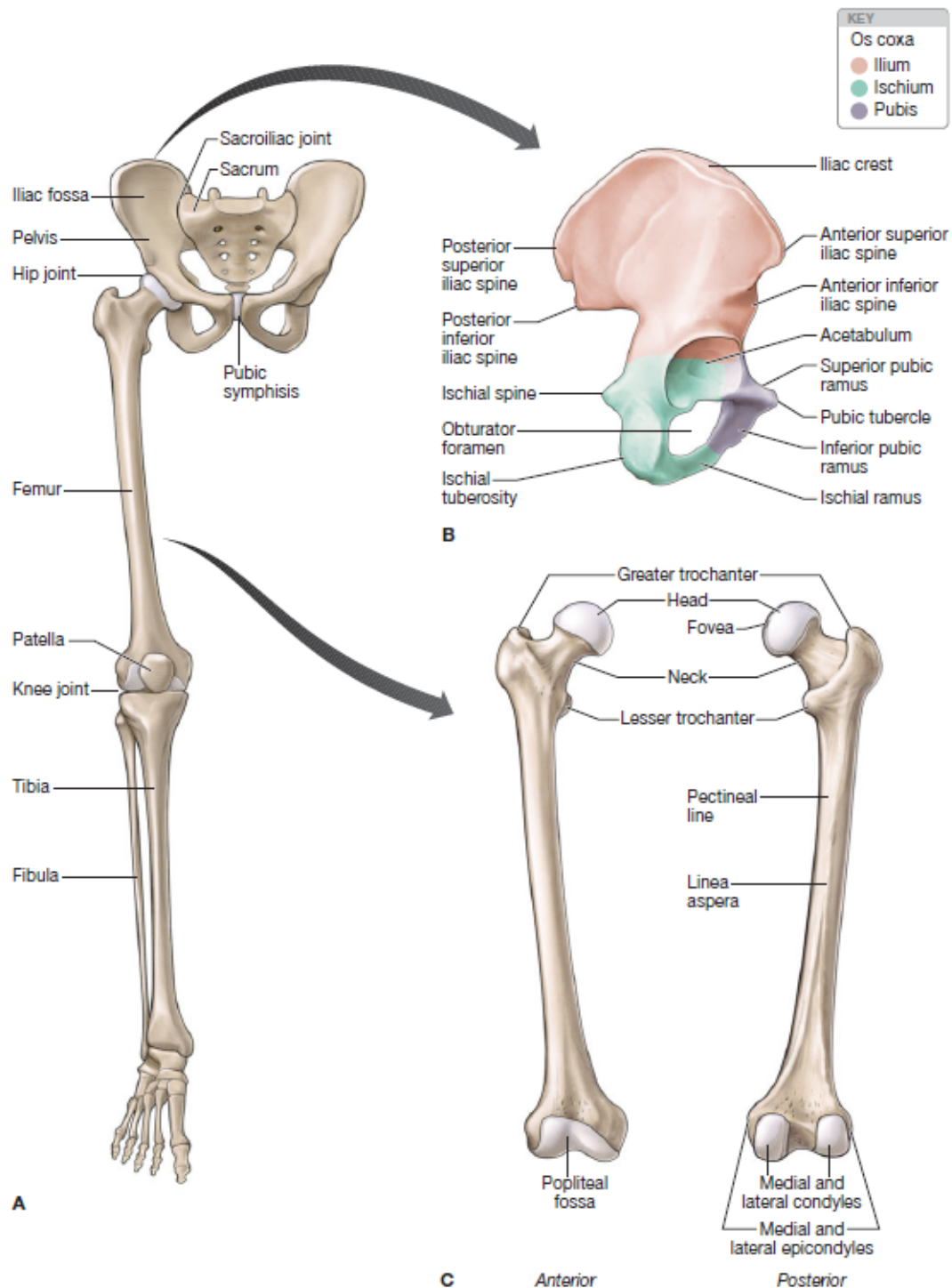


Figure 2.3: A. Skeleton of the lower limb. B. Osteology of the os coxa (pelvic bone). C. Femur. Image from [23].

2.3.3 Kinematics

As classified previously, the hip joint has three DoF, which permits motion in three planes: sagittal (flexion and extension around a transverse axis), frontal (abduction and adduction around an antero-posterior axis), and transverse (internal and external rotation around a vertical axis) (see Figure 2.4). All three of these axes pass through the center of the femoral head. Active range of motion of the hip is variable, but the normal ranges are presented in Table 2.1.

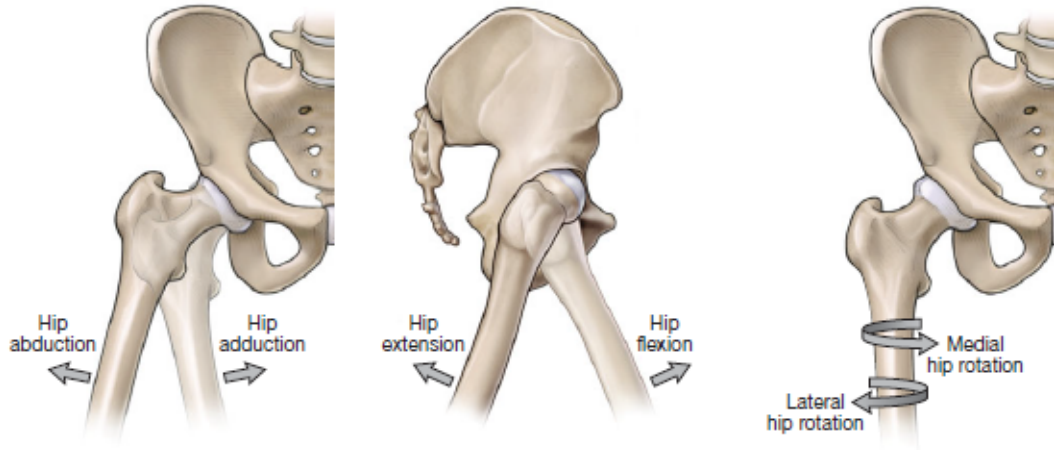


Figure 2.4: Movements of the hip. Images from [23].

Table 2.1: Summary of hip kinematics motion.

Motion	Normal Range of Motion ($^{\circ}$)	Axis of Rotation	Plane
Flexion	0-120	Medial-lateral	Sagittal
Extension	0-20	Medial-lateral	Sagittal
Abduction	0-40	Anterior-posterior	Frontal
Adduction	0-25	Anterior-posterior	Frontal
Internal rotation	0-35	Vertical	Horizontal
External rotation	0-45	Vertical	Horizontal

During walking, the hip extensor muscles have a vital, stabilizing role at the onset of each stride. As body weight is dropped into the forward limb, the extensor muscles contract sharply to preserve upright stance by resisting the forward fall of the pelvis and the trunk [26]. The most stable position of the hip is the normal standing position: hip extension, slight abduction, and slight internal rotation [27]. The commonly cited open-packed (resting) positions of the hip are between 10 and 30 degrees of flexion, 10–30 degrees of abduction, and 0–5 degrees of external rotation [21].

This brief characterisation of the hip joint shows how complex this joint is and how important it is to the human body. Its role in daily life activities such as walking and climbing stairs is crucial and, for that reason, a closer look into how the hip behaves during locomotion is key for this project.

2.4 Human Locomotion

Walking is the fundamental and the most important phenomenon in space and time, reflecting locomotor characteristics of an individual [28]. Therefore, understanding locomotion is crucial in the development of a prosthetic joint and even more so in one as important as the hip. In human walking, a gait cycle is a period beginning with the initial contact between the heel and the ground of one leg up to the subsequent contact between the heel and the ground of the same leg (see Figure 2.5). A basic unit of the human gait is a step which is divided into two basic phases: the stance phase and the swing phase.

In the stance phase, a foot touches the ground and it takes approximately 60% of the overall cycle duration. The stance phase includes the double support, when both limbs touch the

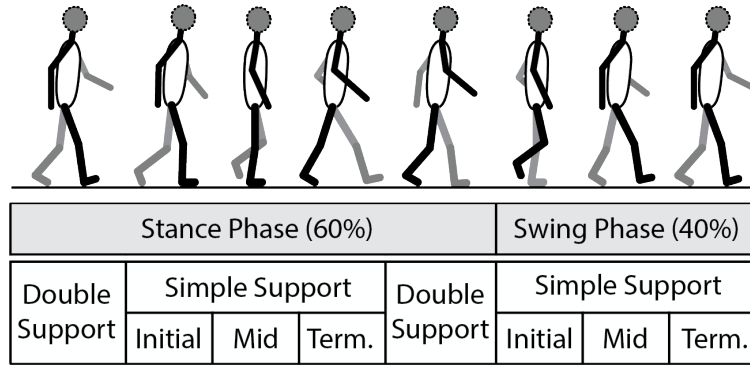


Figure 2.5: Phases of human locomotion.

ground, and the single support, when only one limb touches it. The main task of the support limb is to transfer the pelvis and the upper body part from the back position to the front position in the lowest arch. In this way, it avoids the centre of gravity from lowering too low to then springs-back the impact. After the foot steps onto the ground, the body weight impact evokes the reaction that transfers it into the propulsive push-off force. The push-off begins with the plantar flexion of the ankle, by which the heel rolls away from the ground. Then, the ankle lifts angle-wise upward and forward. The push-off is completed when a foot leaves the ground [29].

The swing phase begins in the end of the stance phase, immediately after the push-off, and takes 40% of the gait cycle. The objectives of the swing phase of gait are foot clearance over the ground, forward swing of the limb and preparation of the limb for stance. The swing phase can be broken down into 3 sub-phases: initial, mid and terminal swing. During initial swing, the hip, knee, and ankle are flexed to begin advancement of the limb forward and create clearance of the foot over the ground. During mid-swing, limb advancement continues and the thigh reaches its peak advancement. During terminal swing, the final advancement of the shank takes place and the foot is positioned for initial foot contact to start the next gait cycle.

The precise duration of all these gait cycle intervals (stance and swing) varies with the person's walking velocity: increasing velocity increases total stance and swing times, while it lengthens single stance and shortens the two double stance intervals [30].

All the same, to successfully understand the way this gait cycle is carried out by the human body and who are the main actors of this functional performance, one should study the control mechanisms of locomotion.

2.4.1 Control in Locomotion

A very useful way to understand the control mechanisms in human locomotion is as presented by Pons et al. [31], in the framework of two European projects (H2R and BETTER). According to the authors, the human musculoskeletal and neural-motor system is highly optimized for efficient locomotion; where efficiency, stability and voluntary modulation of human gait are a result of a combination of features spanning the human musculoskeletal, sensorimotor and neural systems. Main aspects of these systems include the functional morphology, the synergistic coordination of motor activity, the phase modulation of muscle activity and cognitive skills.

They identified the functional morphology (musculoskeletal system) being highly optimized for efficient biped locomotion [32] since it allows exploiting the inherent dynamics to reduce energy consumption and control effort, and consequently, results in natural looking motions. It also contributes to the capability of self-stabilization, since the elastic properties of muscles and tendons increase stability without active control. Muscles often span multiple joints, which results in energy-saving and power transfers when a movement simultaneously requires negative

power at one joint and positive power at another joint. This allows to make effective use of passive elastic properties to generate part of the required force or power without metabolic cost, especially when muscle-tendon units span multiple joints [31]. The main muscles actuating the hip joint are presented in the Appendix A of this document.

Moreover, it is widely accepted that human locomotion depends both on basic patterns generated at the spinal level, and the volitional and reflex-dependent fine control of these patterns at different levels (see Figure 2.6). Motion intentions originate from supraspinal input, which along with afferent feedback serves to modulate basic underlying locomotor patterns within a network of spinal interneurons (central pattern generator). Efferent stimulation is transmitted through motor neurons to individual muscle groups, which are recruited to effect the movement. Afferent feedback, including all sensory neurons, is used to directly modulate motor commands via reflex arcs, thus contributing to the efficiency of gait under normal conditions and stability in the face of unexpected perturbations. Sensory information is also transmitted to the brain, where it is combined with inputs from the visual, auditory, and vestibular systems to provide information required for the maintenance of balance, orientation and control of precise movements [13].

Mapping this control loop and considering the characteristics of the functional morphology in the hip joint provides a framework for the development of the prosthetic module. In Figure 2.6, besides the nominal control loop for locomotion, details of the functional morphology are displayed particularly for the performance of the joint while walking. The main muscles (from section 2.3.3) involved in the gait cycle are presented according to the DoF that they control. Additionally, their periods of activation during the cycle are shown in grey rectangles.

The most important fact to remark from such a complex system and its behavior during walking is the contribution of each DoF and how they interact coordinately. The DoF that govern each of the phases based on the muscles involved is specified below the table. Extension and flexion are, obviously, the main actors of human locomotion and that is why, so far (as presented in section 1.2.3), the majority of prosthetic hip joints have only included this DoF. Nevertheless, abduction and adduction do also play an important role in the gait. In contrast, rotations even though present appear to have a smaller influence in the normal gait cycle. To corroborate such an appreciation, more quantifiable and precise data should be analyzed. In biomechanical terms this is reflected in the kinematic and kinetics of the joint: i.e. angles, moments, power and others characteristics.

2.5 Hip Kinematics and Kinetics During Locomotion

The analysis of the hip kinematics (motion descriptive parameters) and kinetics (motion causing parameters) is vital to capture the characteristics of the biological joint that a prosthetic device should intend to reproduce. In other words, it paves the way towards the specifications of the prosthetic joint. Activities like walking, climbing and sitting are basic for healthy human beings but represent a major challenge to disarticulated amputees. This section will explore the performance of such activities.

2.5.1 Normal Walking

The biomechanics of “normal” walking at a natural pace are well established. Kinematic and kinetic data of the hip [33] for all three planes of motion are given in Figure 2.7. Notice that the moment (torque) and power data is normalized by body weight, Kg. In general terms, all the curves present a consistent profile/shape among the different references in the literature, as well as the ranges of the angles in each plane [33]–[35]. However, there is a big variability in

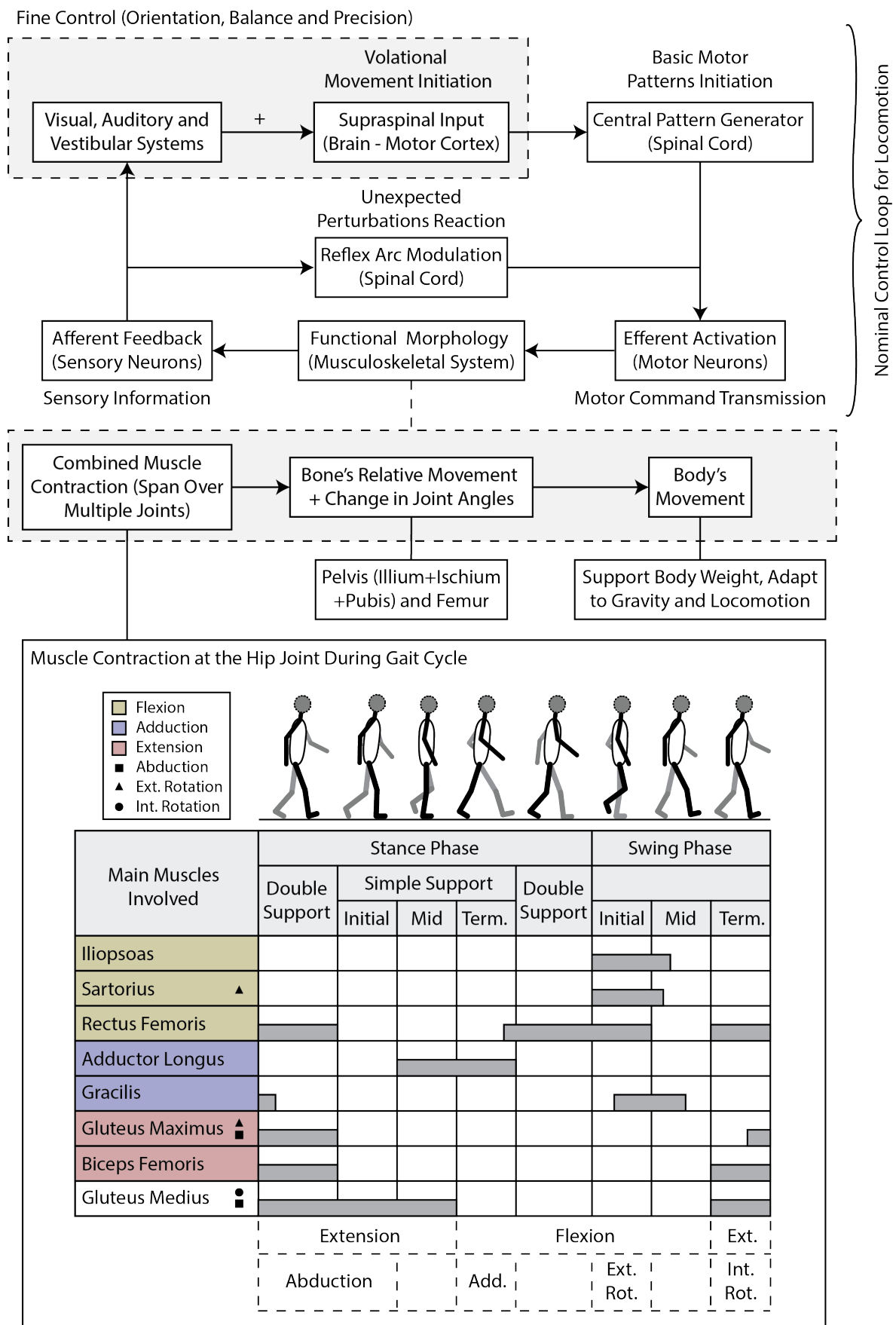


Figure 2.6: Nominal sensory-motor control loop for human locomotion with a specific focus on the role of the hip joint and its different muscle groups. In yellow: muscles involved with hip flexion. In blue: muscles related to adduction of the hip. In red: muscles used during the extension of the hip.

the magnitudes of the torque and power curves, which is something particular of the hip joint and is not that evident in other joints like the knee or ankle. Reasons for this variability are not clear, but one can hypothesise it is due to the way the authors defined the pivot of the hip each time, which may have led to different lever arms in the calculation of the dynamics.

Examining the joint angle data, one finds that the range of motion in the sagittal plane at the hip is around 40° and in the frontal and transverse planes the hip exhibits a range of motion of about 10° . With respect to sagittal plane joint moments, the largest peak extensor moment occurs just after contralateral toe-off (around 10% gait cycle). Meanwhile, the largest peak flexor moment occurs around the 50% gait cycle. The peak hip abductor moment is comparable in magnitude and simultaneous to the peak hip extensor moment, which are considerably higher than the rotation peak moments. This corroborates the necessity of conducting three-dimensional analyzes (or at least two-dimensional) of joint kinetics instead of the typical sagittal plane analysis included in the literature.

To understand the notation in the power curve at the sagittal plane a small clarification should be made. At heel contact, the hip is almost at its maximum flexion angle. Then, an extensor torque is applied to initiate extension of the hip. This requires positive power (H1 region). Hip power shifts from positive to negative at about 20% stride, when the center of mass is at its highest position. Terminal stance is marked by a flexor moment, while the hip is still extending. This means that part of the potential energy released from the falling center of mass is absorbed in the hip (H2 region) to decelerate the thigh and prevent the hip from overextending. At pre-swing, the hip starts to flex. The torque remains flexor, resulting in a positive burst of power (H3 region), helping to lift the foot and initiate swing. The power level increases rapidly as the inertial load of the swinging limb becomes important [36].

When the power data is examined, some different trends exist as compared to the moment data when one considers kinetic measures about the three orthogonal axes during walking. Specifically, the angular velocities are larger in the sagittal plane versus the other two planes, resulting in proportionately larger power values for this primary plane of movement. For example, hip peak power generated in the frontal plane is only one-third the respective sagittal plane values. However, the peak power absorbed at the hip in the frontal and sagittal planes is approximately equal. During normal walking, toe-off occurs at approximately 60% of the gait cycle, and it is evident from these figures that the moments and powers are generally small during the swing phase (60% to 100% gait cycle). A summary of the peak kinetic values associated with all three planes of motion is given in Table 2.2. Values are again presented normalized by body weight, Kg. Which means that, for example, for a 75 Kg person the extensor moment peak would be around 86.25 Nm.

Torque-Angle Diagrams

One common way of visualizing the joint kinetics and kinematics is by means of a torque-angle graph as shown in Figures 2.8, 2.9 and 2.10 for the sagittal, frontal and transversal planes respectively. The main advantage of this approach is how the energy (enclosed area) and the stiffness-like behavior (slope of the torque-angle curve) of the processes can be visualized. However, this results in the erasing of all time-related aspects, which might be adverse for certain analyses. For this reason, for the sagittal and frontal graphs (which have proven to be more influential in the gait cycle) a time correlation was added, marking important events for each one. In them, the direction of the arrows correspond to the direction of moment regarding the percent of gait cycle. When by following the arrows in the curve a clock-wise rotation is done, the enclosed area represents the energy generated in the process. On the contrary, when the direction is counter clock-wise, the area represents the energy that is absorbed in the process.

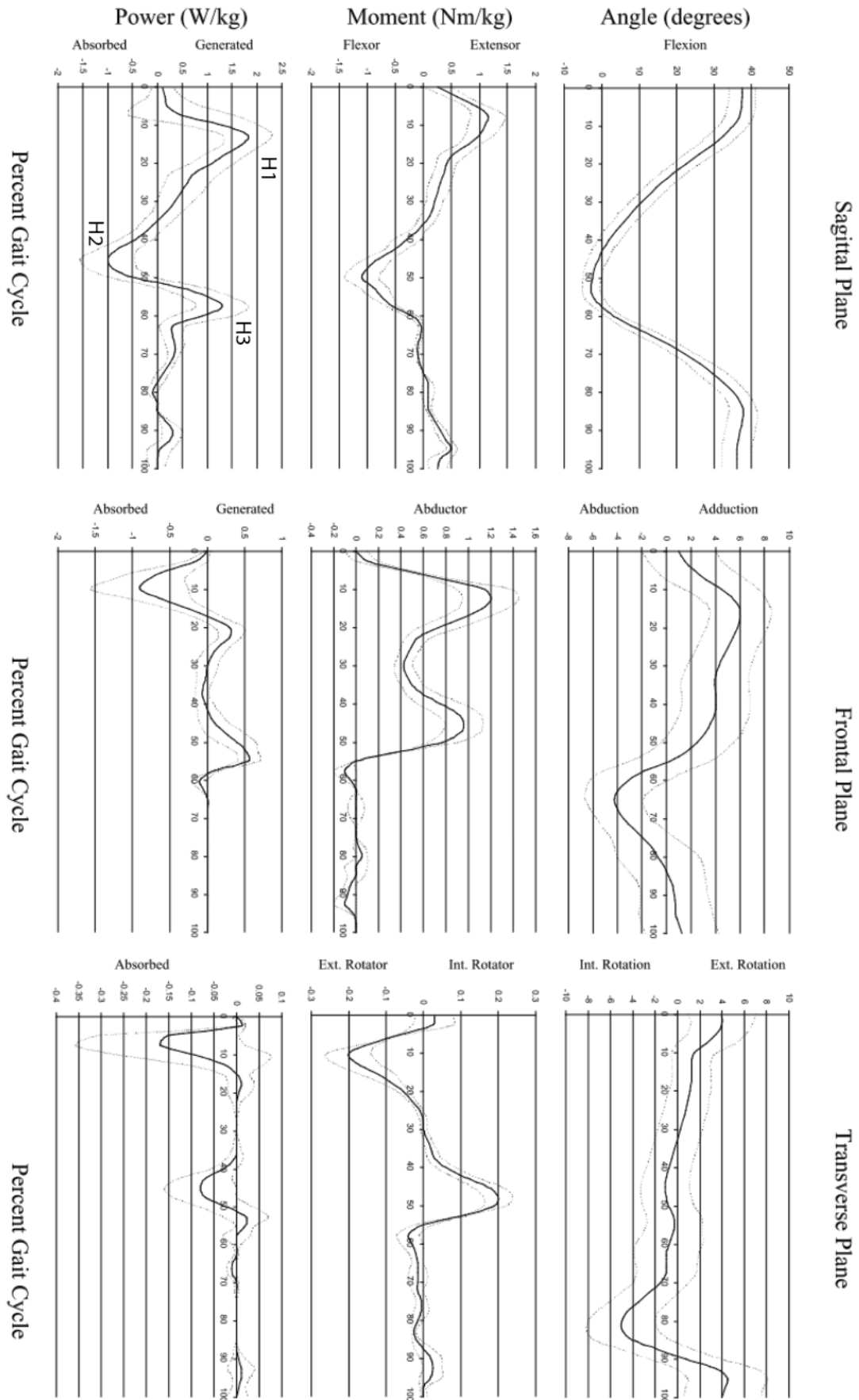


Figure 2.7: Hip joint angles, moments, and powers in the sagittal, frontal, and transverse planes during one cycle of normal walking. (Solid lines are mean values and dotted lines are $\pm$ one standard deviation.). Image from [33].

Table 2.2: Hip joint peak kinetic values during one cycle of normal walking. Data from [33]. Values in brackets correspond to those from Winter [34] for the sagittal plane.

Variable	Plane	Action	Mean $\pm$ SD
Moment (Nm/Kg)	Sagittal	Extensor	1.15 ± 0.30 (0.9)
		Flexor	1.10 ± 0.30
	Frontal	Abductor	1.20 ± 0.25
		Adductor	0.10 ± 0.10
	Transverse	Inversion	0.20 ± 0.04
		Eversion	0.20 ± 0.06
Power (W/Kg)	Sagittal	Generation	1.80 ± 0.50 (1.4)
		Absorption	1.0 ± 0.55
	Frontal	Generation	0.55 ± 0.15
		Absorption	0.90 ± 0.60
	Transverse	Generation	0.02 ± 0.06
		Absorption	0.17 ± 0.16

It is worth remarking once again that the peak values (shown also in the extreme of the areas) for both the sagittal and frontal planes are similar between each other and considerably higher when compared to the transversal plane. The average power in the sagittal plane is positive, while in the others it is roughly neutral. Additionally, the torque-angle relations of the sagittal and frontal planes seem to follow in a certain degree as a spring (either linear or nonlinear). However, this will be addressed in the next section of the document.

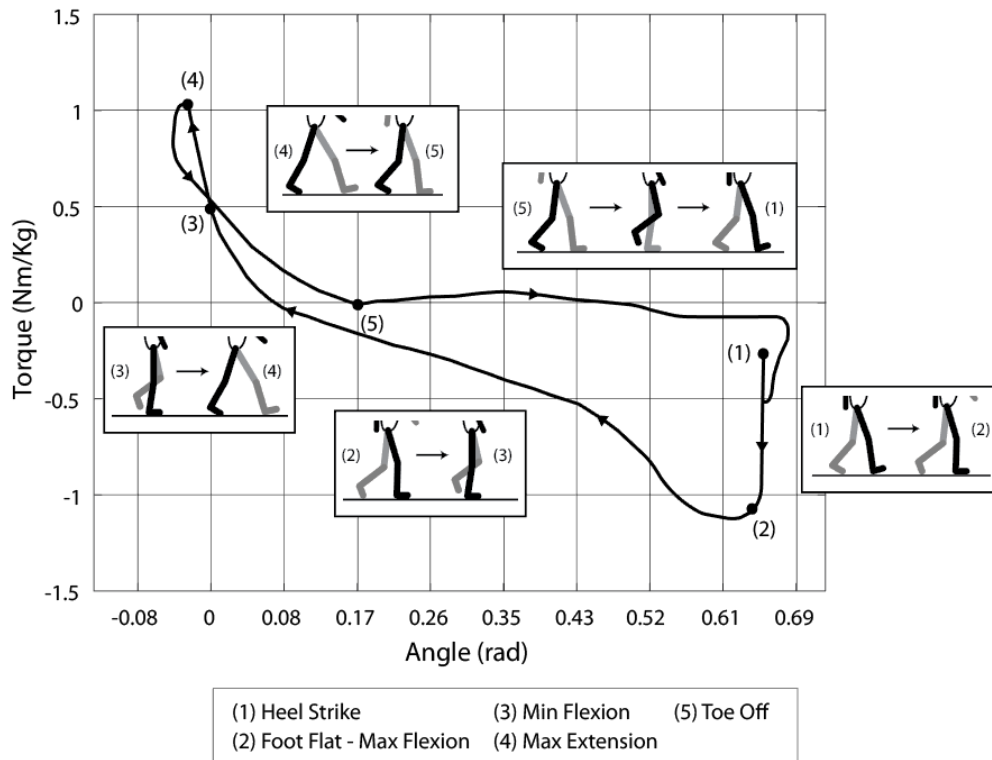


Figure 2.8: Sagittal plane hip torque-angle plot with data extracted from [33]. Main events regarding the gait and the sagittal angles are highlighted.

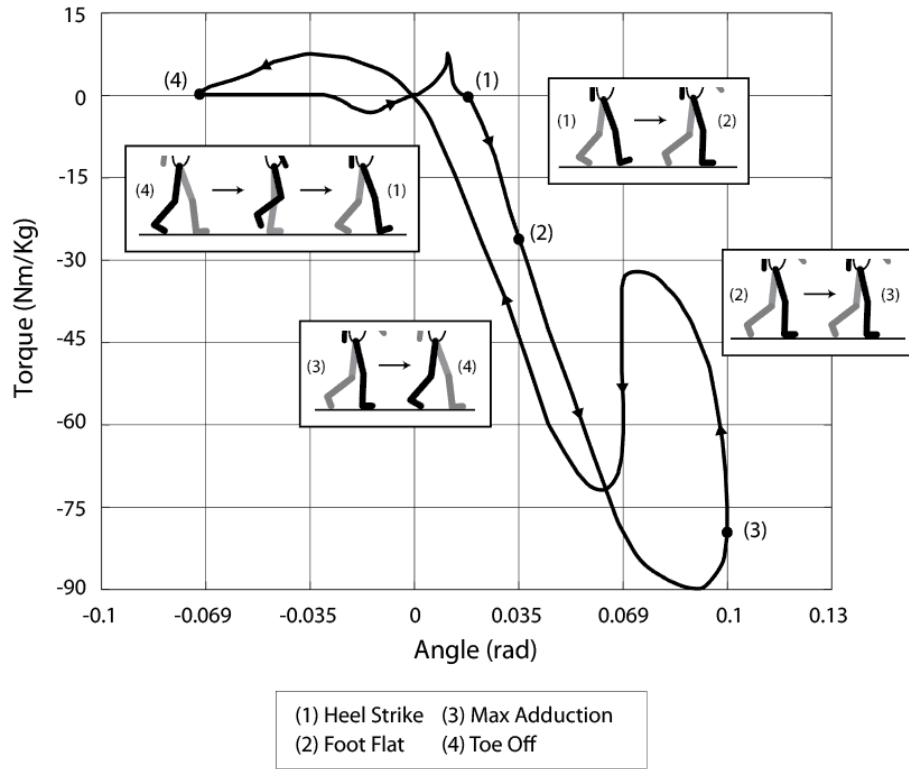


Figure 2.9: Frontal plane hip torque-angle plot with data extracted from [33]. Main events regarding the gait and the frontal angles are highlighted.

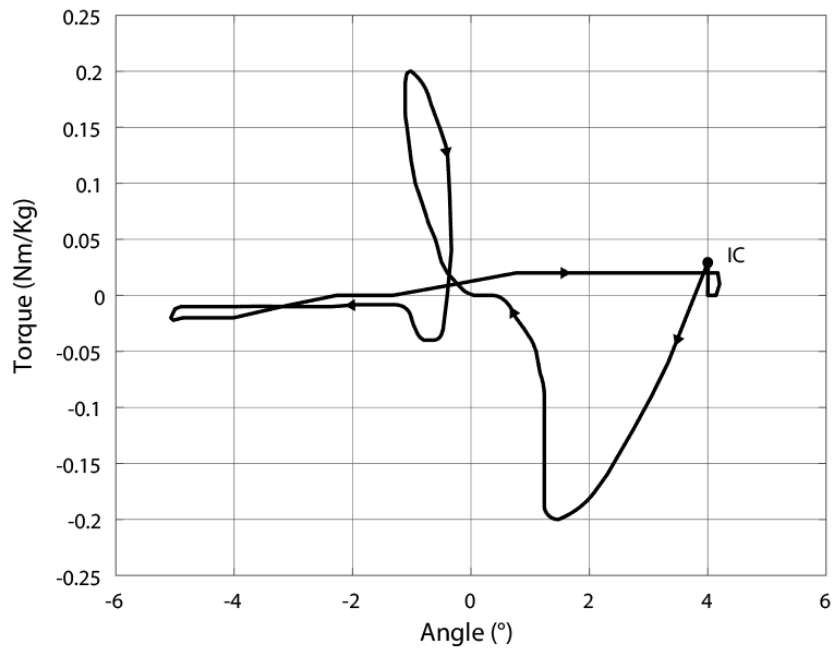


Figure 2.10: Transversal plane hip torque-angle with data extracted from [33]. Initial Contact (IC) is presented.

2.5.2 Walking at Different Speeds

The effects of walking speed on lower extremity biomechanics have been reported [34]. Sagittal plane joint angles and powers were measured and calculated for all three joints at slow (85 steps/min), natural (105 steps/min), and fast (122 steps/min) cadences. The point at which toe-off occurs in the cycle was found to be unaffected by change in walking speed, occurring at approximately 63% of the gait cycle for all three cadences. Although the joint angles for all three joints are seen to vary slightly with change in walking speed, these differences are considered to be very small when compared to the range of motion of each joint (see Figure 2.11). With respect to joint power, the timing of transitions from power generation to absorption or absorption to generation was generally unaffected by change in walking speed; however, a high correlation between peak joint powers and walking velocity was observed.

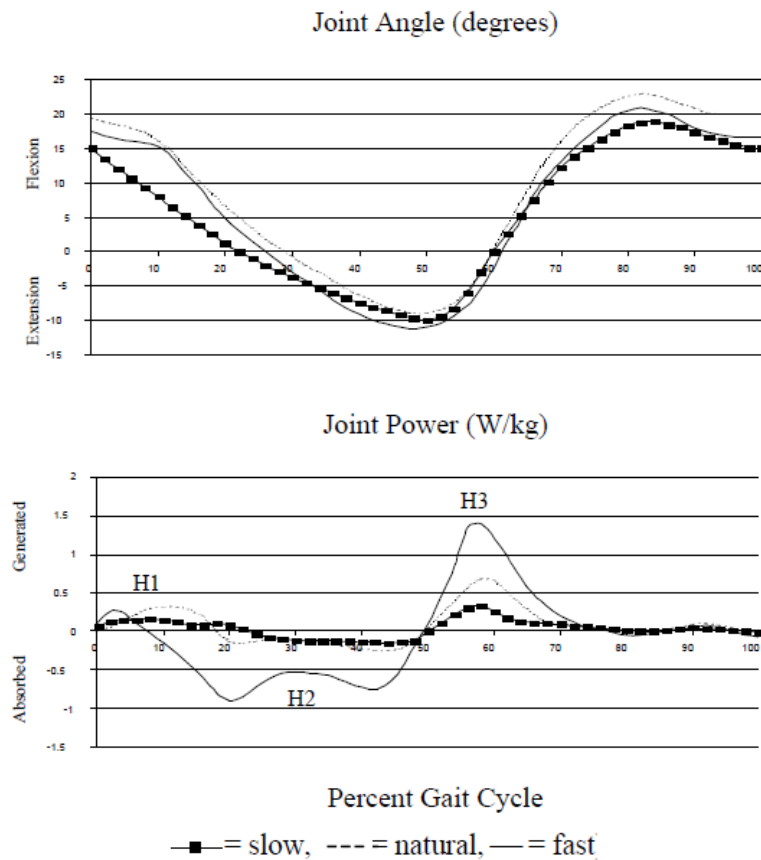


Figure 2.11: Sagittal plane hip angles and powers during one cycle of walking at three cadences. Image from [33].

2.5.3 Climbing and Descending Stairs

Finally, out of the activities involving locomotion that disarticulated amputees face more commonly (discarding highly demanding activities as running, jumping, etc), stair climbing and descending is one worth analyzing biomechanically. A report in 1980 [37] presented the stance phase joint moments for all three planes of motion. In this study, subjects ascended and descended a three-step staircase, and kinematic data were collected for one stride between the first and third step. Joint moments were calculated during the stance phase for all three planes of motion at each joint, and these resulting data are shown in Figure 2.12. A summary of the peak joint moment values obtained for each plane of motion is presented in Table 2.3.

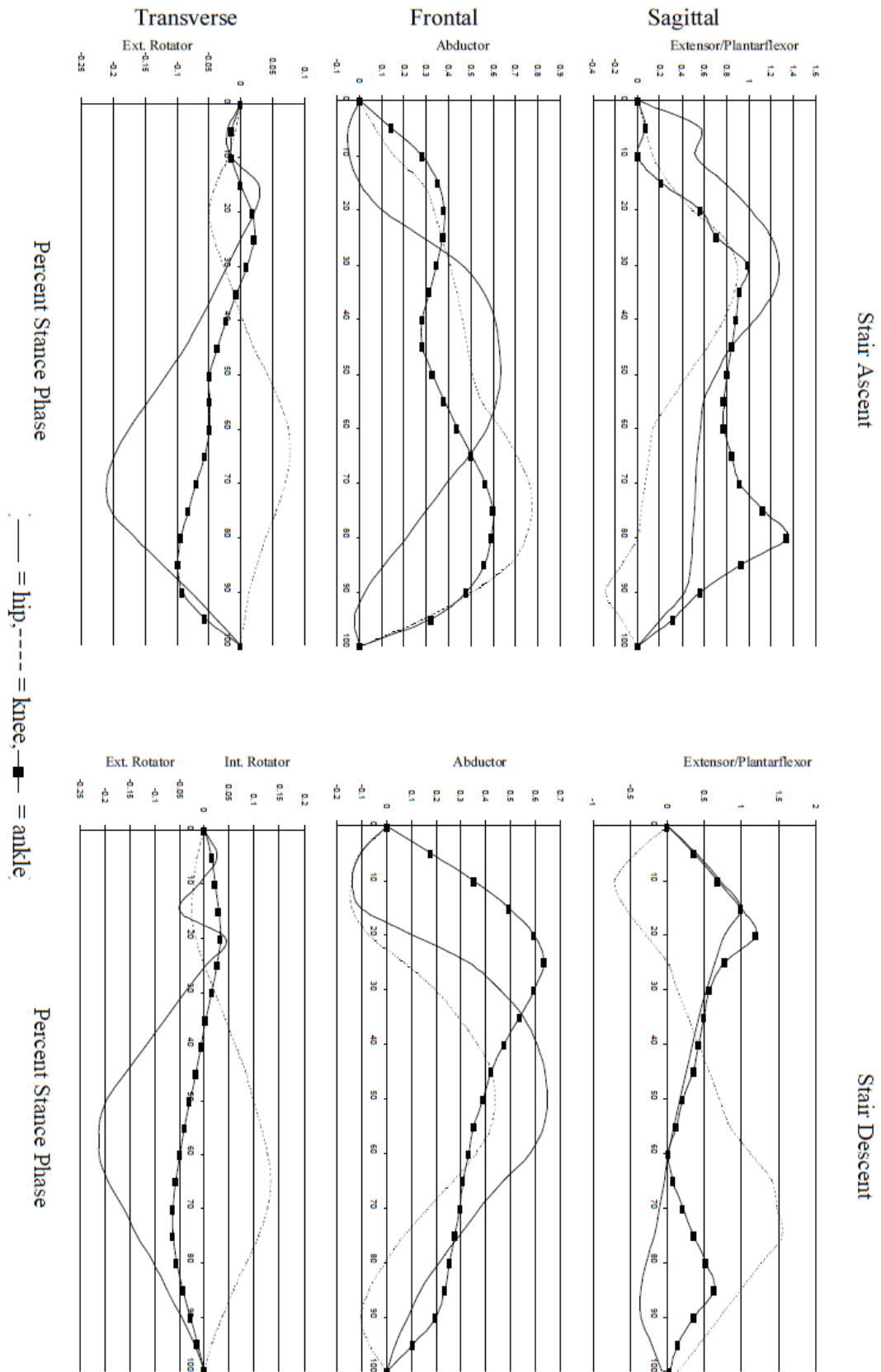


Figure 2.12: Joint moments in the sagittal, frontal, and transverse planes during stance phase of stair ascent and descent (values are in newton meters per kilogram). Image from [33].

During the stance phase of stair ascent, the largest joint moments appear to have occurred in the sagittal plane, but hip peak abductor moments in the frontal plane are also significant (approximately one-half of their corresponding peak extensor moments). Similar is the behavior

Table 2.3: Three-dimensional peak moment values during the stance phase of normal stair ascent and descent. Data from [33].

	Plane	Action	Value (Nm/Kg)
Ascent	Sagittal	Extensor	1.27
		Flexor	0
	Frontal	Abductor	0.63
		Adductor	0.05
	Transverse	Inversion	0.03
		Eversion	0.21
Descent	Sagittal	Extensor	0.99
		Flexor	0.35
	Frontal	Abductor	0.65
		Adductor	0.14
	Transverse	Inversion	0.04
		Eversion	0.21

observed in the stance of stair descent. Transverse moments during the stance phase of both stair ascent and descent appear to be considerably small.

Regarding joint powers another study analyzed the ones produced in the sagittal plane for normal stair ascent and descent [38]. Subjects also ascended and descended a three-step staircase, kinematic data were collected for each lower extremity joint, and joint powers were calculated with a six-degree-of-freedom approach. The results for the hip joint are shown in Figure 2.13. During stair ascent the peak power generation is lower than the corresponding values for normal walking (about 1.5 vs 1.8 ± 0.5 (W/kg)) but still in the confidence range. While during stair descent, peak power generation and absorption values are reduced to approximately one-third of their corresponding values for normal walking (0.65 vs 1.8 ± 0.5 (W/kg)).

Take-Home Messages

After revising the kinematic and kinetic profiles and the peak values corresponding to walking at “normal” and different speeds and climbing stairs, some conclusions towards the specification of a prosthetic module can be made:

- “Normal” walking peak values of torque and power in the sagittal and frontal planes are similar and considerably higher than the transverse values;
- During walking, the average power in the sagittal plane is positive (meaning to be generated by the joint) while in the other planes it is roughly neutral;
- The torque-angle diagrams of walking in the sagittal and frontal planes suggest a spring-like behavior that could be exploited in the design of the joint;
- There is a direct relation between the walking speed and the torque and power peak values. As the speed of walking increases, the maximum values increase. This is, of course, important to take into account given the aforementioned variations in magnitudes at the hip joint dynamics. Nevertheless, disarticulated amputees walk at a similar-low speeds (to be presented in Section 3.2) and peaks variation is rather small;
- Variations in the kinematics due to walking at different speed are considered minor when compared to the range of motion;
- Peak values of torque and power while climbing stairs are at most in the same range as the ones presented during “normal” walking. As a result, the values obtained when walking can be taken as inputs for the design of the prosthetic joint.

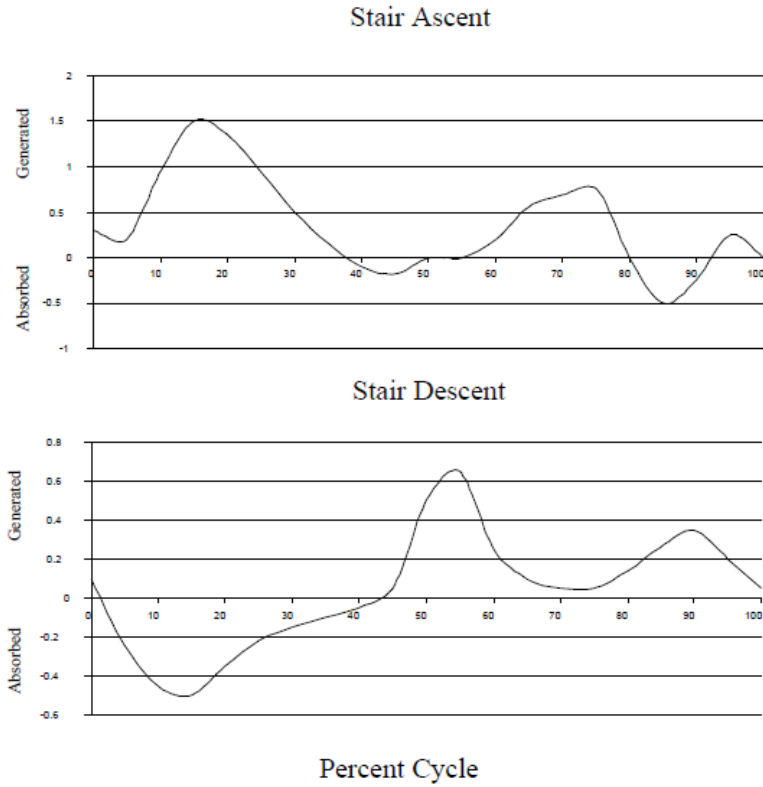


Figure 2.13: Sagittal plane hip powers during one cycle of stair ascent and descent (values are in watts per kilogram). Image from [33].

2.5.4 Compliance

One last concept highly important to analyze in locomotion is the one linked to the intrinsic visco-elastic properties of muscles that produce a compliant behavior at the joint level. Compliance, i.e. the flexibility of physical structures in response to an external force, is beneficial under different conditions and will be explored in this section based on the approach by Torricelli et al. [30]. Common advantages of compliance are the possibility of rapidly and efficiently storing and releasing energy (e.g. bouncing motion), or reacting instantaneously to sudden impacts, with positive effects on stability and thereby safety.

For the following analysis of the compliance in walking (see Figure 2.14), the biomechanical states of the hip–pelvis complex were classified into four categories, characterized by the actions of the internal forces produced by muscle activations from Zajac et al.[39]:

- Motive state, when the internal (muscle-tendon) forces are used to accelerate the joint motion, thus generating positive work. This is achieved in human joints by concentric activity of agonist muscles;
- Resistive state, when the internal forces are used to decelerate the joint motion against external forces or inertia, thus producing negative work. In humans, this is achieved by eccentric contraction of antagonist muscles;
- Stabilizing state, when the internal forces are used to counteract external forces in order to maintain the joint in a fixed position, producing low levels of work. This is achieved in humans by the isometric contraction of agonist and/or antagonist muscles;
- Passive state, when no internal forces are produced as a result of neural recruitment, leaving the joint free to move under the effect of inertial and gravitational forces as well as intrinsic muscle fiber, tendon, and ligament constraints. In this condition, muscles are not activated.

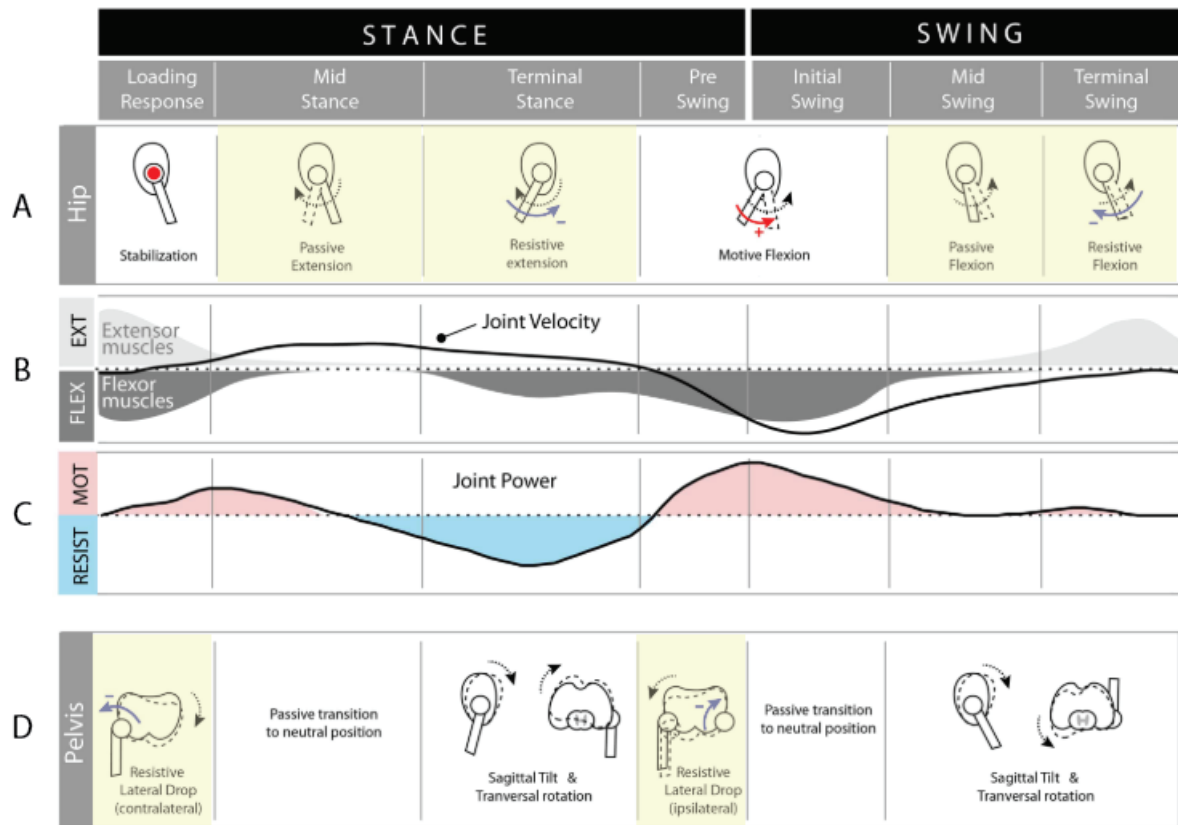


Figure 2.14: A. and D. Schematic view of the functional role of the hip and pelvis. Black dotted arrows show the direction of movement, while colored arrows indicate the direction of net joint torques (red, with a '+' symbol, for motive; blue, with a '-' symbol, for resistive torques). The red color indicates joint stabilization. B. Joint velocity profile and muscle activations of sagittal hip. C. Joint power, obtained by the product between net torque (not shown) and velocity. It does not take into account neither internal opposing forces, nor the forces generated in absence of motion. The plots for velocity, power and muscle activations are given in arbitrary units. Image adapted from [30].

During the loading response phase, a resistive action of hip abductors during the downward movement of approx. 4° , called hip drop, is used to dampen the impact of weight loading (see Figure 2.14.D). During the stance phase, the pelvis shows an anterior rotation of 4° . In order to facilitate the lower limb motion, the pelvis contributes with a transversal rotation of 10° enhancing the leg swing movements. Pelvic rotation and frontal plane tilt contribute to the control of the trajectory of the center of mass of the whole body. In the sagittal plane, the hip assumes a motive state during pre and initial swing (see Figures 2.14.A-B), when the hip flexors contract concentrically to pull the swinging limb upwards and forwards [34]. A resistive action is observed during terminal swing, when the hip extensors decelerate the limb forward momentum to prepare for ground contact. A passive role is prevalent during mid stance and midswing, in which the hip is passively extended under the gravitational and inertial forces, accelerating the thigh, with no relevant muscle contribution [40].

The hip assumes a stabilizing role during loading response, when the co-activation of extensors and flexors stiffen the joint to ensure postural stability during body weight transfer [40]. The hip frontal movement has been related to balance and posture stability during walking, which is obtained through the control of the lateral foot placement by hip abductor/adductor muscles during swing. This action is important to maintain the center of mass within the lateral borders of the base of support [34].

In addition to the schematic functions of human joints, highlighted in yellow are the phases where humans exhibit compliant behavior. This is mostly observed when the joints assume ‘resistive’ and ‘passive’ roles. In both cases, compliance of the joint is expressed with deviations of joint position under the action of external forces or inertia. In the ‘stabilizing’ state, instead, the joint acts to minimize such deviations, maintaining the joint in a constant position in presence of external forces. In the ‘motive’ state, compliance is not visibly expressed, because muscle activations are used to generate motive torques rather than reacting to an external force or inertia [30].

As seen throughout this manuscript, solutions for prosthetic hip joint have used stiff actuation only and do not consider the advantages of having compliant mechanisms at the hip–pelvis complex. Therefore, one may say that compliance is an unexplored characteristic worth studying given its remarked contributions to human locomotion.

2.6 Conclusions

This chapter presented the biomechanical concepts needed to understand the role of the hip joint in the movement and stability of the human body. In this context, the anatomical and functional characteristics of the hip were described taking into account the different tissues and their organization and influence. A precise analysis of the normal hip joint kinetics and kinematics while walking, as the fundamental activity of locomotion, was presented; focusing on the ranges of motion in all body planes (flexion/extension, abduction/adduction, and internal/external rotation), force couples and joint power. Variations in the walking speed and the activity of climbing stairs were also analyzed based on the previous parameters. Evidence showed that the sagittal DoF is the main DoF while walking, in terms of angle, torque and power profiles and peaks. Nonetheless, the importance of the secondary DoFs cannot be overlooked. In particular, the frontal DoF proved to be important as well based on the torque and power requirements. In addition, the inclusion of “compliant” behavior and related concepts becomes important to capture more accurately the biomechanics of the hip. The next chapter will present the biomechanical impact of hip disarticulation while walking, together with the first experiences with active prosthetic hip joints.

Chapter 3

Biomechanics of Hip Disarticulated Amputees and Active Hip Prosthesis

3.1 Introduction

As it was previously stated, lower-limb amputations affect the amputee's gait biomechanics in a different degrees depending on the level at which they are performed. Hip disarticulation, being the highest level amputation, has the greatest impact on the biomechanics of the amputees. In this chapter, the biomechanics of walking of hip disarticulated using the Helix3D prosthetic joint will be presented. Most of the times it will be done by comparing this prosthetic joint with pre-existing hip prostheses. Finally, the few active prototypes of prosthetic hip joints will be addressed.

3.2 Degradation of Locomotion Physiology

After having provided an extensive review of the hip biomechanics during walking in the previous chapter, this section will focus on the degradations of this healthy behavior for disarticulated patients wearing the prosthetic hip joint 7E10 Helix3D by Ottobock [16], considered to be the actual gold-standard. The advantages of such a device will be presented and the effects it has on the hip disarticulated individuals will be analyzed by examining the few real cases available. After a broad literature review, where the focus lay on “studies measuring functional outcomes with the Helix 3D hip joint prosthesis for individuals with a hip disarticulation amputation”, a total of four articles was found. The authors of three of them are affiliated with Ottobock which may represent a conflict of interest in the way outcomes were measured. However, the results that will be presented are important to build the foundations of future designs and proper biomechanical performances.

The New Helix 3D Hip Joint - Initial Results and Experiences with the Hip Joint (2008) [17] briefly summarizes the experiences and initial results for the described hip joint and trial or test fittings for 18 hip disarticulation amputees and three hemipelvectomy amputees (an extended version was released on a Thesis report available only in German). The study reported that movement characteristics typical for conventional hip joints (7E7 hip joint by Ottobock) changed significantly in the sagittal plane. Hip joint extension, from the beginning of the stance phase until the extension stop is reached, took longer than just the 15 % of the gait cycle achieved by the 7E7. It is marked that, since the Helix hydraulics dampens extension, the abrupt stop was eliminated, which then also helped to optimize the stance phase resistance and the step length. Finally, amputees described the effects of the combined movement (due to

a polycentric configuration; flexion and rotation) during the stance phase while walking slowly as especially advantageous. However, the report remains a first qualitative-approach.

Biomechanical differences between two exoprosthetic hip joint systems during level walking (2010) [18] analyzed the gait pattern of six hip disarticulation amputee subjects again by comparing the Helix 3D with the 7E7 joint. Kinematics and kinetics were recorded by an optoelectronic camera system including force plates. The differences between both joints in regards to the walking speed and prosthetic and contralateral step length were insignificant. Regarding the pattern of the hip joint moments, angles and angular velocity were compared (see Figure 3.1).

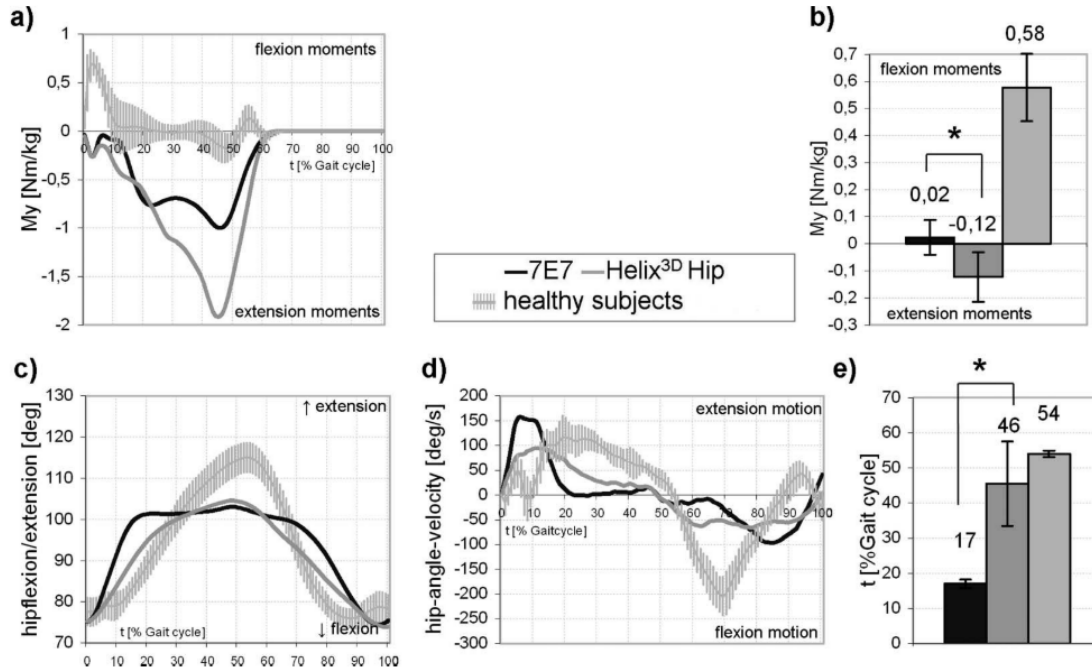


Figure 3.1: A. Mean pattern of sagittal hip moments. B. Mean maximum sagittal hip moments within the first 20% of the gait cycle. C. Mean pattern of hip flexion/extension. D. Mean pattern of hip angle velocity. E. Mean time of maximum hip extension. *significant, $p < 0.05$. Image from [18].

With all subjects using the Helix3D, the external hip moments produced during stance were extension moments, while when walking with the 7E7, a short flexion moment within the first 20% of the gait cycle was observed with five subjects. Both of them were far from the moments produced by healthy subjects. The pattern of angle and angular velocity for both hip joints showed differences. With the 7E7, maximum extension was reached at 17% of the gait cycle, meanwhile with the Helix3D it was reached at 46% of the gait cycle; probably due to its adjustable resistance. This effect led to the conclusion that the hip joint was continuously moving during single limb support on the prosthetic side and was more in line with the hip motion of the contralateral limb. Something that the subjects described positively. Furthermore, until flexion was initiated after 70% of the gait cycle, the hip angle of the 7E7 changed only slightly. With the Helix3D, however, flexion started again immediately after maximum extension and was relatively closer to the behavior observed in healthy subjects. Without digging extensively in the angular velocity pattern, significant differences in peaks and slopes were observed for the prosthetic joints and the healthy subject as well. In general terms, the pattern of hip angular motion and angular velocity with the Helix3D was, when compared to other systems, qualitatively more similar to the hip motion observed in non-amputees. Additionally, it was

found that the stance phase damping of the Helix3D allowed the amputee to extend the hip joint more slowly and to control the pelvic posterior tilt.

In this case, even though the patterns are positively affected by the joint, the fact that no flexion moments (due to the lack of actuation) are achieved is still a major challenge towards a more healthy-gait. This is something also related to the range of motion that this joint can offer, being around three-fourths of the peak value expected in a normal gait cycle.

In Functional Outcome Measurements of a Veteran With a Hip Disarticulation Using a Helix 3D Hip Joint: A Case Report (2011) [41] the authors reported the results of the Timed Up and Go (TUG) and the 2-minute walk tests performed by a 30-year-old male with a traumatic hip disarticulation amputation. Tests were conducted at the time of delivery of the permanent Helix 3D hip joint prosthesis, and then again after eight and at 15 weeks. The tests were also performed with the 7E7 joint but only at the 15th week. Although no function outcome measures have been validated specifically for persons with hip disarticulation amputations, the TUG and the 2-minute walk tests were chosen as they are simple to perform in a clinic setting and are both well studied.

The subject had a progressively improving TUG test time during the 15-week time period while using the Helix3D prosthesis. In addition, the standard deviation decreased in week 15, indicating he became more consistent in performing the TUG test. However, regarding the 2-minute walk test, he did not show similar improvements in this measure over the time of his 15 weeks of rehabilitation. In order to explain this phenomenon, the authors justified that the TUG has a component of transferring from sit to stand, and stand to sit in addition to ambulation. Thus, the subject may have shown improvement as he gained experience in transferring with the prosthesis with physical therapy and home use. It was also found that he ambulated at an average speed of 1.06 m/s using the Helix3D, which is slower than the average speed for his age and sex but is faster than reported speeds for hip disarticulation amputees using conventional hip joints. According to [11] the average self-selected walking speed for individuals with hip disarticulation amputations secondary to malignant tumor resection is 0.79 m/s.

The walking speed with the 7E7 prosthesis was similar to those reported for a demographically similar veteran ambulating with a comparable prosthesis. At three months post-injury, a walking speed of 0.57 m/s, and at 38 months post-injury, a walking speed of 0.86 m/s in a 27-year-old soldier status post-traumatic hip disarticulation, ambulating on a single-axis hip joint with an Ottobock C-Leg was previously reported [42]. The difference in walking speed suggested a potential difference in functional independence based on the prosthetic joint.

In this case report, valuable information regarding the speed of ambulation and real clinical test results with the prosthetic joint was obtained. However, more detailed measurements regarding gait symmetry through kinematic analysis and energy expenditure could complement the study; which is something acknowledged by the authors and planned as future studies.

The last one, **Evaluation of the Benefits of a New Prosthetic Hip Joint System in Activities of Daily Function in Patients after Hip Disarticulation or Hemipelvec-tomy (2013)** [43] is the more complete study conducted with the Helix3D to the knowledge of the author. This study included a survey and mobility tests of thirteen subjects with hip disarticulation before and after fitting a new prosthetic hip joint system.

The clinical test performed included (1) the 10-m walking test, (2) downstairs 10 steps, (3) walking downhill 7-m and (4) uphill 7-m. Mobility and independence were assessed by the implementation of the Locomotor Capabilities Index-5 (LCI-5) and a functional assessment specific for the use of a hip prosthesis (see Figure 3.2). The LCI-5 is a highly validated instrument that is well correlated with performance-based outcome measures for amputee rehabilitation such as the 2-minute walk test and daily step activity.

The common question is "whether or not you wear your prosthesis at the present time would you say that you are able to do the following activities with your prosthesis on?"				old system	Helix ^{3D} Hip Joint System	p- value	Box-Whisker-Diagram point = Median box = IQR line = range	
							old system	Helix
basic ambulation skills	1	Get up from a chair	Median Min Max	4 (4) 3 (3) 4 (4)	4 (4) 3 (3) 4 (4)	0.317 n.s.		
	4	Walk in the house	Median Min Max	4 (4) 3 (3) 4 (4)	4 (4) 2 (2) 4 (4)	0.593 n.s.		
	5	Walk outside on even ground	Median Min Max	4 (4) 3 (3) 4 (4)	4 (4) 3 (3) 4 (4)	0.317 n.s.		
	8	Go up the stairs with a hand-rail	Median Min Max	4 (4) 3 (3) 4 (4)	4 (4) 3 (3) 4 (4)	0.180 n.s.		
	9	Go down the stairs with a hand-rail	Median Min Max	4 (4) 1 (3) 4 (4)	4 (4) 3 (3) 4 (4)	0.655 n.s.		
	10	Step up a sidewalk curb	Median Min Max	4 (4) 3 (3) 4 (4)	4 (4) 3 (3) 4 (4)	1.000 n.s.		
	11	Step down a sidewalk curb	Median Min Max	4 (4) 3 (3) 4 (4)	4 (4) 3 (3) 4 (4)	1.000 n.s.		
total score of the basic ambulation skills			Median Min Max	28 (28) 21 (21) 28 (28)	28 (28) 20 (20) 28 (28)	0.500 n.s.		
advanced ambulation skills	2	Pick up an object from the floor when you are standing up with your prosthesis	Median Min Max	4 (4) 0 (3) 4 (4)	4 (4) 3 (3) 4 (4)	0.285 n.s.		
	3	Get up from the floor (eg, if you fell)	Median Min Max	4 (4) 1 (1) 4 (4)	4 (4) 2 (2) 4 (4)	0.317 n.s.		
	6	Walk outside on uneven ground (eg. Grass, gravel, slope)	Median Min Max	3 (4) 3 (3) 4 (4)	4 (4) 3 (3) 4 (4)	0.068 n.s.		
	7	Walk outside in inclement weather (eg.snow, rain, ice)	Median Min Max	3 (3) 0 (3) 4 (4)	4 (4) 3 (3) 4 (4)	0.028 *		
	12	Go up a few steps (stairs) without a hand-rail	Median Min Max	3 (3) 0 4	4 (4) 2 4	0.018 *		
	13	Go down a few steps (stairs) without a hand-rail	Median Min Max	3 (3) 0 (0) 4 (4)	4 (4) 2 (2) 4 (4)	0.028 *		
	14	Walk while carrying an object	Median Min Max	4 (4) 0 (0) 4 (4)	4 (4) 3 (3) 4 (4)	0.180 n.s.		
total score of the advanced ambulation skills			Median Min Max	22 (25) 10 (10) 28 (28)	27 (28) 18 (18) 28 (28)	0.005 **		
total score of the LCI-5			Median Min Max	46 (53) 31 (31) 56 (56)	55 (56) 38 (38) 56 (56)	0.003 **		
ambulation skills explicitly difficult for HD and HP	N1	Sit down on a chair with weight on both legs	Median Min Max	0 (4) 0 (4) 4 (4)	4 (4) 0 (0) 4 (4)	0.037 *		
	N2	Step up on a sidewalk curb without stopping first	Median Min Max	0 (0) 0 (0) 4 (4)	4 (4) 0 (0) 4 (4)	0.018 *		
	N3	Step down a sidewalk curb without stopping first	Median Min Max	3 (3) 0 (0) 4 (4)	4 (4) 0 (0) 4 (4)	0.128 n.s.		
	N4	Walk in the house without looking down at the ground	Median Min Max	4 (4) 0 (0) 4 (4)	4 (4) 3 (3) 4 (4)	0.109 n.s.		
	N5	Walk outside on even ground without looking down at the ground	Median Min Max	4 (4) 0 (0) 4 (4)	4 (4) 1 (3) 4 (4)	0.281 n.s.		
	N6	Walk outside on uneven ground without looking down at the ground	Median Min Max	3 (3) 0 (0) 4 (4)	4 (4) 0 (0) 4 (4)	0.028 *		
total score of the supplementary ambulation skills			Median Min Max	15 (12) 4 (7) 24 (24)	24 (24) 6 (6) 24 (24)	0.005 **		

In parentheses are the values of the three subjects who had already been using a C-Leg previously.
LCI-5, Locomotor Capabilities Index-5; HD, hip disarticulation; HP, hemipelvectomy; NS, not significant; *significant with 0.05 > p > 0.01; **highly significant with p < 0.01.

Figure 3.2: Results of the Locomotor Capabilities Index-5 (LCI-5), basic ambulation skills, advanced ambulation skills, and ambulation skills explicitly difficult for hip disarticulation patients. Image from [43].

The LCI-5 was used as the primary outcome measure in this study. According to this index, mobility and independence of lower-limb amputees are determined based on 14 activities of daily living. The subjects were asked to rate how independently they were able to perform each activity and points were awarded as follows:

- 0 = not possible
- 1 = yes, if someone helps me
- 2 = yes, if someone is near me
- 3 = yes, alone, with ambulation aids
- 4 = yes, alone, without ambulation aids

Therefore, the maximum possible score was 56; and the higher the score, the higher the amputee's mobility and independence in everyday life. Additionally, another questionnaire containing six activities (N1YN6) that are considered very demanding for hip disarticulation amputees was developed. With the same scaling of the LCI-5 a maximum score of 24 can be reached. Furthermore, the functional assessments specific for the use of a hip prosthesis evaluated: (1) comfort while standing, (2) safety while standing, (3) safety during loading response of the prosthesis, (4) ease of initiation of swing phase, (5) extension characteristics of the hip joint ending at the extension stop, (6) walking speed and (7) start to walk from a standing position. These aspects might be influenced by the known characteristics of the Helix3D and somehow represent a conflict of interest. The mobility levels were determined by subjective judgment using the definition of the Medicare Functional Classification Levels.

The basic ambulation skills of the LCI-5 were performed almost equally well with either hip prostheses as they consist of activities that are typically done by patients with lower mobility levels similar to those enrolled in this study. The results for the advanced ambulation skills and ambulation skills that were considered explicitly difficult for this amputee group, however, showed a significant improvement with the Helix3D. The following are the activities in which benefits were perceived, according to the feature of the hip joint to which they were attributed:

- The hydraulically controlled swing phase damping provided for a constant step length of the prosthetic leg, which has been proven highly effective for amputees, especially when walking on different terrains (LCI activity 6) and on black ice (LCI activity 7);
- The PU elements caused the hip to flex when weight was taken off, resulting in the prosthesis swinging forward. This supported dragging it to the next step and, therefore, facilitate the swing phase initiation on uneven ground, while minimizing necessary visual control (activity N6). This elements may also be a reason for improved performance of the 12th LCI activity: walking upstairs without using the handrail;
- Another special activity, stepping up a sidewalk curb without stopping first (activity N2), was performed with less aids, which may be attributed to higher protection against hip flexion at loading response that was provided by the polycentric movement of the joint;
- The median values found for the Helix3D Hip Joint System and the previous prosthesis were identical in the other 12 of the 20 activities (1,4,5,8,9,10,11,2,3,14,N4 and N5);
- With respect to the specific functional outcome measures, the Helix3D hip joint system was rated significantly better than the previous prosthesis (as it was expected).

The overall results of the LCI-5 demonstrated a statistically significant improvement in walking ability, mobility, and independence with the amputees using the Helix3D hip joint system as compared with their previous prosthesis. Additionally, a reduction in time required for walking down a defined ramp or stair with the Helix3D was found. On the other hand, a significant increase in walking velocity on level ground due to the Helix3D as described in the previously presented case reports [41], [42] was not confirmed.

Among the limitations of this study is the fact that the subjects were aware beforehand of the claimed benefits of the Helix3D Hip Joint and the C-Leg. Furthermore, in most cases, the

whole prosthesis was changed, making it difficult to determine the proportion of effects of every component for the demonstrated improvements in function.

Take-Home Messages

The previous analysis of the results obtained with the implementation of the Helix3D joint in disarticulated amputees constitutes the baseline for the new designs. The main outcomes to keep in mind can be summarized as:

- The polycentric “3D” configuration of the Helix3D joint has been reported as advantageous by its users, in terms of comfort and stability;
- Passive joints (either the Helix3D or the 7E7) are incapable of reproducing the hip moments of healthy patients. They generate extension rather than flexion torques, which as previously seen, is the opposite of a healthy pattern;
- The hip joint sagittal angle profile is better represented by the Helix3D joint than other hip joints, even though there is still a difference in magnitude with stand values;
- A patient using the Helix3D joint showed considerable improvements in the transferring from sitting to standing positions (TUG test) over 3 months of rehabilitation therapy. His ambulating speed was reported to be higher in comparison with speeds of individuals using conventional hip prosthetic joints; which suggests a potential difference in terms of independence due to the joint;
- For the first time, a validated and suited questionnaire/survey was developed to functionally assess hip disarticulation: The LCI-5 with the supplementary ambulation skills, containing activities that are considered demanding for the patients;
- With this assessment, no differences were found for the basic ambulation skills when comparing the Helix3D with conventional joints. However, for the activities explicitly demanding for disarticulated amputees, significant improvements were found.

3.3 Hip Active Prosthesis

The repercussions of hip disarticulation in the locomotion are, at this point, clear to the readers. And, even though the Helix3D has proven to improve many functional skills of disarticulated amputees, there seems to be a roof for advances in passive joints. In this section, the available active technologies for hip disarticulation will be explored.

In studies performed in other fields (non-amputees centered), assistance of the hip joint has shown to improve gait in amputees as assisted hip motion was found to not only improve spatiotemporal gait parameters, i.e. velocity, cadence, step time, stance time, swing time, double support time, stride length and spatial asymmetry in stroke survivors [44], but also to reduce the metabolic cost during walking in healthy subjects [45]. Additionally, the effects of hip torque during step-to-step transition during human walking has been examined with numerical simulation. The models demonstrated that the hip torque actuator can improve the energetics and stability of walking; which led the authors to claim that the findings may be helpful for the improvement of the bipedal robots, prosthesis and exoskeleton [45], [46].

As stated in Section 1.2.3, the main purpose of active (robotic) prostheses is to provide active actuation during gait to improve the biomechanics and metabolic costs for amputees [13]. To the knowledge of the author only a few developments of active hip prostheses have been reported. The first contribution by Hata and colleagues [47], is in Japanese and there is no translation available of the manuscript. The second one corresponds to a patent application for a powered hip joint submitted in 2018 by the Icelandic company Ossur [48]. The patent pertains

to a mechanism that connects the powered prosthetic thigh to the socket. However, the design specifics and device performance are unknown because there are currently no detailed reports.

The last one belongs to Ueyama and colleagues [49], who developed a **robotic hip disarticulation prosthesis (HDP)**, and analyzed the gaits of a non-amputee using the robotic and non-powered dummy prosthesis composed of standard prosthetic parts. Both the robotic and the non-powered prosthesis were attached to dummy sockets to allow the non-amputee to wear them, as presented in Figure 3.3. The robotic device was composed of two DC motors and motor drivers at the hip and knee joints through pulleys and timing belts (i.e. there was one degree of freedom at each hip and knee joint).

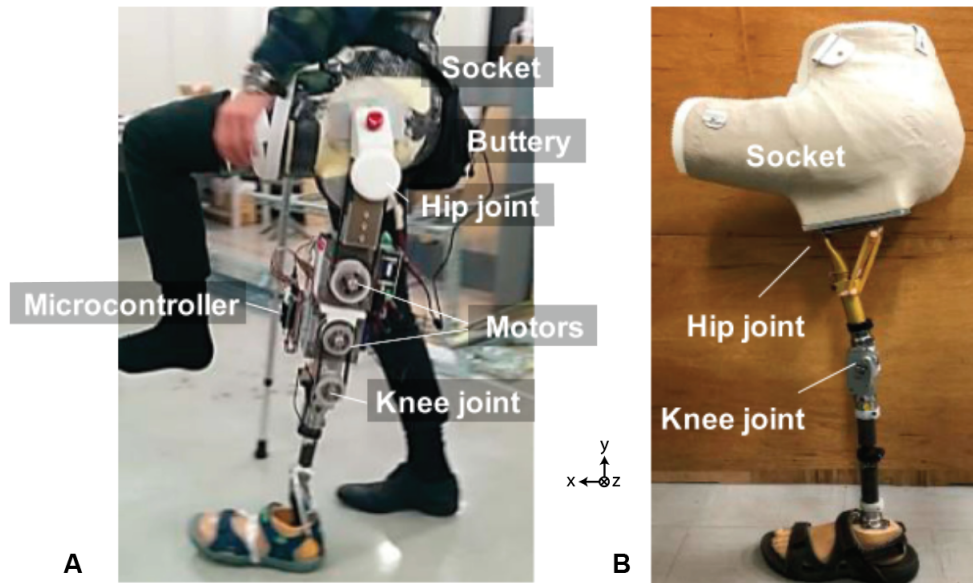


Figure 3.3: Robotic and non-powered hip-disarticulation prosthesis with dummy sockets. A. Robotic HDP substituting for the left leg. The robotic HDP has hip and knee joints powered by two motors. B. Non-powered HDP. The hip and knee joints are not powered, but can be moved passively. Image from [49].

The subject trained in walking using the prosthetic for 1 h/day over 10 days in a month using the non-powered HDP, as for most healthy people walking using an HDP without training is a difficult task. On the last training day, the subject practised with the robotic HDP for only about 10 min. The prosthesis was adjusted to the subject to coordinate several settings. Among them, the length of the lower leg, the fine positioning and rotation of the joints, and the damping strength of the knee joint.

Regarding the experiment using the non-powered HDP, the step lengths of both limbs were found to be extremely small and the subject leaned forward while walking. In the prosthetic limb only small changes in the hip and knee joints were observed (see Figure 3.4.A, left). In the non-prosthetic limb changes were larger than those, but the hip joint was extended almost without flexion (see Figure 3.4.A, right). These characteristics match what was mentioned in the previous section, with respect to the impossibility of passive devices to reproduce healthy behaviors. On the other hand, with the robotic HDP there was no trunk lean by the subject. Although the step lengths of the non-prosthetic limb were longer than those of the prosthetic limb; the joint angles were almost equivalent to the reference joint angles for non-amputees.

By using a dynamic optimisation software for a human musculoskeletal model (BoB; BoB-Biomechanics, Coventry, UK) ground reaction forces and hip contact force (HCF) were estimated. In the case of the non-powered HDP, the non-prosthetic limb induced greater HCF than the prosthetic limb and the difference was significant (see Figure 3.4.B and C).

For the robotic HDP, the non-prosthetic limb also induced slightly greater HCF than the prosthetic limb. However, the difference was not significant this time.

These results, the authors claimed, imply that a robotic prosthesis may induce proper gait and maintain the burdens of the prosthetic and non-prosthetic limbs symmetrically and present potential advantages of the robotic prosthesis to improve mobility of hip-disarticulated. Nevertheless, they acknowledged that the robotic HDP was much heavier than a standard HDP, which could imply that fatigue would not be reduced. Thus, the prosthesis needs to be re-designed to reduce its weight.

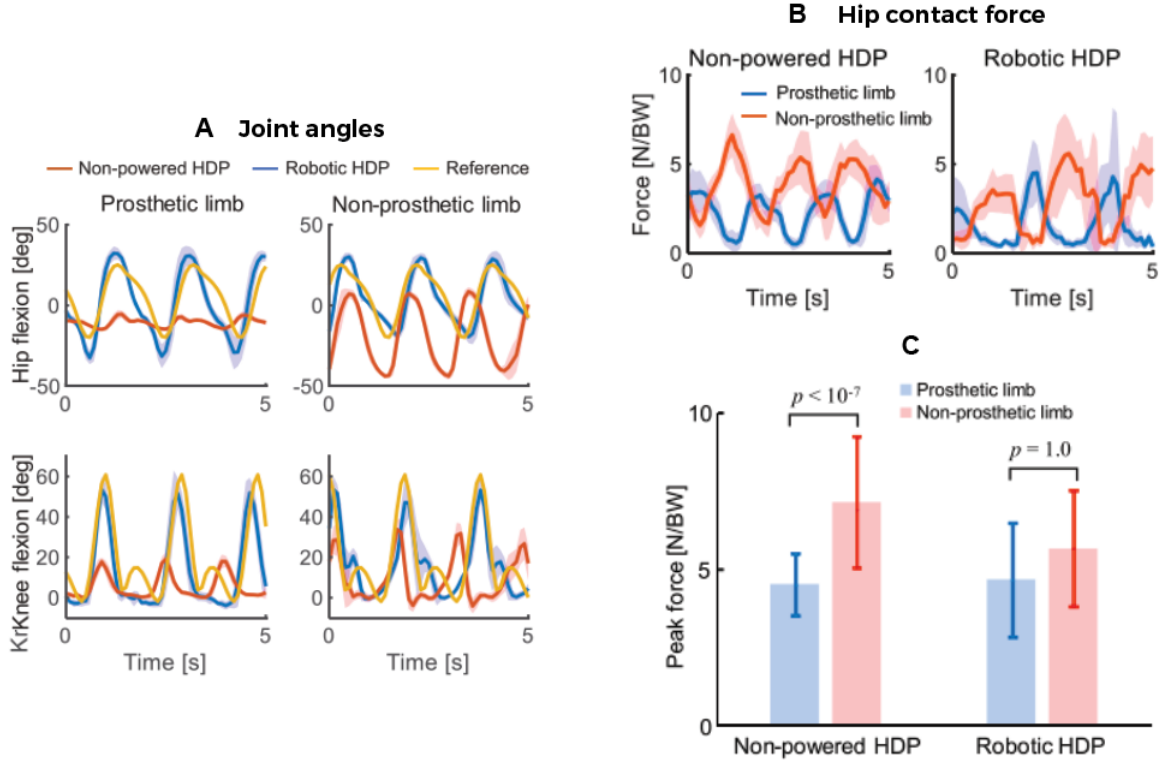


Figure 3.4: A. Time profiles of the hip joint and knee joint flexion angles of the prosthetic and non-prosthetic limbs. The positive angles are defined as flex direction of the joints “Reference” indicates standard joint angles for non-amputees without the prosthesis. B. Time profiles of the hip contact forces. C. Mean peak of the hip contact forces. p-value of the comparison between the prosthetic and non-prosthetic limbs by the unpaired two-tailed Welch’s t-test. Image adapted from [49].

3.4 Conclusions

This chapter presented the biomechanical impact of hip disarticulation in amputees using the Helix3D prosthetic joint, which is considered to be the best prosthetic hip joint currently available. The advantages of its implementation were noticeable when patients were performing activities explicitly demanding for disarticulated amputees. However, most of the characteristics of the prosthetic hip joint having been reported as beneficial by the users still lack scientific quantification. Additionally, the few active prototypes of prosthetic hip joints to date were presented, along with the evidence supporting the need of actuation in the prosthetic hip joint to correctly replace the human leg. The next chapter will present the design processes that followed this analysis towards an active hip module.

Chapter 4

Mechanical Design: Methodology and Results

4.1 Introduction

Throughout history there have been many attempts to draw up maps or models for designing successful different type of products. Some of these models simply describe the sequence of activities that typically occur in a design while others attempt to prescribe a better or more appropriate pattern of activities. These latter models try to persuade or encourage designers to adopt improved ways of working. They usually offer a more systematic procedure to follow, and are often regarded as providing a particular design methodology. The Cross model [50], which is the design model followed in this project, is based on guidelines written by the german professional engineers' society, Verein Deutscher Ingenieure (VDI). These VDI principles considered the breaking of an overall problem into sub-problems to then find suitable sub-solutions and combine them into an overall solution; and was based on seven stages, each with a particular output. The Cross model contemplates, in addition to the previous logical progressions, the symmetrical and commutative relationship between problem and solution and sub-problems and sub-solutions, as illustrated in Figure 4.1. Taking these interconnections into account, the model attempts to capture the essential nature of the design process, in which understanding the problem and solution are developed together. The way how the seven stages relate to each other and to the symmetrical problem-solution model is also presented in the Figure. For example, clarifying objectives is appropriate both to understand the problem-solution relationship as well as to develop the overall problem into sub-problems.

On top of the general approach, the design methods (any procedures, techniques or tools for designing) have also evolved and multiplied over time. There is a wide range of rational design methods covering all aspects of the design process from problem clarification to detail design. The Cross model includes into the process the ones considered to be the most relevant and widely-used methods, covering the whole design process. The selected set is detailed in Figure 4.2, with the stage to which it belongs in the design process shown on the left, and the method relevant to this stage shown on the right.

The design process carried on in this project followed the order of the stages and was implemented as close as possible to the methods suggested by the Cross model. The development of each of them is presented in the next sections in accordance, naturally, with the outcomes found in the previous chapter.

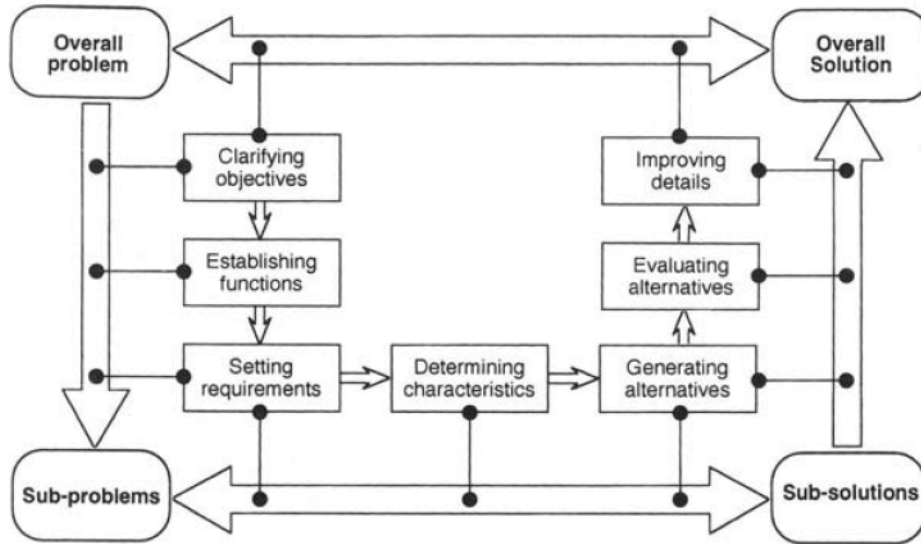


Figure 4.1: Seven stages of the design process within the symmetrical problem/solution model. Image from [50].

Stage in the design process	Method relevant to this stage
Clarifying objectives	Objective Tree Aim: To clarify design objectives and sub-objectives, and the relationships between them
Establishing functions	Function Analysis Aim: To establish the functions required, and the system boundary, of a new design
Setting requirements	Performance Specification Aim: To make an accurate specification of the performance required of a design solution
Determining characteristics	Quality Function Deployment Aim: To set targets to be achieved for the engineering characteristics of a product, such that they satisfy customer requirements
Generating alternatives	Morphological Chart Aim: To generate the complete range of alternative design solutions for a product, and hence to widen the search for potential new solutions
Evaluating alternatives	Weighted Objectives Aim: To compare the utility values of alternative design proposals, on the basis of performance against differentially weighted objectives
Improving details	Value Engineering Aim: To increase or maintain the value of a product to its purchaser whilst reducing its cost to its producer

Figure 4.2: Selection of the most relevant and widely-used design methods. Image from [50].

4.2 Objectives Clarification

Even after the overview presented in the previous chapters, the clear statement of the objectives and the details surrounding is crucial to start the design process. The method chosen to do it is the objectives tree. It shows in a diagrammatic form the ways in which different objectives are related to each other, as well as, the hierarchical pattern of objectives and sub-objectives. The procedure of conceptualizing an objectives tree helps to clarify the objectives and to reach agreement between different members of the research team.

Under normal circumstances this team would include the designers, medical staff and patients related to the subject; however, *the special health situation developed at the beginning of the year 2020* restricted the integration of the last two groups of participants in the project. Therefore, the establishment of the objectives and the following methods were based on testimonies by some disarticulated amputees [51], [52], video material uploaded by some of them showing their experiences [53]–[56] and the literature publications presented in the last chapter. In particular elements from the study by Ludwigs et al. [43], in which the LCI-5 was complemented with a set of activities considered to be demanding for hip disarticulation amputees (see Section 3.2 for further explanation).

Before starting with the objectives tree one last brief-review of the current stages in the procedure of construction of prostheses and the limitations of hip joint prostheses is valuable to understand matching or disruptive considerations of the design. According to Rajt’uková et al. [28] some of the important factors for the creation of properly functioning prostheses include: (1) The selection of appropriate components that depends primarily on the physical and mental condition of the user, user’s activity, and method of use. The principal factors for the selection of prosthetic components are a patient’s weight and physical activity. Depending on the user’s weight, the material of prosthetic parts is selected so that sufficient strength and average weight of the prosthesis is ensured. In this project it is clear that to the design process, all of these factors apply. (2) The residual limb’s conditions, amputation height, residual limb’s shape (conic-smaller circumference on the distal part than on the proximal part, pear-shaped, cylindrical-the same circumference on the distal and proximal end), amputation scar, as well as other problems or diseases. In this case, amputation height is of course taken into account; however, the details of the scar and disease-related features are not studied as they correspond to the interaction with the socket rather than directly with the joint. (3) The construction of the prosthesis that can be divided into the following steps: basic construction, static and dynamic correction. The overall aim of this project lies in the preliminary design.

Coming back to the objectives clarification method, one of the best ways to organize the objectives tree is by answering to the following questions: **1. What?**, **2. Who?**, **3. How?**, **4. When?**, **5. Where?**, and possibly, **6. Recycling?**. After a meticulous analysis the objectives tree of the process is presented in Figure 4.3, following the same enumeration of the questions. To address it, each branch was also color-coded.

Given the context and information provided so far, the development of the leaves number **1. What** (in light red) and **5. Where** (in purple) are rather explicit at this point. Therefore, let’s start with the branch number **2. Who**, colored in yellow. When one discusses about a (electro)mechanical device there is a set of basic elements associated to its physical embodiment and configuration. For the hip joint four major attributes were found. The first one is the weight, which as presented in Section 1.2.2, is one of the factors increasing energy expenditure in patients using passive prostheses. Although in this active approach it might be compensated by the actuation in terms of burden to the patient, is it definitively an important feature to consider, as mentioned at the end of Section 3.3. The second one is the volume or space to size up when assembling all the components in a fashion suitable for the patient, i.e as symmetrical as possible to the other lower limb.

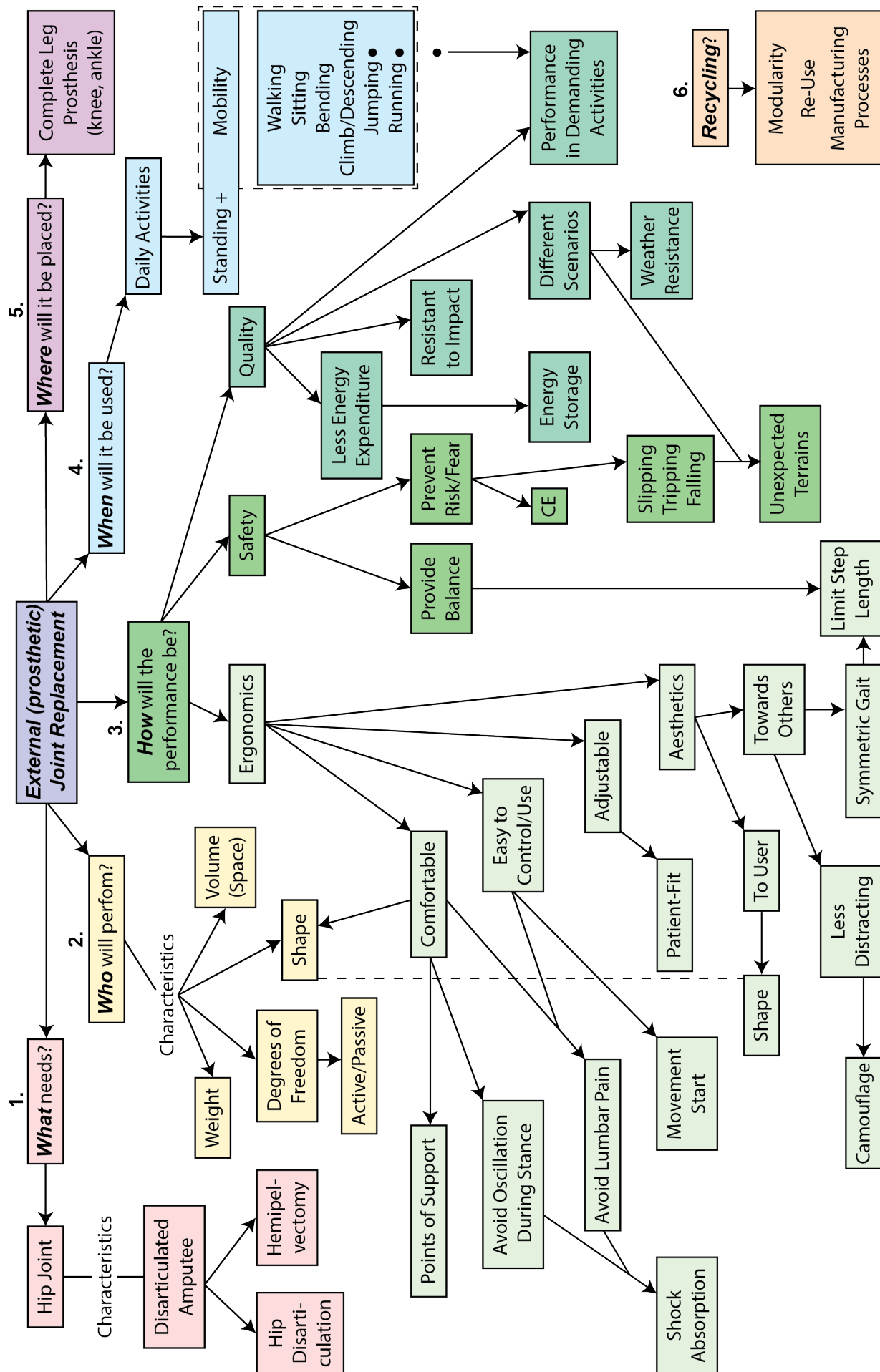


Figure 4.3: Objectives tree for the development of an exoprosthetic hip joint.

The third one is the DoF in the joint. Considering the three active DoF on the hip and the single one (two at most) embed passively on the Canadian prosthesis, the first approach will lie between these two extremes, as a trade-off considering the device cost, weight, complexity, etc. (see Section 2.5). The last one is the shape of the joint. Even though it does not seem to be a particularly challenging/influencing item, it appeared recurrently in the experiences gathered. It is not a coincidence that it was traced (dotted line on the Figure 4.3) to other concepts, as it plays a major role in the way an amputee can relate to his/her prosthesis [51]. More arguments behind the notion of the shape will be presented in other leaves of the tree.

One of the main questions behind this design is presented in the leaf number **3. How**, in green color. The functionality is guided by three principles that lead each of the branches: *(i) ergonomics*, *(ii) safety* and *(iii) quality*. *(i) Ergonomics*, in light green, is the process of designing products so that they fit the people who use them and was approached by four main concepts. Of them, only one will not be studied in this project as it refers to the control implementation of the device which is contemplated as part of the future work. Comfort was studied in terms of the type of support that the prosthetic joint provides to the patient and how the interaction with the socket can be done. As seen, the Canadian prosthesis is supported in the anterior to the gravity line, which has to be revisited for this design. Two other concepts are related to the physical relief of lumbar pain and pelvic thrust, mentioned many times as critical for disarticulated amputees. Both of them have been addressed by the inclusion of hydraulic systems in the past, but the principle behind relays in the shock absorption capabilities of the system (also related to the compliance analysis of the joint in Section 2.5.4). Since the variability of the patients' physical characteristics is high, the adjustment of elements either internally or in external elements like the crural segment must be considered. To conclude with the ergonomics section there is the aesthetics. Even though this notion is more linked to a commercial device and the perception of it by the user and the environment, some hidden points are worth thinking about from the beginning. In accordance with the users, the shape is again highlighted; e.g. an amputee's impressions of the Helix3D joint pointed out that the leg-shape of the device made him feel better as it allows him to do simple tasks as placing a tray over his thighs when sitting on a chair [51]. The other one is how to eventually "camouflage" it visually and sonorously, which can be part of the social attribute of the device.

(ii) Safety, in a stronger green, covers a major challenge that is present in the daily life of disarticulated amputees: balance and risk/fear of slipping, tripping and falling [52]. They are drawn separately in the Figure but should be addressed together in the sense that the feeling of being at risk comes from the instability or lack of balance with the prosthesis. This is again linked to a phenomenon of the symmetric gait that has proven to work a lot with the Helix3D joint, which is the fact of being able to limit the step length. In this order of ideas, this also contributes to walking in unexpected terrains without falling or tripping that so far has been targeted by the control of the knee joint coupled with the Helix3D joint and has worked satisfactorily [57]. On a side note, thinking towards future commercialization requirements on CE marking and ISO norms should be included; especially as they provide the guidelines for a safe device. Nonetheless, this will not be studied at the time being.

Finally, *(iii) Quality*, in aqua green, is devoted to the standards that the device is supposed to achieve when working. The first and one of the most important of all is the decrease of energy expenditure for the patient. As seen, the inclusion of active modules has dealt with it in many cases and is one of the advantages of such configurations. But in the same way, elements of energy storage point towards positively affecting energy expenditure, even though this effect has not been tested with joints as the Helix3D [41]. The second one is the resistance to impact. In spite of the prevention of falling and the care that should be taken when walking in uneven terrains, the device will eventually receive some impact. Either coming from collisions with outside elements or as an effect of the cyclic loading of walking it is crucial to include this

criteria in the quality standards. The third one is the performance under different scenarios and, although it was mentioned regarding the terrains, it also includes a concept of weather resistance. In other words, the device should be able to withstand light exposure to weather elements such as the sun, rain, and wind; which is related to the sealing and casing of the device. The last item is the performance under demanding activities, which is out of the scope of this project and will be better explained in the following leaf of the objectives tree.

The next leaf, number **4. When** is presented in light blue. Starting from the testimonies and merging them with the list of activities established by the medical group in the LCI-5 with N1YN6 questionnaire (see Section 3.2) as principal sources of information, the activities were selected. In the Figure, they are presented as a combination of standing in upright position plus mobility or locomotion, as previously denominated. Among the activities that are related to mobility the focus will be placed on walking, sitting and climbing or descending stairs, ramps, and others; while more demanding activities will not be included. As presented in Section 2.5, the moment demand of the hip is similar when walking and stair ascending/descending only in the sagittal plane, for which considering the moments of walking would be a fair assumption. The moments in all the other planes were significantly higher when walking, which reinforces the idea. Power demands follow the same behavior as the torque, only that this time there is a difference between the mean peak in the sagittal plane of stair ascent and the one obtained when walking. Thus, power requirements could also be taken from walking and cover the ranges of the other activities included. Regarding the joint angles, mainly in the sagittal plane, one can see that the ranges are insufficient to be considered the maximum ranges of the device. Even other activities, like climbing stairs, fail to represent the maximum ranges desired in daily activities. Therefore, one should try to come as close as possible to the active range of motion (recall Section 2.3.3) to encourage activities like leaning and bending of the joint.

The last leaf to analyze is number **6. Recycle**, in light orange. This concept, although sometimes skipped, is very important in a design process and even more nowadays. The reason of the existence of this bullet point is the contribution of a conscious design to the optimization and treatment of all sources of waste. Therefore it is addressed by the implementation of modularity in the design, the re-use of components (when possible and under quality standards) and the selection of the adequate manufacturing processes to the components that are not commercially available.

It is evident from the objectives tree method that the design procedure can have many different levels of generality or detail. Therefore, the definition of some boundaries in the design and the establishment of concrete requirements is needed. To accomplish this task a process of refinement will be presented.

4.3 Functions and Specifications

The first part of this section is devoted to expressing the essential functions (i.e. those that the device must satisfy, no matter what physical components might be used) of the device. In this way, the problem level is consolidated by establishing a “boundary” around a coherent set of functions. The second part corresponds to the definition of specific requirements.

4.3.1 The Function Analysis Method

The first method implemented in this section is the function analysis method. The starting point for this method is to concentrate on what has to be achieved by a new design, and not on how it is to be achieved. The simplest and most basic way of expressing this is to represent the product or device to be designed as simply a “black box” which converts certain inputs into

desired outputs. The black box contains all the functions which are necessary for converting the inputs into the outputs. The next step is to break down the overall function into a set of essential sub-functions, followed by drawing a block diagram showing the interaction between them and, finally, drawing the system boundaries [50].

In order to be able to visualize the set of inputs and outputs of the device one has to go back to the principles governing the human body's locomotion. In Section 2.4.1, the nominal control loop for locomotion with details of the functional morphology for the performance of the joint while walking was presented. Having Figure 2.6 as the base line, the function analysis method was conducted and Figure 4.4 illustrates the results.

The first thing to discuss was the placement of the broader systems, i.e. the complete leg exoprosthesis. In Figure 4.4 it is represented by the gray rectangle in dotted lines and the relationship is set to be with the musculoskeletal system of the patient. Some prostheses, developed for above knee amputations, have pushed this interaction a bit further and have linked the prosthesis to efferent and afferent neurons (usually called neuroprosthetics); however, this project does not contemplate this kind of interaction. The first module inside the prosthesis is, therefore, the inclusion of sensors that can capture relevant biomechanical information (angles, forces, acceleration or orientation, etc) and convert them into data to which the second block, the electromechanical embodiment of the prosthesis, can react. Once this embodiment actuates according to the characteristics given, a third and optional block would be the inclusion of stimulation sensors that provide feedback to the body and are linked to the sensory information sent to close the loop of control. Nonetheless, this block is dotted in the Figure as it will not be studied in the project.

Contained in a simplified diagram of the electromechanical embodiment, one can find the three basic joint modules that interact together to achieve locomotion. From them, the hip module is expanded and the role is presented as concise as possible. Since this module is located at the top of the leg, it is the one in charge of the adjustment to the socket, which is the physical entry in the Figure. After it, the condition of stability is checked to know if either the patient has the intention of standing stable or to initiate movement, which in general terms is classified as sitting or bending situations (like leaning or crouching) or traveling activities such as walking or stair climbing. These activities were separated according to the angle and power requirements of the hip as studied in the previous section. At last and to close the loop in the joint is the phenomenon of shock absorption, also developed previously, more present in the second group of activities.

According to this characterization there are four main sub-functions to study and include in the design: (i) the actuation of the joint, by means of the powered mechanism, (ii) the shock absorption, also related to compliance, (iii) the fixation to the socket and (iv) the sensing, meaning the variables to measure and the sensors.

4.3.2 The Performance Specification Method

The second method is called the performance specification method. It is intended to help define the problem, leaving the appropriate amount of freedom so that the designer can manoeuvre over the ways and means of achieving a satisfactory design solution. The method emphasizes in declaring each specification to define a required performance, and not a required product [50]. A specification list is then generated according to the concepts and descriptions found with the objectives tree and the functional graph. It is important to make distinctions between Functions (F) which always relate to an action performed by the product or in which it is involved, and Constraints (C) which mostly relate to adapting the product to its environment and users. Then, for each one, performance criteria and levels were defined as Functional Requirements (FR) and Constraints Requirements (CR).

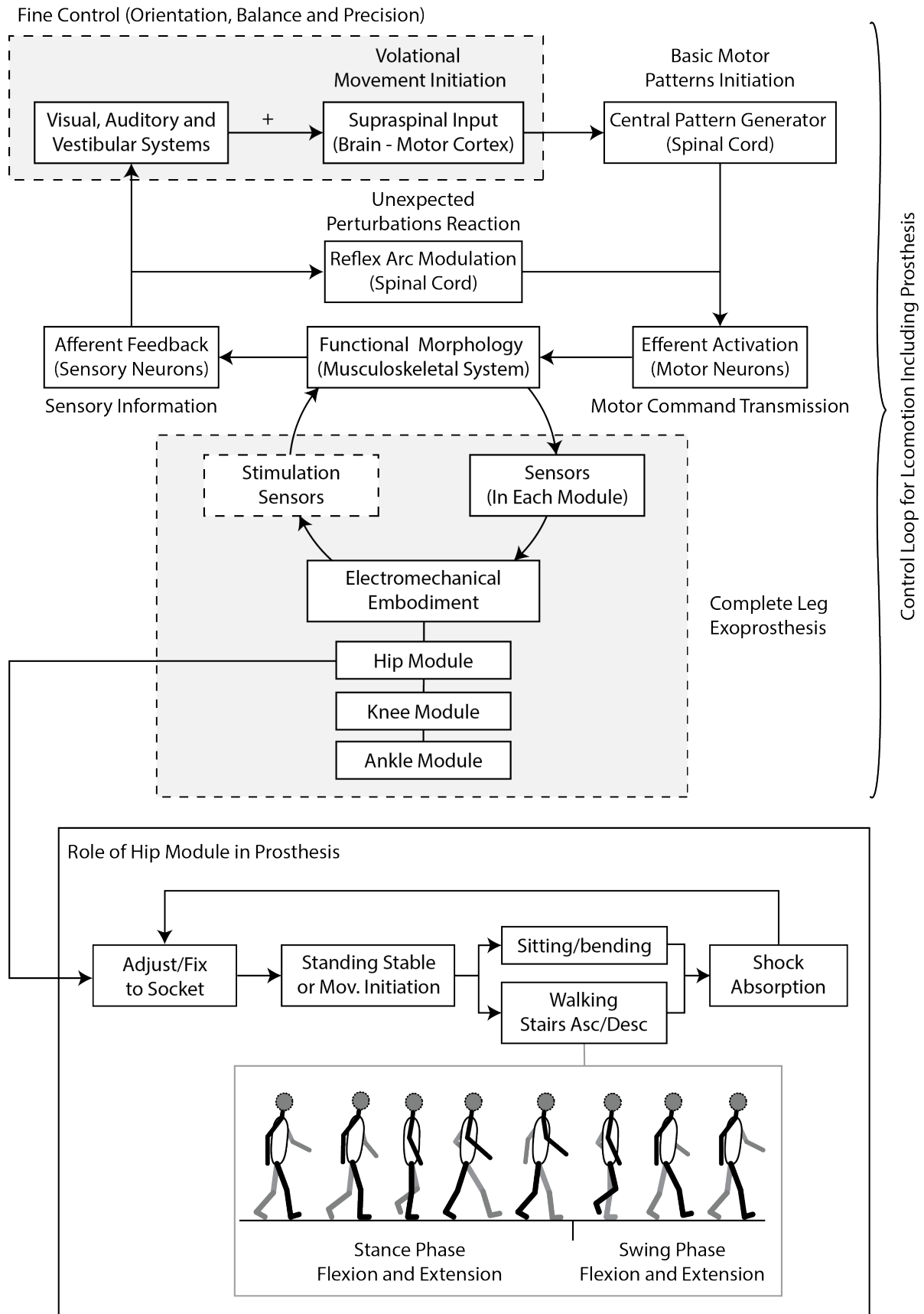


Figure 4.4: Transparent box model of the prosthesis functionality.

Specifications		
Context		An external prosthetic hip joint for disarticulated amputees to perform daily-life activities as close as possible to the performance of non-amputees.
Origin	Criteria	Requirements
Client	Functions (F)	F.1. Support body weight and provide stability
		F.2. Allow locomotion based on lower limbs (sitting and walking)
Client/ Designer	Constraints (C)	C.1. Be part of a complete prosthetic leg
		C.2. Be easy to control/use
		C.3. Be comfortable and pleasant
		C.4. Be safe to use
Designer	Functional Requirements (FR)	FR.1.1. Load line of body weight should fit between the medial-lateral axis of rotation of the hip and the pivot point of knee joint
		FR.1.2. Adjustable to different patients (crural segment)
		FR.1.3. Body weight to support (e.g. Helix = 100 Kg, abductors force up to 2 times weight during walking).
		FR.2.1. Consider interaction with knee joint of the prosthesis
		FR.2.2. Degrees of freedom: actuation in sagittal and frontal planes, passive in transversal plane
		FR.2.3. Range of motion from other activities different from gait (as close as active range of motion of hip joint as possible)
		FR.2.4. Torque and power peaks from walking biomechanics
		FR.2.5. Localization of center of rotation (fixed vs virtual rotation)
Client/ Designer	Constraints Requirements (CR)	CR.1.1. Volume to occupy (e.g. Helix3D = 146 mm height)
		CR.1.2. Consider weight of the joint (e.g. Helix3D = 990 gr + 275 gr tech, active components will increase it)
		CR.1.3. Operating temperatures (e.g. Helix3D = -10 to 60 C)
		CR.1.4. Connection to socket and crural segment
		CR.1.5. Consider interaction with other joints (out of the scope)
		CR.1.6. Be modular and easy to assemble
		CR.2.1. Signals to control (angles, forces, acceleration, orientation)
		CR.2.2. Initiation of movement (e.g. spring in Helix3D)
		CR.2.3. Avoid lumbar stress and thus, pain (compliance)
		CR.3.1. Upper leg-shaped (e.g. placing a tray)
		CR.3.2. Support points in the socket (single contact vs multiple points)
		CR.3.3. Avoid oscillations during stance (compliance)
		CR.3.4. Absorption of shock during gait
		CR.3.5. Reduce the energy expenditure (active and compliant)
		CR.3.6. Limit step length
		CR.3.7. Eventually, be hidden/well enclosed
		CR.3.8. Good finishing operations (for manufacturing of pieces)
		CR.3.9. No noise production, smell or leakage of substances
		CR.4.1. CE regulations (choice of material, infection contamination, risk of injury, related to electrical/mechanical components failures and monitoring of power supply (alarms))
		CR.4.2. Performance in unexpected terrains (e.g. weather resistance)
		CR.4.3. Resistance to impact

The process led to the specification list here displayed. Two main functions and four constraints, coming from both the “client” and design team, were attributed to the joint. From it, a long list of concepts was developed and, even though, not all of them are directly referenced to quantitative data (as there is not a precise reference to look after), some allusion was made to the Helix3D as baseline. Mainly, regarding its dimensions. The analysis of the available space was performed with the use of a video compilation of the patients experiences with the prosthetic hip joint as presented in Figure 4.5.

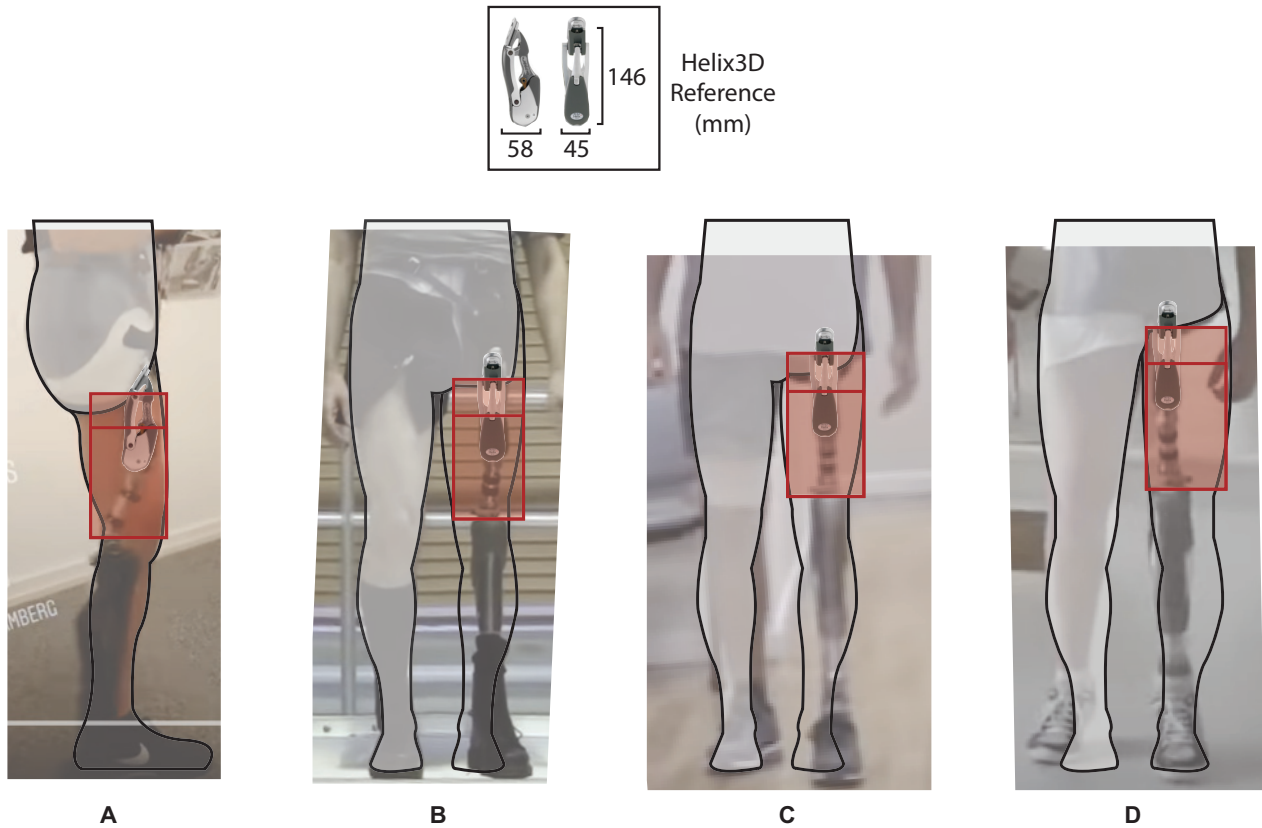


Figure 4.5: Analysis of the volume to occupy by means of video compilation of patients experiences [53]–[56], in that order. The reference of the Helix3D joint is presented and superposed to scale the images.

The videos were selected if they displayed patients in a standing position (or as close as possible) and the camera captured the lower limbs with the lens relatively parallel to them. An anatomical average silhouette (in blurry gray) was fitted to the snapshots images according to the width/depth and height of the patient’s hip and non-prosthetic leg. The reference of the Helix3D joint was used to equally scale all the images and roughly draw the volume (red boxes) that the device could occupy, taking into account the characteristics of the healthy leg. Two boxes were drawn per image: one from the attachment point of the prosthetic joint to the socket (including the 4-bar mechanism) until the beginning of the knee joint; and a smaller one only from the body of the Helix3D until the knee. By subtracting the second from the first, and considering the dimensions of the Helix3D, one could obtain the volume of a possible moving mechanism if needed. The dimensions of these boxes and their average are presented in Table 4.1. Based on them and trying to match a cylindrical-leg-shape, a circumference of diameter $\varnothing 115$ mm was selected. Regarding the height, the most conservative approach was taken, by selecting the smallest heights that corresponded to the ones in Figure 4.1.B: 212 and 157 mm. Additionally, this big height corresponds to one half of the minimum femur length (1/8 of human’s total height) [58], [59], which can also be used as a reference.

Table 4.1: Approximation of the available space for the placement of the prosthetic joint. The values between parenthesis correspond to the inner box in each case.

Figure 4.5	Height big box (mm)	Height small box (mm)	Width (mm)	Depth (mm)
A	218.7	167.08	-	115.5
B	212.3	157.03	106.4	-
C	217.7	159.75	115.8	-
D	245.9	190.77	121.1	-
Avg.	223.6	168.65	114.4	115.5

There is another item worth checking for lower limb prostheses and it is mentioned in the FR.1.1 of the list. As mentioned in Section 2.3.2, the load line is key to the proper weight distribution and balance. Thus, when constructing a prosthesis, the load line should run through the joint centers as it does in sound individuals [28]. In the sagittal plane, it means that it should run through the centre of gravity, then 2 mm posterior from the hip joint, 15 mm anterior from the knee joint, and 60 mm anterior from the ankle joint. Other basic characteristics mentioned on the list have been presented extensively in this document.

Additionally to the list specifications, a set of specifications for the design based on the torque and power requirements seen in Section 2.5 was established. In Table 4.2 the values for a 75 Kg person, which is the point of reference selected, are presented.

Table 4.2: Design specifications for a 75 Kg person.

Variable	Value
Weight (kg)	Approx. 4 (Helix joint: 1.2 to Human thigh: 7.8)
Maximal flexion (°)	Walking: 38 Sitting: 120
Maximal extension (°)	Walking: 3 Sitting: 0
Peak torque (Nm)	Flexor: 87.75 Extensor: 80.25
Peak velocity (rad/s)	3.24
Peak power (W)	142.73
Net work done (J)	21 at 1.06 m/s

After setting up the context, functions, sub-functions, requirements and details that enclose the device, the floor is given to the next section for the potential solutions to start appearing.

4.4 Alternatives Generation

The generation of solutions is, of course, the essential and central aspect of designing. Whether one sees it as a mysterious act of creativity or as a logical process of problem solving, the purpose of design can be in (1) making a proposal for something new, which does not yet exist (it is not the case most of the times given the contexts) and/or in (2) making a variation from or modification to an existing product (clients usually want improvements rather than novelties). Actually, the second one is also considered to be the way in which much creative thinking actually develops. In particular, creativity can often be seen as the re-ordering or re-combination of existing elements [50].

The morphological chart method relies on this phenomenon, and encourages the designer to identify novel combinations of elements. The idea is to generate a chart with the complete range of elements, components or sub-solutions that can be combined together to make a solution. The number of possible combinations is therefore high, and includes from conventional solutions to a wide range of variations and completely novel solutions. Morphology is the study of shape or form; so a morphological analysis is a systematic attempt to analyze the form that a product or machine might take, and a morphological chart is a summary of this analysis [50]. It consists on, first, listing the sub-functions essential to the product (the ones found in Section 4.3.1, with the function analysis method). Then, for each of them, the means by which it might be achieved are listed and drawn into the chart. This is done until the chart contains all the possible sub-solutions. Finally, the feasible combinations of sub-solutions are identified. Brainstorming, analogy method and other activities are useful and were implemented and organized in the chart presented in Figures 4.6 to 4.9.

Starting with the actuation, in Figure 4.6, three major possibilities could eventually be included, each of them including several subcategories: (i) *Electric actuators*, meaning electrically powered. They could either be linear or rotatory. The linear actuators consist of motors, linear guides, and drive mechanisms and are used primarily in automation applications when a machine component or tool requires a controlled movement to a particular position. These actuators can be driven by several means including ball or lead screws and belts among others. The rotatory actuators consist of motors and output shaft mechanisms with limited rotary travel and are used primarily in automation applications when a mechanism requires controlled movement to particular rotational positions. These actuators are usually driven by various motor types. (ii) *Hydraulic actuators* and (iii) *Pneumatic actuators*, which are fluid powered devices by means of hydraulic fluid, gas, or differential air pressure, respectively. They could also be linear or rotatory. Linear actuators consist of cylinder and piston mechanisms that produce linear displacement; while rotatory actuators include cylinders and piston mechanisms, gearing, and output shafts giving limited rotational travel [60].

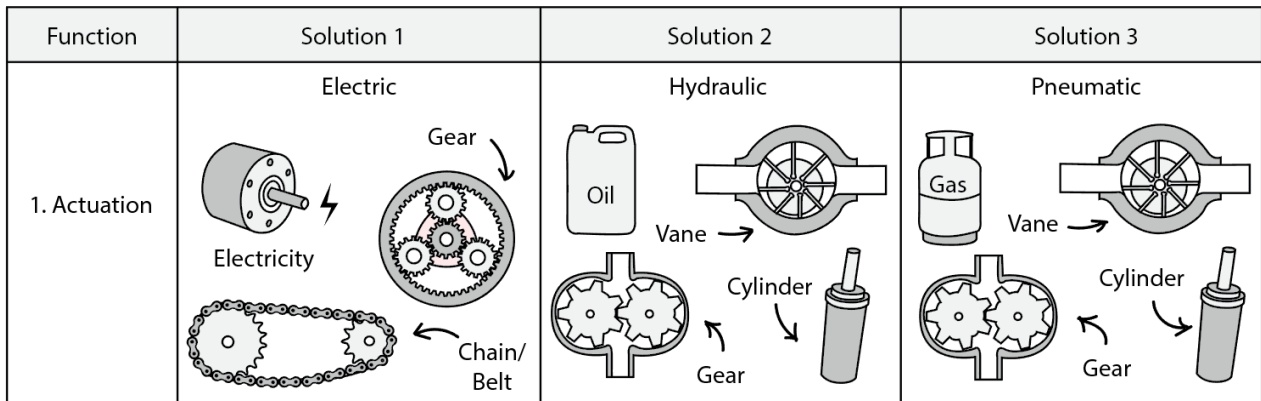


Figure 4.6: Morphological chart for the actuation of a exoprosthetic hip joint.

Regarding the shock absorption, in Figure 4.7 and also related to the compliance previously revised, the alternatives include first, cylinders either hydraulic or pneumatic. An hydraulic approach would follow the ones included in the Helix3D system, which have been reported to be beneficial. Pneumatic ones could be in the form of air springs as it has been included in prosthetic knee joints. The second solution is dampers, which are commonly hydraulic. They could be as presented in the chart or eventually include a metallic spring around them. The last solution is a metallic spring, which, depending on how the load force is applied to them, could be either linear (tension/compression, resisting to axial forces) or torsional (resisting to torque or twisting forces), with either a constant or variable stiffness.

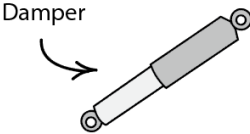
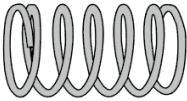
Function	Solution 1	Solution 2	Solution 3
2. Shock Absorption/ Compliance	 Cylinder	 Damper	 Metallic Spring

Figure 4.7: Morphological chart for the shock absorption in an exoprosthetic hip joint.

Concerning the socket support, in Figure 4.8, three conditions were proposed. A vertical adjustment where the leg is located underneath the socket, as it was done in some of the first Canadian prostheses. A lateral adjustment similar to the one reviewed in Section 3.3 for the active hip prosthesis. And finally, an anterior adjustment as it is commonly placed in current prosthetic legs.

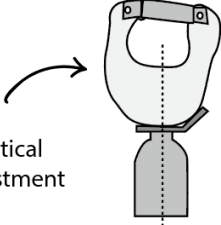
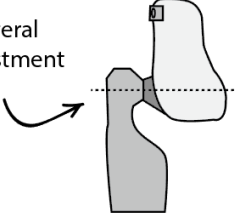
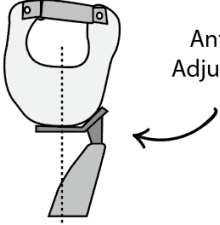
Function	Solution 1	Solution 2	Solution 3
3. Fixation to Socket	 Vertical Adjustment	 Lateral Adjustment	 Anterior Adjustment

Figure 4.8: Morphological chart for the fixation to the socket of an exoprosthetic hip joint.

Finally, in respect to the sensing module in Figure 4.9, the most common parameters to be controlled during gait and related activities in robotics were included. This means angles and position (e.g. by means of potentiometers or encoders), forces and deformations (e.g. by means of strain gauges) and acceleration/ orientation (e.g. by means of inertial movement units).

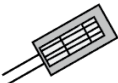
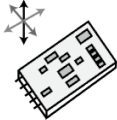
Function	Solution 1	Solution 2	Solution 3
4. Sensing	 Angles/ Position	 Force/ Deformation	 Acceleration/ Orientation

Figure 4.9: Morphological chart for the sensing in an exoprosthetic hip joint.

With all the alternatives, the next step would be to provide as many combinations as possible to then assess. Nevertheless, following strictly these guidelines, some of the specifications could be ignored or not fully achieved. Therefore, the assessment of the alternatives will be performed at the level of the sub-solutions.

4.5 Relative Assessment

In a design process, the designer is faced with the problem of selecting the best alternative from a range of potential solutions. In cases like this one, there are also decisions of choice

to be made between alternative sub-solutions that might be incorporated into a final design; which shows that choosing between alternatives is therefore a common feature of design activity. Choices can be made by guesswork, by intuition, by experience, or by arbitrary decision. However, if the choice can be made following a rational procedure, it strengthens the solution and allows other members of the design team to assess the validity of the choice [50]. For each of the functions, and taking into account the objectives and specifications previously stated, a small discussion of assessment was held.

1. Actuation

In regards to the actuation many characteristics and applications have been extensively documented and well understood. The first alternative is **electric actuators**. The main **advantages** of this type of actuators are: first, they offer the highest precision-control positioning, which is needed in applications like this one. Second, their setups are scalable for any purpose or force requirement, and are quiet (in terms of noise, they are quieter than pneumatic and hydraulic actuators), smooth, and repeatable. Finally, since there are no fluids involved, no leakages and related hazards are expected. These properties match with the specifications listed before (CR.3.9). Among the main **disadvantages** is the initial unit cost of an electrical actuator which is higher than that of pneumatic and hydraulic actuators. However, over the long run it tends to be mitigated. Additionally, they are not suited for all environments, therefore they would need proper sealing and protective systems for certain conditions (related to CR.4.2). However, they can work outdoors where air systems can freeze. Last, the motor chosen locks in the actuator's force, thrust, and speed limits to a fixed setting. Which means that parameters should be selected to match maximum value cases (related to FR.2.3,2.4 and CR.1.6.).

In the case of **hydraulic actuators**, the main **advantages** are: first, that they are rugged and suited for high-force applications. They can produce forces 25 times greater than pneumatic cylinders of equal size and have higher horsepower-to-weight ratio than a pneumatic motor. Second, an hydraulic actuator can hold force and torque constant without the pump supplying more fluid or pressure due to the incompressibility of fluids (possibly linked to CR.3.4). And finally, they can have their pumps and motors located considerably distant to each other with minimal loss of power (related to CR.3.7). On the other hand, the main **disadvantages** include that, as mentioned, hydraulics usually leak fluid, which leads to less efficiency and potential damage to surrounding components and areas (CR.3.9 again). Additionally, they require many companion parts, including a fluid reservoir, motors, pumps, valves, and others, along with noise-reduction equipment. This could make some systems difficult to accommodate (related to CR.1.2,3.7).

In the literature, hydraulics have been used mainly in prosthetic knees and the Helix3D hip joint. Even though results have been mainly positive, issues related to failure have arisen. It has been reported that malfunction of hydraulic knees can be difficult to assess by the patient and, thus, early detection is difficult. Since it takes time to notice, failure results in oil leakage and loss of support, producing embarrassing and dangerous situations for the amputee [61].

To conclude, the **pneumatic actuators** are known to have as main **advantages** the fact that they are rather simple actuators and can work in hostile environments, meaning in high temperatures, explosive environments and humid rooms (related to CR.4.2). Also, the cost of pneumatic actuators is low compared to other actuators. However, as its been said, among their **disadvantages** are the pressure losses and air's compressibility makes them less efficient. This means that for them to be truly efficient, pneumatic actuators must be sized for a specific job. In addition, even though the air is easily available, it can be contaminated by oil or lubrication, which leads to the inclusion of air compressors as part of the system (related to CR1.6,3.7).

Pneumatic systems have also been used in many fields in the literature. In the book ‘Delft Pneumatic Biped’ [62], authors state that the main drawback of pneumatic actuation is the lack of sufficiently lightweight commercially available components, which means that many of the components must be specifically developed for research projects. A second drawback found is the difficulty of implementing high-bandwidth control, which they say, may be a reason for them to switch to electric DC motors for their future prototypes.

The important attributes of an actuator targeted to mimic the hip joint capabilities can be summarized in terms of: (i) the output torque, power, efficiency and the maximum speed (stated in Table 4.2), (ii) the assembly configuration relative to the rest of the device, (iii) the space itself occupies (in Table 4.1), (iv) the undesired features (noise, smell or leakage), (v) the operating conditions and finally (vi) the cost. For an overall assessment, based on the advantages and disadvantages previously presented, it becomes clear that the *electric actuators are the most suitable actuators for this application*. A discussion about the means of implementation will be addressed further in the section.

2. Shock absorption and compliance

The following decision is related to resistance and shock absorption. The first two alternatives presented in Table 4.7 have been extensively implemented in the literature. Mainly in knee prosthetics, given the biomechanical nature of the knee joint, but also in the Helix3D hip joint. Therefore, the characteristics have already been presented at this point. Regarding the springs themselves and considering the selection of electrical actuators, one field of study that has gained attention over the last years is the one of compliant actuators. Series elastic actuators (SEA) and parallel elastic actuators (PEA) are part of this type of actuators that have been implemented in devices interacting with humans as they allow force to be controlled by a simple principle. The SEA is essentially a spring in series with a stiff actuator. The compliance is determined by the spring constant and is therefore not adjustable during operation as presented in Figure 4.10. In a typical setup of an SEA, elongation of the spring is used as force measurement and fed back in the control loop [63]. In this field, the main purpose of using a series spring is to store the energy and reuse it during power-demanding moments during gait, thus, reducing the peak motor power. On the other hand, but trying to follow a similar approach, PEA implements a spring in parallel with the stiff actuator. In this case, the parallel spring has been used as it reduces both the torque and power demand on motor and gears [35].

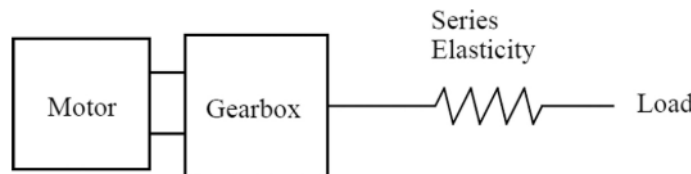


Figure 4.10: Series elastic actuator schematic. Image from [64].

Since the references in knees and the Helix3D are based on passive systems, a brief documentation on actuated devices was conducted for this assessment. The first contribution, although not in the field of prosthetics, is the one by Hobbelen et al. [65]. In their work, they present a system overview of bipedal robots Flame and TULip, and discuss their choice of SEA. They said: “The use of electric DC motors generally allows high torque control bandwidth with relatively little system overhead. In comparison to pneumatic or hydraulic actuators this makes the electric motor an attractive candidate for autonomous robots. The application of a geared motor instead of direct drive is chosen because in case of direct drive the magnitude of the desired torques in

our robots would require a heavy motor with overly high power rating. However, the use of a geared electric DC motor typically results in a high actuator impedance (i.e., high reflected inertia and damping/friction). This undesirable property is best solved by the application of torque control through a series elastic/compliant element". The second contribution here studied, comes back to the field of prosthetics as Grimmer et al. [66] simulated monoarticular SEAs in an actuated prosthetic leg for walking and running and evaluated energetic and peak power changes. They found that in walking, including a series elastic spring at the knee and the hip would decrease the energy requirements. They sustained that elastic structures may increase gait robustness to uneven terrain or sudden changes in the ground. Furthermore, they said that if the benefits in energy or peak power requirements are small, this could be a reason to consider the elastic structures for the system design.

This being said, the fact of implementing a compliant actuator is highly in accordance with what was explored in Section 2.5.4, where compliance was analyzed in the hip joint. Exploiting the capabilities of either *SEA*, *PEA* or a combination of the two is therefore the alternative selected for this function.

3. Fixation to the Socket

The next decision is in regards to the third function, the fixation to the socket. Underlying this function, that might seems like a simple placement of the prosthetic joint, there are some concepts that should be considered. First, the positioning of the leg with respect to the socket conditions its axis of rotation. At the same time, if the axis of rotation is selected to be fixed or virtually rotating, the mechanism of motion is also conditioned. Consequently, the space for placing the actuator is affected. In order to assess the different alternatives, the few experiences with existing prostheses were valuably considered.

Among the first documentations on the release of a prosthetic hip joint to the market are the ones by Ottobock. In the prosthetic follow-up presented by Shurr et al. [67], the Ottobock's modular prosthesis is mentioned for the first time due to its lighter overall weight, improved cosmetics and interchangeable and adjustable components. During their study, they interviewed six hip disarticulated and analyzed the factors which contributed to or detract from prosthetic use. They found that some of the factors affecting the use of a prosthesis included discomfort while in a sitting position, due to the placement of the hip joint, and the fragility in the rubber bands implemented in the models (basic Canadian prosthesis configuration). To solve those inconveniences, the module was re-designed and released as the 7E7 (see Figure 4.11), already mentioned in this document in Sections 1.2.3 and 3.2, which later evolved into the Helix3D.

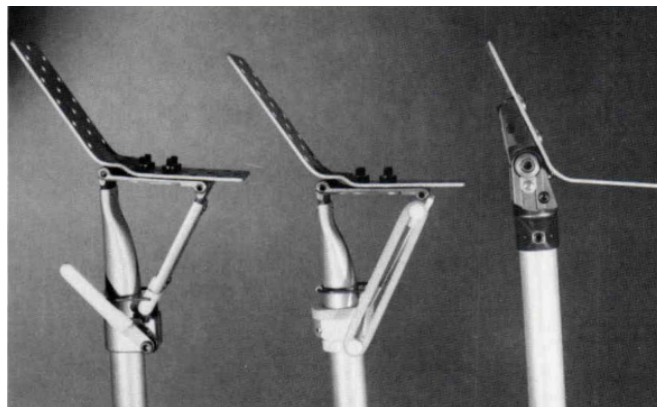


Figure 4.11: The chronological progression of the Ottobock Modular Prosthetic Hip Joints. To the right the 7E7 prosthesis. Image from [67].

As it can be seen, the 7E7 was positioned more anterior than distal in reference to the socket. From then on, most of the passive hip joints have been located in the same fashion. Conversely, the only powered prosthetic hip joint was placed laterally to the socket [49] to more easily stay at the same height of the biological joint. However this configuration makes the prosthesis grow laterally in a way that can obstruct activities like sitting in a chair and alter the cosmetics (also referred to as the aesthetics in this manuscript) which have a high value for the users. The socket area enveloping the entire pelvis already adds significantly to the hip and waist circumference, the less distortion of the hip area, the better for the user. Given that the anterior placement has proven to be stable and comfortable in the reviewed experiences with both the 7E7 and Helix3D joints, this configuration is selected to advance in the design. The challenge then remains in placing a virtual center of rotation close to the biological one, but with the anterior location and allowing a sitting position.

Once again, the experiences were of high importance. In 1988, in the Netherlands a study by Tuil [68] decided to implement a hip joint using the Ottobock four-bar knee disarticulation joint mounted in reverse (see Figure 4.12). There he said that even though at first it was questionable whether the joint would be strong enough, it proved not to be an issue. On the contrary, he found two advantages in the use of the four-bar hip joint. The first one was that the patients walked with a lower energy expenditure because the prosthesis shortened the swing phase. He claimed that, in contrast to patients who had worn older style hip disarticulation prostheses, the patients with the new style prosthesis walked more and had indicated that they used less energy. However, this effect was not quantitatively proven and even nowadays it is missing assessments with the Helix3D. The second advantage was that there was hardly any strain on the cosmetic cover, so much less damage was done. He also mentioned that with the four-bar joint the construction could be less critical and the whole prosthesis could be readily adjusted, noting that “many adjustments are possible with regard to rotation in the hip joint itself. The lack of facility to adjust abduction has never been a problem”. His work was acknowledged by other studies. In Ottawa, Canada, the approach was reproduced [69] and they emphasized that even if Mr. Tuil’s article dealt with pure speculation, it was very helpful in proving that a knee joint could be successfully used as a hip joint.

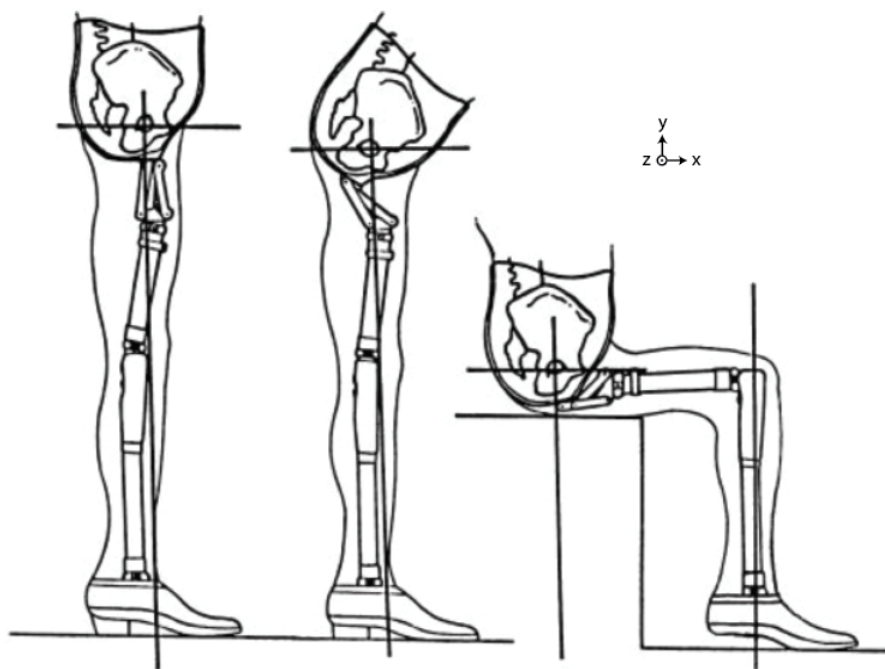


Figure 4.12: Schematic of motion permitted by using an inverted polycentric knee joint as a hip mechanism, first advocated by Tuil of the Netherlands. Image from [70].

Although, to the knowledge of the author, there are no mathematical or biomechanical models that prove the advantages of a polycentric hip joint design in terms of torques and force load, the evidence gathered here, together with the performance showed by the Helix3D (a polycentric joint) is considered to be a good omen for the design. Therefore, *a polycentric design with anterior fixation to the socket is selected for this function*. That being said, it is clear that further simulations and models to assess the configuration proposed by the end of this project will be needed.

4. Sensing

The last decision is related to the type of sensor to implement. However, this is not a situation like the previous where selecting one would discard the others. Actually, all of them could eventually be included as they measure valuable parameters for the control of the prosthesis and complement each other. Hence, the decision will lie in characteristics of other elements (e.g. some motors include encoders and SEA could be used in force control) and will be made farther in the design, trying to incorporate as many as possible.

4.6 Conclusions

This chapter presented the design methodology followed to guide the design proposal for an exoprosthetic hip joint. The methodology was based on the model developed by Cross. The model understands the design process as a decomposition of the overall problem into sub-problems, for which sub-solutions are evaluated and combined to generate an overall solution. Since the process intends to develop solutions as the problem is being understood, seven stages were defined to easily keep track of the different outcomes. For each of them, a specific method was implemented in this chapter. A broad analysis of the problem was presented with the objectives tree, followed by a functional analysis in which major functions were established. Then, based on them, a list of specifications and solution alternatives was developed. Finally, the initial assessment of the functions was done to select the major elements of the design. A polycentric, compliant (still to define if by a SEA or PEA) electromechanical prosthetic hip joint, fixed anterior to the socket and with angle, position, force and acceleration sensing was chosen. However, there are still many features to be defined and dimensioned. The next chapter will present the mechanical implementation that followed this analysis, in terms of the gross dimensioning of the main components, the selection of the most important technological component in catalog and a proposal of their assembly.

Chapter 5

Mechanical Implementation

5.1 Introduction

Having a pre-defined configuration of the hip module in terms of the actuation and physical disposal, the next step is to dimension and select the suitable components. It is important to mention that when this step is reached in a design process, the dimensioning is key as it narrows down the list of options available in the catalogs or sets requirements for components that need to be manufactured/developed (when they are not found in the market or additional constraints prevent their acquisition). The first section of this chapter will be devoted to the effect of adding an SEA or PEA and the selection of an appropriate stiffness based on the chosen actuator. Once the motor characteristics have been re-evaluated for the stiffness, the selection of a motor and gearbox in the catalog will be done. Finally, a proposal for the assembly and the integration with a polycentric design will be presented.

5.2 Component Selection

Three different approaches were evaluated for the design of a compliant actuator. The first one was based on the inclusion of a pure SEA. Then, the effects of only a PEA were analyzed. At last, given the properties that will be presented, a series-stiffness for a PEA was also contemplated. The actuator peak power requirements were derived similar to the ones presented in [35], [66]. Therefore, before explaining the optimization method, some assumptions were made following their work:

- For both SEA and PEA, linear springs were considered;
- Frictions due to motor and transmission were neglected and the efficiencies of the motor and gears were not included, as motor copper and iron losses were not included;
- Parallel springs engaged and disengaged when the spring is at rest (zero spring force) to avoid chattering;
- The inertias of the motor and gears were not included. Thus, the power to accelerate the motor and gear was not included;
- The motor required the same amount of power for the driving and braking modes;
- In this project, the author chose to optimize the design for a 75 kg test subject.

5.2.1 SEA

The main purpose of using a series spring is to store energy and reuse it during power-demanding moments, so that the peak power demand on the motor can be reduced [35].

Motor power P_m was calculated as

$$P_m = T_m * \dot{\theta}_m \quad (5.1)$$

where T_m is the motor torque and $\dot{\theta}_m$ is the motor angular velocity. T_m is equal to the torsional spring torque, T_s , and both are oppose of the external torque, T_{ext} (the joint torque). $\dot{\theta}_m$ is the derivative of the motor angle, θ_m , calculated from total actuator angle θ_g and spring angle θ_s .

$$\theta_m = \theta_g - \theta_s \quad (5.2)$$

The spring angle was calculated based on Hook's law

$$T_{ext} = K_s * \Delta\theta_s \quad (5.3)$$

where K_s is the spring constant. The deflection of the spring, $\Delta\theta_s$, was calculated by subtracting the instantaneous spring angle, θ_s , from the spring angle at rest, θ_0 .

$$\Delta\theta_s = \theta_0 - \theta_s \quad (5.4)$$

Finally, the equation for calculating spring angle is

$$\theta_s = \theta_0 - \frac{T_{ext}}{K_s} \quad (5.5)$$

By using θ_s in Equation 5.2, motor angle θ_m was calculated.

$$\theta_m = \theta_g + \frac{T_{ext}}{K_s} - \theta_0 \quad (5.6)$$

The derivation of Equation 5.6 results in motor angular velocity $\dot{\theta}_m$.

$$\dot{\theta}_m = \dot{\theta}_g + \frac{\dot{T}_{ext}}{K_s} \quad (5.7)$$

The final equation for calculating motor power is

$$P_m = T_m * \left(\dot{\theta}_g + \frac{\dot{T}_{ext}}{K_s} \right) \quad (5.8)$$

For calculating the peak power, only absolute values are taken into account. And, since the motor has mimic the joint torque profile, the power is

$$P_m = \left| T_{ext} * \left(\dot{\theta}_g + \frac{\dot{T}_{ext}}{K_s} \right) \right| \quad (5.9)$$

If one considers the case where the stiffness, K_s , is nearly infinite (i.e. direct drive), the spring power term drops to zero and the motor must provide the absolute value of normal gait power. In the opposite case, if the stiffness is near zero, the power requirements tend toward infinity. The objective is to find a spring stiffness that minimizes the peak motor power.

The optimization was conducted with the data presented in Section 2.5 for a healthy individual of 75 Kg. Optimal spring stiffness values for minimizing peak power requirements were determined by a brute-force search method. Stiffness values between 1 Nm/rad and 100000 Nm/rad with 1 Nm/rad increments were included in Equation 5.9. The brute-force search method was used because computations require only some seconds to get a precise result from all possible solutions.

Figure 5.1 and 5.2 present the peak motor power as a function of the spring stiffness to determine the optimal stiffness. Both reveal an interesting relationship, as the extreme cases described earlier have shown. At a stiffness value near zero, infinite motor power would be required; and at high stiffness, the power asymptotically approaches the peak gait power (direct drive, in green).

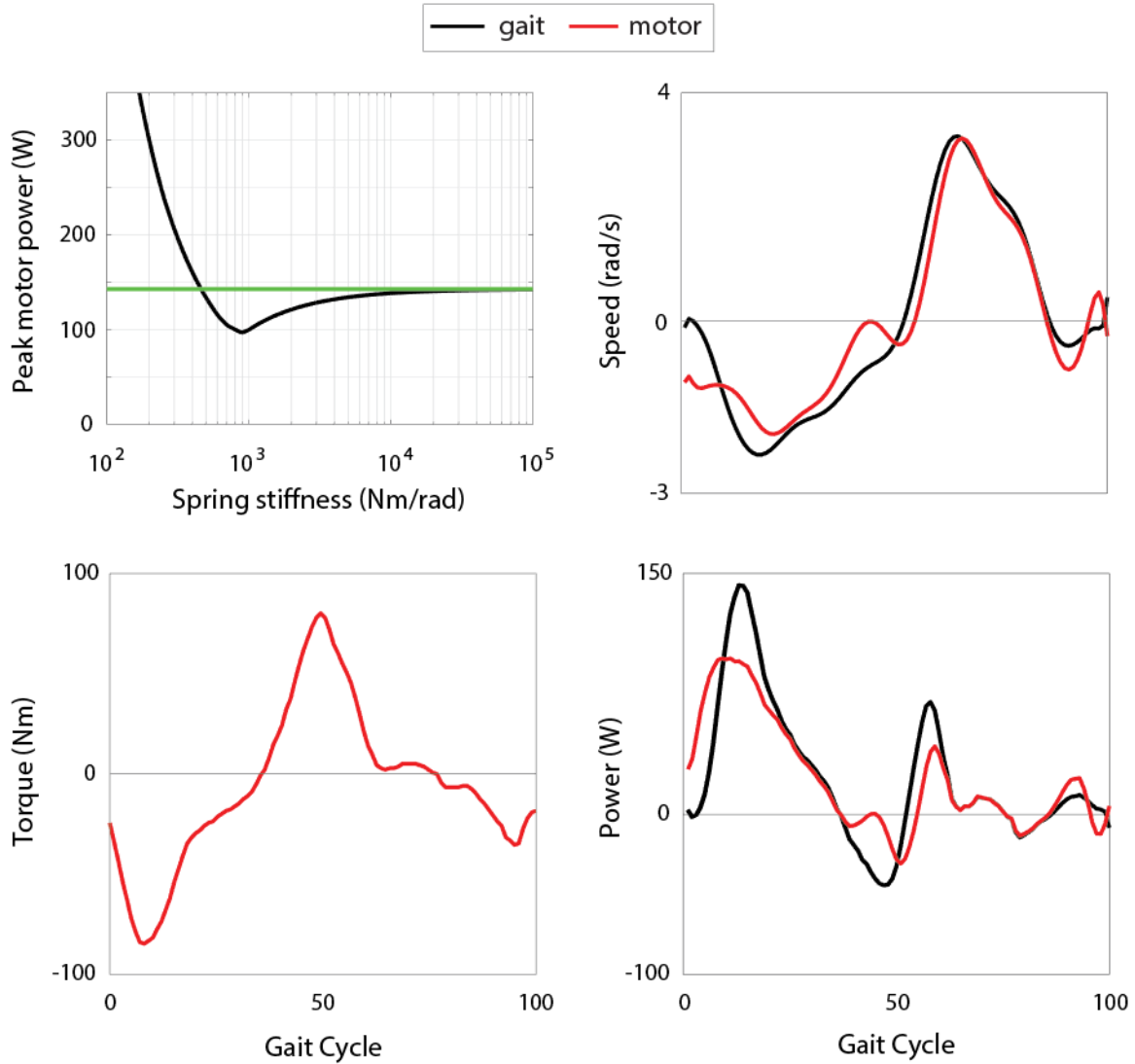


Figure 5.1: SEA for sagittal hip of a 75 Kg person. The biological curves are presented in black and the motor curves in red: the motor speed, torque and power. Optimal spring reduces the motor power by reducing the motor speed, and the motor torque is the same as joint torque. In green the original direct drive peak power.

After selecting the optimal stiffness for each case (see Table 5.1) the angular velocity, torque and power were calculated for the SEAs. As it can be seen, the series elasticity can reduce the power demands as it influences the angular velocity of the actuator. Nevertheless, it cannot reduce the demand on motor torque and this is the reason why the torque pattern is not altered.

Table 5.1: Predicted optimal SEA stiffness for sagittal and frontal planes of the hip for a 75 Kg person.

Variable	Sagittal	Frontal
SEA Stiffness (Nm/rad)	908	1457

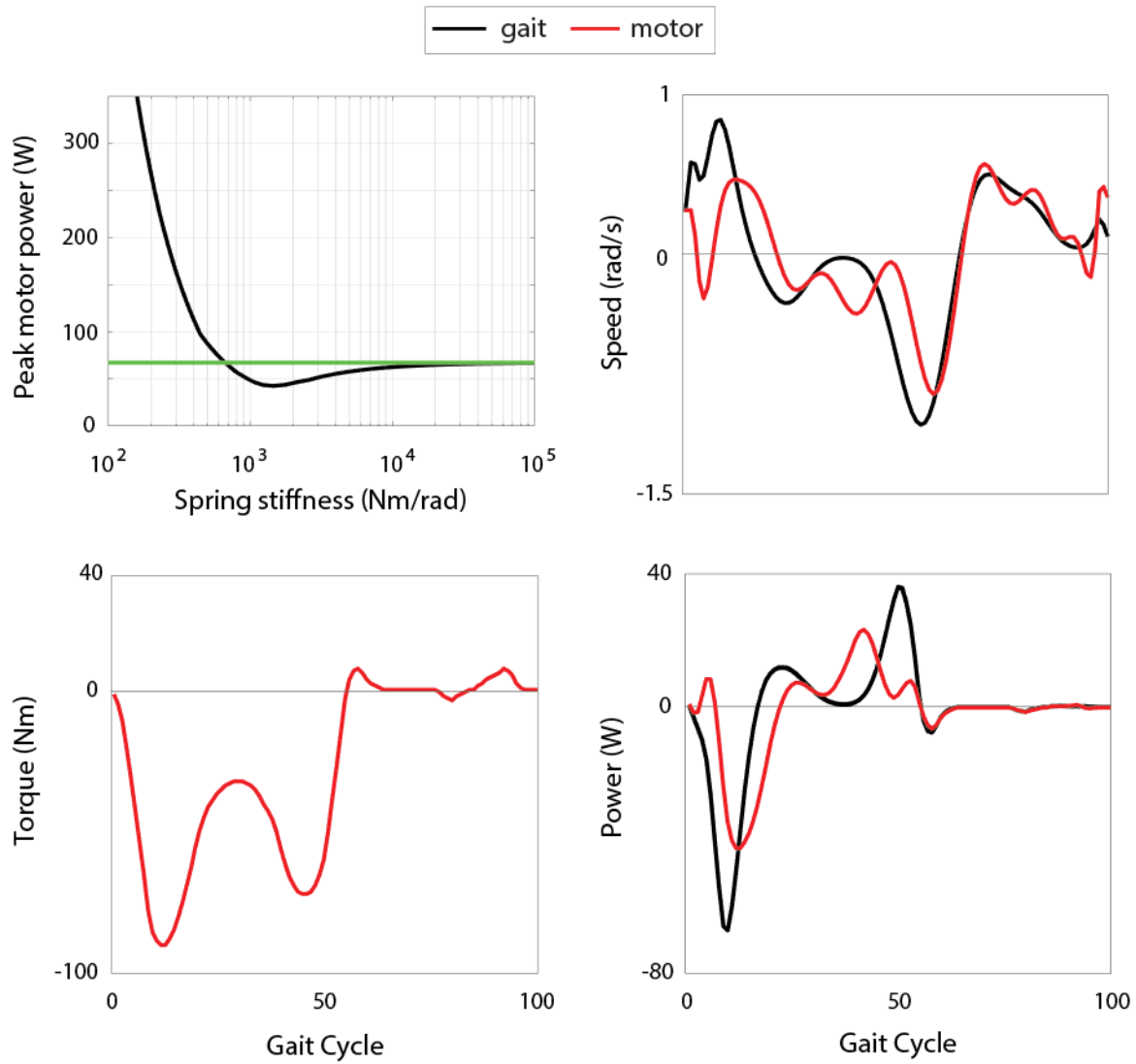


Figure 5.2: SEA for frontal hip of a 75 Kg person. The biological curves are presented in black and the motor curves in red: the motor speed, torque and power. Optimal spring reduces the motor power by reducing the motor speed, and the motor torque is the same as joint torque. In green the original direct drive peak power.

The maximal angular velocity, peak and RMS power demands of the SEA at the sagittal and frontal planes are presented in Table 5.2. As it can be seen, there is a considerable reduction of the peak and RMS power for both planes and it is similar in magnitude.

Table 5.2: Predicted velocity and power effects of the implementation of an SEA at the sagittal and frontal planes of the hip for a 75 Kg person.

Variable	Sagittal			Frontal		
	Direct Drive	SEA Motor	Change (%)	Direct Drive	SEA Motor	Change (%)
Peak velocity (rad/s)	3.24	3.20	1	1.06	0.87	17.9
Peak power (W)	142.7	97	32	67.1	42.3	36.9
RMS power (W)	49	41	16.3	17.7	13.23	25.2

In [35] an even higher reduction for the frontal plane was obtained (peak power by 60% and RMS power by 49%) and no obvious reduction in the sagittal plane was reported. On the other hand, in [66] a similar reduction was found (peak power by 26.6%) for a walking speed of 1 m/s in the sagittal plane, which is roughly the same speed considered for the simulation. The variation among the results obtained by those studies and the ones found here may be attributed to the characteristics of the power profile of the hip joint previously mentioned in Section 2.5. However, the results obtained here seem to reconcile both previous findings, as they are in between both cases. In this sense, at this point it seems to be advisable to include an SEA for both planes. The next trial is to analyze the effects of a PEA to determine the best alternative.

5.2.2 PEA

The primary goal of using a linear parallel spring is to reduce the peak torque demand on the motor and gears during gait, and thus, its power requirements. The methodology used was the fitting of a line (linear spring) in the torque-angle curve by means of the least squares (LS) method. The LS method distributes the torque evenly around the spring torque-angle curve [35]. It was implemented in Matlab® with the FIT function and a linear polynomial curve.

Figure 5.3 and 5.4 show the torque-angle curves with the fitted curves for a parallel spring and Table 5.3 presents the predicted stiffness for both cases. When a parallel spring is added to the joint, the motor is in charge of delivering only part of the joint torque, which is the difference between the required joint torque and spring torque. The torque and power profiles at the hip are shown in Figure 5.5.

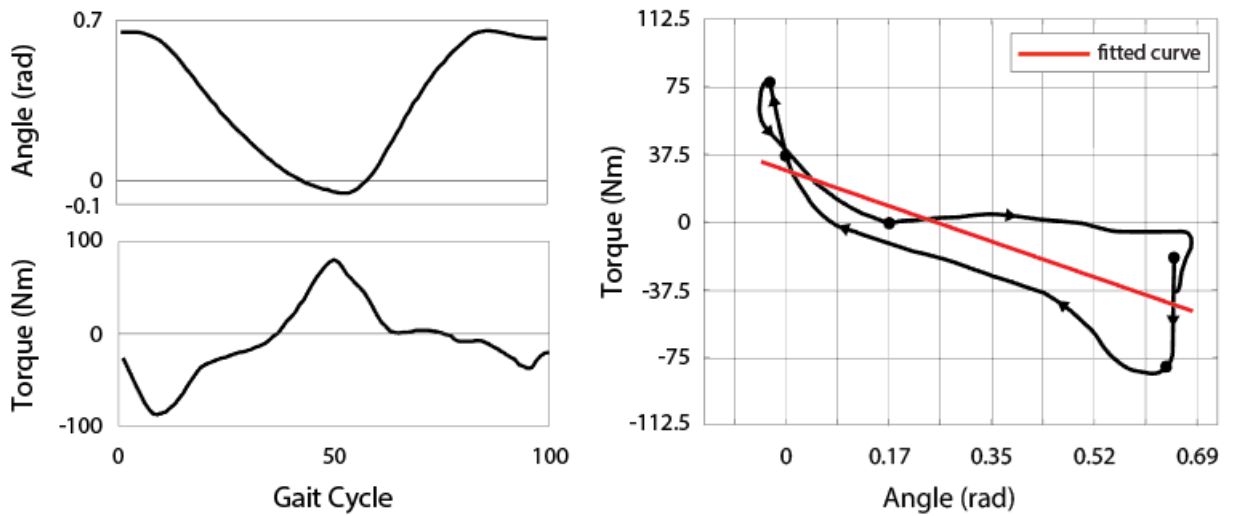


Figure 5.3: Sagittal torque-angle plot with parallel spring fitted by a linear regression, for the hip joint of a 75 Kg person.

Table 5.3: Predicted optimal PEA stiffness for sagittal and frontal planes of the hip for a 75 Kg person.

Variable	Sagittal	Frontal
PEA Stiffness (Nm/rad)	116	472

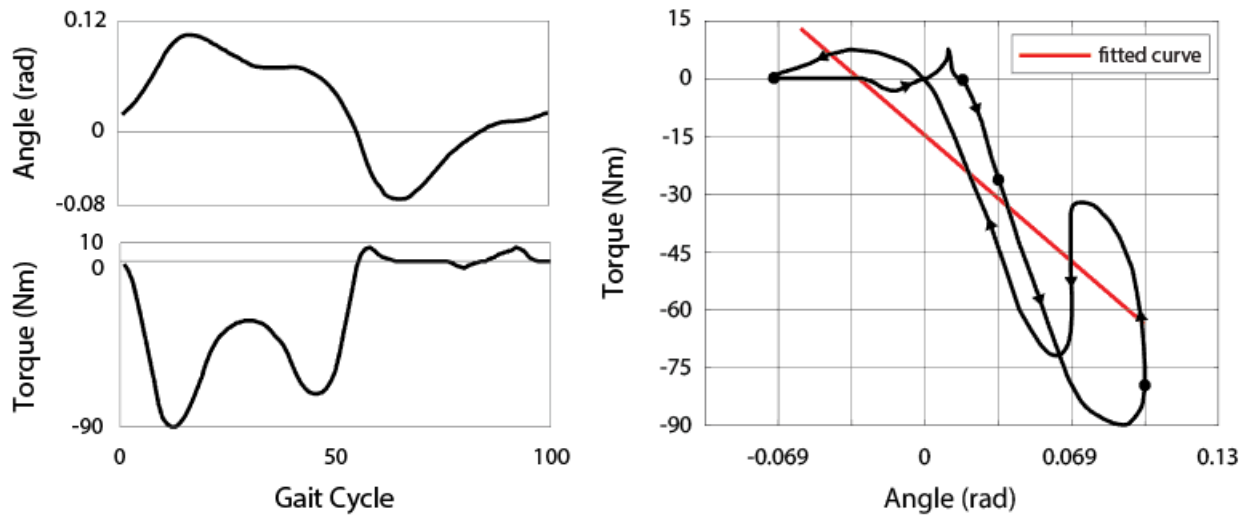


Figure 5.4: Frontal torque-angle plot with parallel spring fitted by a linear regression, of the hip joint of a 75 Kg person.

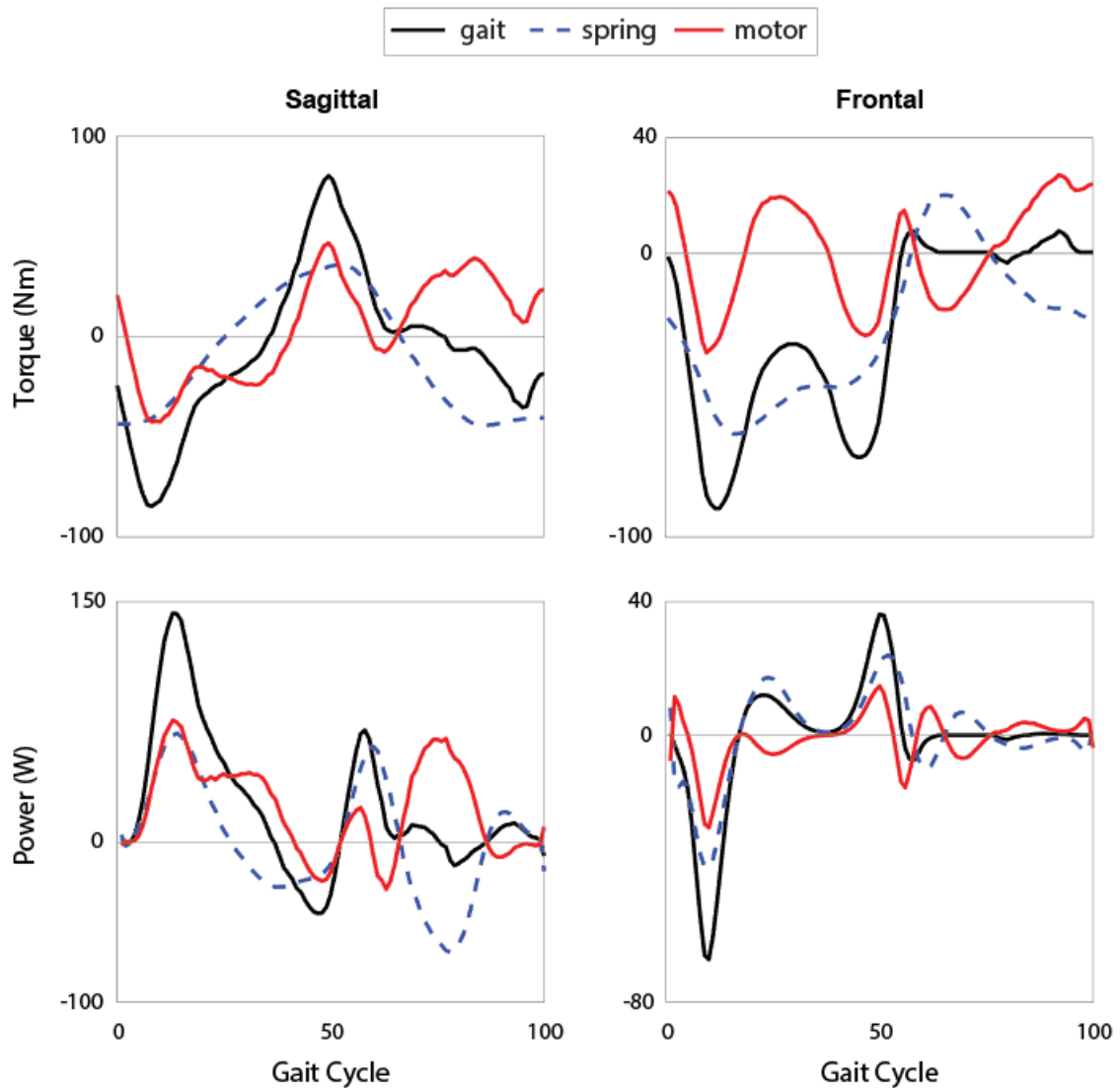


Figure 5.5: Sagittal and frontal torque and power curves after the fitting of the parallel spring. The biological curves are presented in black, the spring curves in blue and the motor curves in red.

In this case, the torques are the parameters altered by the spring and the power requirements are, as a consequence, different as well. The peak and RMS torques and power demands of the PEA at the sagittal and frontal planes are presented in Table 5.4. Once again, a considerable reduction is appreciated for both planes, although this time, not close in magnitude.

Table 5.4: Predicted torque and power effects of the implementation of a PEA at the sagittal and frontal planes of the hip for a 75 Kg person.

Variable	Sagittal			Frontal		
	Direct Drive	PEA Motor	Change (%)	Direct Drive	PEA Motor	Change (%)
Peak torque (Nm)	87.8	44.9	48.8	90	35.3	60.7
RMS torque (Nm)	40.8	25.6	37.2	41.2	18.1	56.1
Peak power (W)	142.7	76.1	46.67	67.1	27.7	58.7
RMS power (W)	49	35.36	27.83	17.7	7.6	60

The only reference to compare this finding with is [35]. They found a reduction of similar magnitude but higher for the sagittal plane compared to the frontal plane, which is the opposite to the ones here found. However, the real concern is the comparison with the capabilities presented by the SEA. The percentages of change in peak and RMS power with a parallel spring are higher than the ones presented by the SEA. Additionally, the PEA reduced demands of torque, which are valuable to the dimensioning of the motor. Therefore, the PEA is considered more suitable for this application. A final question related to compliance is to analyze if the addition of a series elasticity to the PEA might further improve the energy efficiency.

5.2.3 SEA for the PEA

Using the optimal parallel spring stiffness and residual torques that the motor has to deliver, the optimization process for series elastic actuation (see Section 5.2.1) was performed. Figure 5.6 presents the peak motor power as a function of the spring stiffness to determine the optimal stiffness.

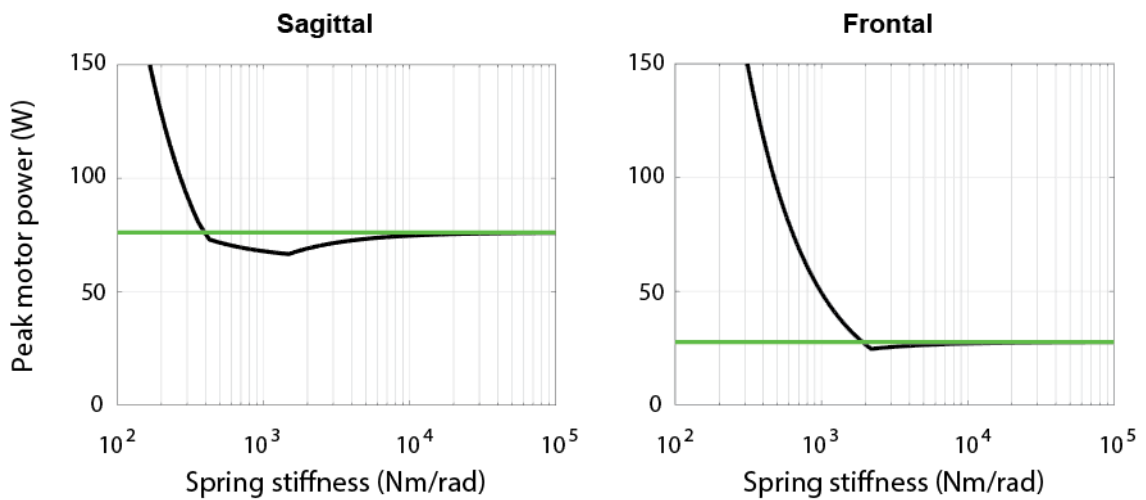


Figure 5.6: Peak motor power of the PEA hip joint of a 75 Kg person, in terms of an SEA stiffness variation. In green the original PEA peak power. In the sagittal plane peak value decreased from 76.1 W to 66.6 W. In the frontal plane peak value decreased from 27.7 W to 24.7 W.

In this case, even though there was an optimal stiffness, the percentages of change were rather small compared to previous results by the SEA and PEA, as presented in Table 5.5. Considering that this peak power reduction would only be reduced with a very fine tune of the spring and the mounting and control challenges in the implementation of an SEA, the results were not found advantageous.

Table 5.5: Predicted optimal SEA for PEA stiffness for minimizing peak power at the sagittal and frontal planes of the hip for a 75 Kg person.

Variable	Sagittal	Frontal
SEA Stiffness (Nm/rad)	1489	2188
Peak change (%)	12.5	10.8

The decision is then to only include a **spring in parallel** to the direct drive. Additionally, another decision is taken at this point. So far, both DoF (sagittal and frontal) have been considered in the analysis as it was suggested in Section 2.6. Nevertheless, this proposal being the first approach towards an active compliant polycentric module, it is considered more appropriate to start by implementing only the sagittal DoF. The inclusion of the frontal plane will remain for future studies when the addition of another DoF is pertinent. The design of the spring in a polycentric configuration is not a straightforward process as there is not a fixed center of rotation, however, a pre-dimensioning of the spring is now explored.

5.2.4 Pre-dimensioning of the Parallel Spring

There are multiple ways to implement a parallel spring as the one selected for this prosthetic joint. Even though this deserves a very detailed analysis, a first approach is by means of torsion springs. This type of springs is initially selected as it can be easily placed between two links to provide a torque proportional to the angle in between. Additionally, since most manufacturers offer custom-made torsion springs, by calculating the main parameters of the spring, there is no need to depend on catalogs. Following the procedure in [36], the goal is to achieve a torsion spring constant K which matches the optimization torsion spring constant in Table 5.3. The torsion spring constant K (in Nmm/rad) is calculated as

$$K = \frac{d^4 E}{64 D_m n_w} \quad (5.10)$$

where d is the wire diameter (in mm), E is the young modulus of the spring material (in N/mm^2 , e.g. $190000 N/mm^2$ for stainless steel), D_m is the coil diameter (in mm) and n_w is the number of active coils; as presented in Figure 5.7.

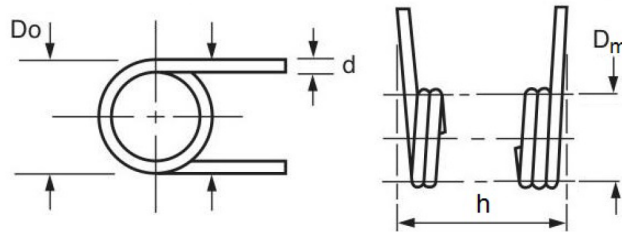


Figure 5.7: Torsion spring parameter definition. Image adapted from [71].

An optimal design for this spring is a compact one, meaning that both the torsion spring height (h in Equation 5.11) and the outer diameter (D_o in Equation 5.12) should be small.

$$h = n_w d \quad (5.11)$$

$$D_o = D_m + d \quad (5.12)$$

Furthermore, the spring must be able to withstand the extension it is subjected to. The strength is checked according to European Standard EN-10270-1-NS (steel wire for mechanical springs). There, the tensile strength for a stainless steel torsion spring is

$$R_m = 0.85 \cdot (2220 - 820 \cdot \log(d)) \quad (5.13)$$

And the permissible bending stress σ_{max} (in N/mm^2) is taken at $0.7 R_m$

$$\sigma_{max} = 0.7 \cdot 0.85 \cdot (2220 - 820 \cdot \log(d)) \quad (5.14)$$

Which must be higher than actual bending stress σ_b (in N/mm^2) found as

$$\sigma_b = \frac{32M}{\pi d^3} \quad (5.15)$$

where M is the moment acting on the spring (in Nmm).

Following the previous reasoning, three different alternatives of stainless steel springs were calculated and are presented in Table 5.6.

Table 5.6: Torsion spring design.

Variable	Spring 1	Spring 2	Spring 3
Wire diameter d (mm)	7	7.5	8
Mean diameter D_m	30.7	40.5	52.4
Outer diameter D_o	37.7	48	60.4
Number of active coild n_w	2	2	2
Height of spring h (mm)	14	15	16
Spring constant K (Nm/rad)	116.09	115.97	116.03
PEA spring constant K (Nm/rad)	116		

This pre-dimensioning of the parallel spring is intended to open a conversation around the spring design of the module, even though more detailed analyses regarding its location must be considered. Mainly, because torsion springs, like the ones suggested here, act along one axis of rotation and in this design a polycentric design was adopted; which alters the linear elastic behavior of the spring. The next step in this ‘component selection’ process is to determine the suitable motor and gearbox given the modified specifications for the PEA.

5.2.5 Motor and GearBox

Given the task that the actuator has to perform, the motor and gearbox have to be of high quality and reliable. To select the appropriate ones, Maxon Motors was chosen as the first company as their products are well known for having high quality and efficiency and in their catalog there are many alternatives. In a case like this one, where one needs high torques at low speeds, the suggestion is to (i) start by looking for the gearhead. Then, (ii) narrow down the list of articles by selecting the preferred product range and parameters. Once the gearbox is selected, (iii) one can combine assembled gearhead-motor-sensor combinations from the Maxon modular system and select matching controllers and accessories. Since many alternatives may be feasible, iterations over those steps is recommended until finding the best solution.

The main parameter for the gearbox search was the “Rated torque”, which is the maximum torque load that can be applied to the output shaft for continuous operation under ideal installation conditions. Meaning that if the torque load exceeds this rated value, the service life may be compromised. The best fit was the family of Planetary Gearhead GP 52 C (ratios from 43:1 to 936:1) with torque of 30 Nm. In this family two main groups of characteristics are presented and in Table 5.7 they are compared.

Table 5.7: Parameters of possible gearbox for prosthetic hip sagittal actuation.

Variable	GP 52 C (43:1 to 156:1)	GP 52 C (186:1 to 936:1)
Max cont. torque (Nm)	30	30
Max inter. torque (Nm)	45	45
Max. cont. speed (rpm)	6000	6000
Max. inter. speed (rpm)	6000	6000
Max. efficiency (%)	75	68
Weight (Kg)	0.77	0.92
Total length (mm)	78.5 (+29 of pin)	92 (+29 of pin)

For both groups, a search for a pairing motor was then conducted. After careful consideration that both the maximal speed (30 rpm, according to Table 4.2) and the continuous and peak torques (25.6 and 44.9 Nm, respectively according to Table 5.4) were achieved by the system, the more suitable combination was found for the gearbox of ratio 113:1. The motor found was the EC 60 flat brushless motor with Hall sensors. Its main parameters are: (i) dimensions; diameter of 60 mm and length of 38 mm, (ii) nominal voltage of 24 V, (iii) power of 150 W and maximum torque of 401 mNm. The motor already comes with the Hall sensors, which is a magnetic field sensor used to determine rotor’s position and speed, and thus, include the first of the solutions of the “sensing” function (see Section 4.4). Additionally, if it is desired, an encoder from the MILE family (512 - 4096 counts per turn) can also be integrated into the motor. A summary of the results for the PEA set-up is presented in Table 5.8.

Table 5.8: Summary of the results for the PEA set-up. * Dimensions without the MILE encoder are 1mm less. TD: To be defined.

Req	EC 60 Maxon Motor	GP 52 C (113:1)	+ Custom ratio	Total	+ Extra
Pw: 76 W	150 W	< 170 W	Miter (1:1)	✓	Driver
Trq: 45 Nm	Nom: 0.401 Nm	45.3 Nm		✓	K =
Vel: 30 rpm	Nom: 3480 rpm	30.8 rpm		✓	116 Nm/rad
Mass: $\approx$ 4 Kg	0.35 Kg	0.77 Kg	TBD	$\approx$ 1.12 Kg	
H: $\approx$ 205 mm	41.8 mm*	78.5 mm	TBD	$\approx$ 120.3 mm*	
$\varnothing \approx$ 115 mm	60 mm	52 mm	TBD	$\approx$ 60 mm	

Furthermore, several electronics can be added to the selected combination. A list of the ones recommended by Maxon are displayed in Figure 5.8. The ESCON modules are powerful 4-quadrant PWM servo controllers that provide power and highly precise speed control for Maxon motors, in a compact design; and the EPOS modules are digital positioning controllers by Maxon. They are suitable for permanent magnet-activated motors plus encoders with a range of 1 to 1050 W continuous output power. The data-sheets of the motor and gear box are presented in the Appendix B of this document.

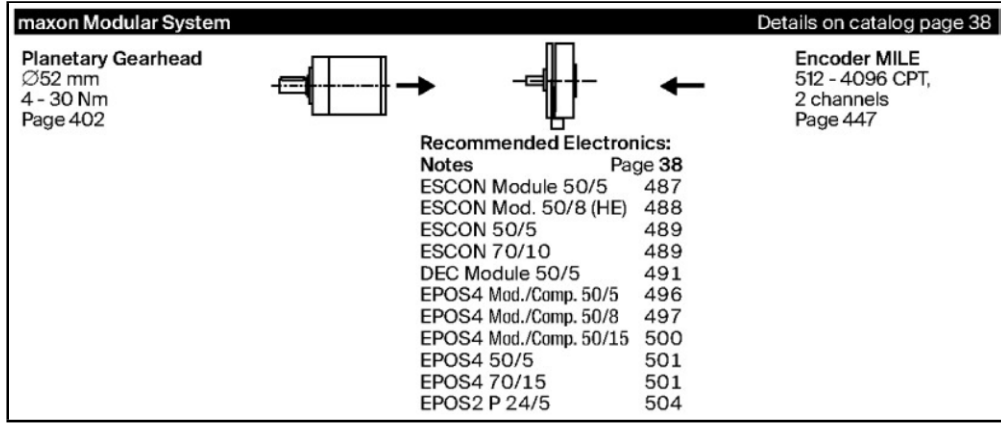


Figure 5.8: Maxon components for PEA: gearbox, motor, encoder and electronics. Image from the online catalog of Maxon.

Even though the characteristics of the actuator are reasonable so far, the dimensions are close to the established limits and, consequently, are conditioning the assembly in the space (see Section 4.3.2). Therefore, other companies as Faulhaber were considered but no satisfactory combinations of motor and gearboxes were found for the application. Finally, another alternative was presented: the motor produced by Harmonic Drive, in their FHA-Mini Rotary Actuator series, which was implemented in [49] and presented in Section 3.3. This actuator delivers high torque and combines a high precision Harmonic Drive gear with a brushless servo motor and an incremental encoder to provide high resolution velocity and position feedback (once again, included the “sensing” function of Section 4.4). Given the maximal torque of the actuator, an extra gear of ratio 2:1 should be added as presented in Table 5.9. As complementary electronics for this motor, the driver HA-680-6-24 provides position, speed and torque control. The main pages of the data-sheet of this actuator are presented in the Appendix B of this document.

Table 5.9: Summary of the results for the PEA set-up. TBD: To be defined.

Req	FHA-14C-100-E200E	+ Custom ratio	Total	+ Extra
Pw: 76 W	-	Helic. (2:1)	✓	Driver
Trq: 45 Nm	28 Nm	56 Nm	✓	K =
Vel: 30 rpm	60 rpm	30 rpm	✓	116 Nm/rad
Mass: $\approx$ 4 Kg	1.2 Kg	TBD	$\approx$ 1.2 Kg	
H: $\approx$ 205 mm	75 mm	TBD	$\approx$ 75 mm	
$\varnothing \approx$ 115 mm	75 x 66 mm	TBD	$\approx$ 66 mm	

Both the Maxon and Harmonic Drive systems will be considered in the assembly of the components. The configuration needed for each of them, regarding the orientations of the axis of rotation and the assembly of the components, will be explored to finally select one of them.

5.3 Conceptual Design

The conceptual design is presented in terms of the kinematic schematics for the frontal and sagittal planes. The objective of this section is to suggest an alternative to the placement of the main components in the module. With that in mind, the dimensions of other components in the design are considered approximations to guide the decisions towards the best assembly. Reminding the space volume definition of the prosthetic hip module defined in Section 4.3.2, a

cylindrical-leg-shape, a circumference of diameter $\varnothing 115$ mm and heights 212 and 157 mm (two boxes corresponding to attachment point and body of the joint) were selected. The geometry produced is represented in Figures 5.9 to 5.11 in the red dashed lines.

The first kinematic schematic, in Figure 5.9, corresponds to the Maxon PEA defined in Table 5.8. Given the dimensions of the motor and gearbox, the orientation of the actuator can only be set vertically, which then implies the addition of the bevel gear mentioned on the Table. Since the actuator can provide all the torque requirements, the ratio of the gear is only of 1:1, which is known as a ‘miter gear’. To achieve a polycentric design as it was intended, a timing belt (again of ratio 1:1) with the top pulley fixed was added, placed as close as possible to the axis of rotation. To ensure proper movement of the joint, free-rotating segments of fixed length were added (i) between the same two axis of the timing belt and parallel to it and (ii) from the attachment point to the socket to the chassis. Inspired by the morphology of the Helix3D joint, a central segment was used (i) and two lateral segments were used (ii), to provide symmetry. The spring is depicted, at this point, not to forget that it should be included in the assembly, however many possible placements can be suitable and it is a question of further analysis (e.g. by means of a torsional spring as shown, or by implementing a linear equivalent).

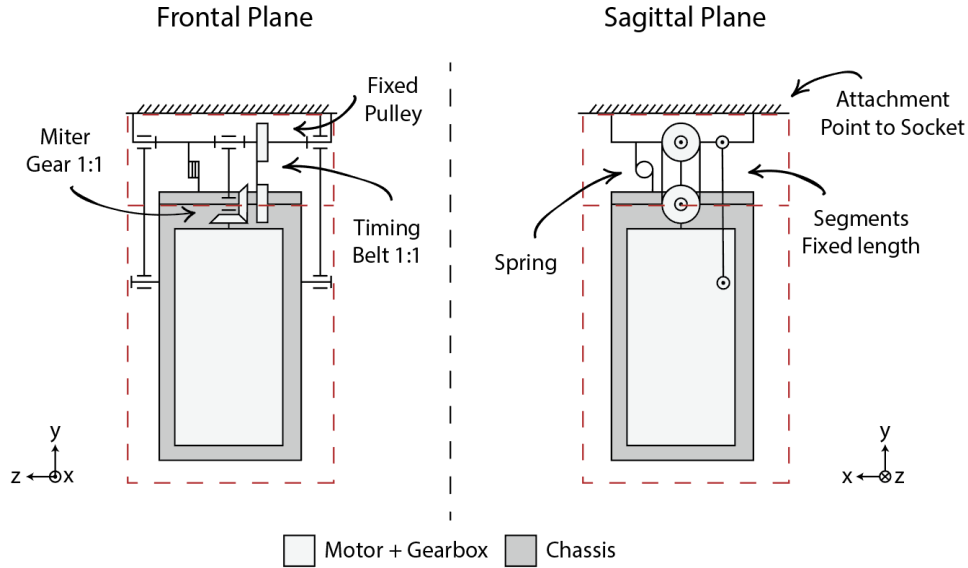


Figure 5.9: Kinematic schematic for the Maxon actuator.

One can notice that this configuration is nearly close to the bounds, mainly on its height, and yet the length of the pulleys in the timing belt and the gears in the miter gear do not have appropriate dimensioning. Not to count the space that should be considered to place the additional electronics and other elements of the modules (crural segment, rotational adaptors, etc). Performing a brief consultation on catalogs for miter gears [72](data-sheet in the Appendix B) and for metal timing pulleys [73], the gears and timing pulleys that could fit the requirements for the torque would have the approximate dimensions illustrated in Figure 5.10. In this assembly, it is pretty evident that the dimensions are out of the limits and do not satisfy the compact design that is looked for. Therefore, the Maxon combination is not contemplated as part of the design.

On the other hand, the kinematic schematic in Figure 5.11, corresponds to the Harmonic Drive PEA defined in Table 5.9. Since the dimensions of this actuator do not condition its orientation in the designated space, the axis of rotation is set to be horizontal. This time, however, the actuator can not provide all the torque requirements, and this is why a timing belt of ratio 2:1 with the top pulley fixed was added, as in the previous design.

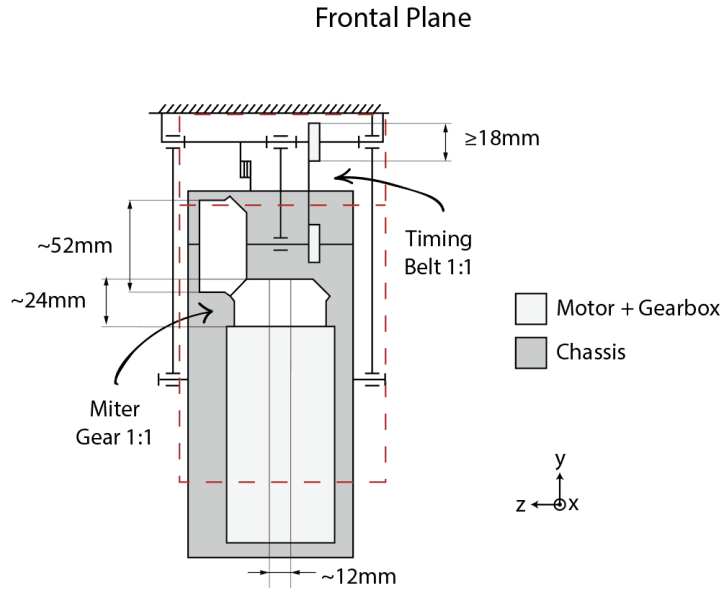


Figure 5.10: Kinematic schematic for the Maxon actuator.

Once again, free-rotating segments of fixed length were added, inspired by the morphology of the Helix3D joint. Although this time, the actuation is placed in the more distal segment. One can notice that in this configuration there is no problem regarding the limits of the device. On the contrary, since there is no need to add a bevel gear and the actuator is much more compact than the other one, there is still room to fit the additional electronics and the other elements of the modules (crural segment, rotational adaptors, etc).

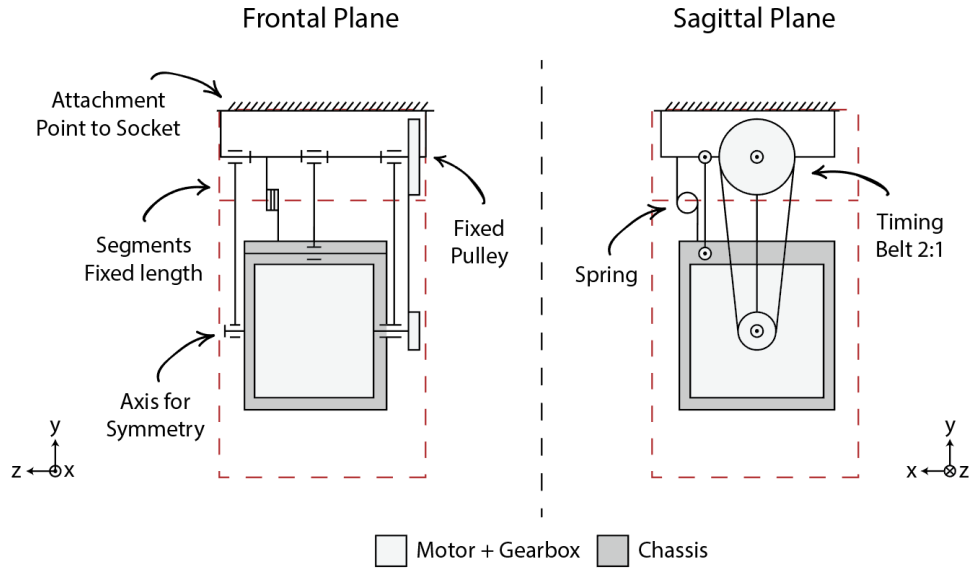


Figure 5.11: Kinematic schematic for the Harmonic Drive actuator.

In order to better understand how the actuation would be done, a schematic of the sagittal movement of the simplified joint module is presented in Figure 5.12.A. The movement is compared to a representation of the movement performed by the Helix3D joint, presented in Figure 5.12.B. The assembly offers polycentric-like advantages and in terms of space it is a very flexible approach as it allows the inclusion of extra components. Both the dimensions and the weight are below the expected requirements at this stage, which is a good sign towards next steps of the design.

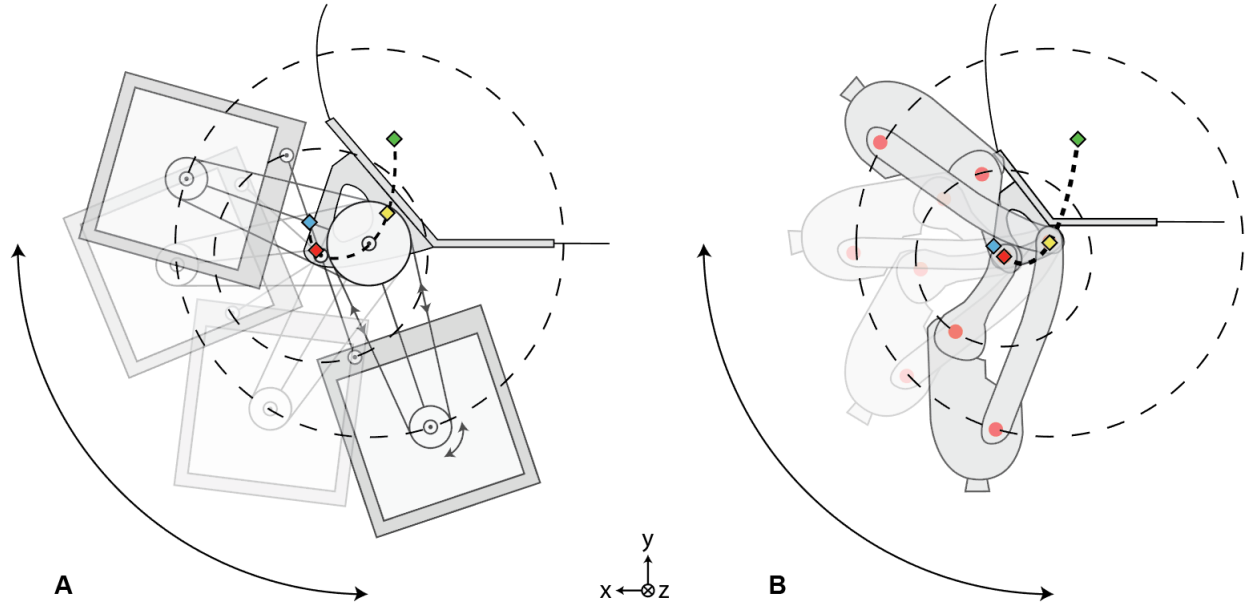


Figure 5.12: Schematic of the movement in the sagittal plane. A. Proposal for the Harmonic Drive actuator. B. Simplification of the Helix3D. Diamonds in color represent the instant center of rotation trajectory.

To conclude the mechanical implementation proposed in this document, a preliminary technical drawing of the elements considered so far is presented in Figure 5.13. The dimension of the main motor shaft was set according to the characteristic of the hollow configuration of the motor (data-sheet in the Appendix B). The other shafts were selected to have the same diameter, for the sake of simplification. The dimensions of the bearings and the timing belt are merely illustrative and were selected without the appropriate calculation, but following standard measures of those components in their respective catalogs and making sure they fitted the selected shafts. Nevertheless, the 2:1 ratio of the pulleys is respected and the dimensions of the pulleys correspond to real-life proportions. Other dimensions not specified are open to upcoming designs, provided they respect the boundaries set (shown, once again, by the red dotted box).

In the Figure, four sections are worth commenting. In (a), the discontinuity convention lines were drawn to show that, even though in this frontal plane view the shaft may be perceived as one, there are actually two. Something that can be corroborated if one looks at the sagittal plane view in Figure 5.11. In (b), the design of a shaft to attach the central segment was done by means of two passing-through pieces. The shaft can be then fixed with screws or other means, which are not depicted on the drawing. In (c), some space is provided to access the motor and the cabling. Finally, in (d), the way in which the motor is fixed to the chassis is presented. To avoid an hyperstatic system, only the pulley is fixed to the shaft. The segment of fixed length is attached by the bearing to the chassis.

The system here describe constitutes the preliminary technical drawing of the system. It intends to be a baseline for future designs where other components are included.

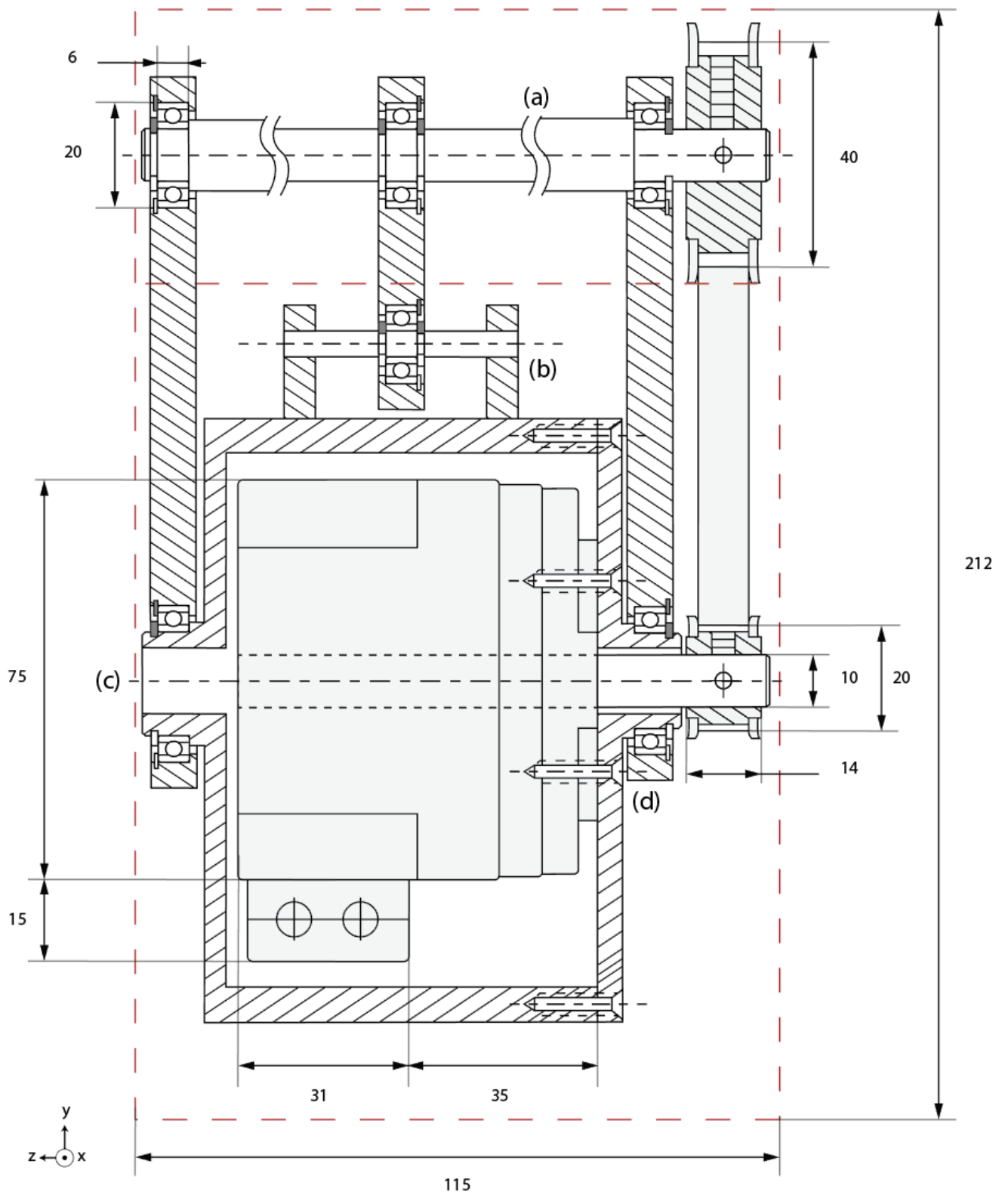


Figure 5.13: Frontal view of the preliminary technical drawing of the proposed prosthetic hip joint.

Chapter 6

Conclusions and Future Work

The goal of this thesis was to start the design of an active module of the hip joint of a complete-leg prosthesis for disarticulated amputees. A literature study on the functional and psychological repercussions of this type of amputation, and the existing technologies developed specifically for it, revealed the importance of defining the right requirements for the active prosthetic hip joint. Therefore, a significant portion of the time reserved for this thesis was dedicated to specifying and understanding these requirements.

Starting by a detailed analysis of the hip joint and its kinematics, the complexity and importance of this human joint was shown. Afterwards, its role in the most fundamental activity reflecting human locomotion: walking, was analyzed. The kinematics and kinetics of walking and considerations regarding the torque-angle diagrams of each of the planes of movement of the hip were presented and compared with other activities as stair climbing and descending. Based on the phenomenon observed and the study of the “compliance” as a highly important concept of the human body, the characteristics of healthy locomotion were understood. The sagittal DoF is without a doubt the main DoF to consider in the design of a hip joint, even though the frontal DoF cannot be overlooked in such a design.

In order to understand the biomechanics of the hip in disarticulated amputees and, thus be able to compare them to sound patients, another literature study was conducted. A big effort was done to gather the existing publications and experiences with this population into one document, something that had not been done before, to the knowledge of the author. The Helix3D prosthetic joint by Ottobock was found to be the best configuration so far achieved for a passive joint thanks to its polycentric “3D” mechanism, hydraulic damping, and elastic elements. Additionally, for the first time, a validated and suited questionnaire was recognized to functionally assess hip disarticulation (the LCI-5 with supplementary activities). Finally, the only robotic prosthesis for hip disarticulation published to date was studied. With all the context set, the inputs and characterization of both prosthetic joints (the passive Helix3D and the robotic), and following the Cross Model for engineering design a *polycentric and compliant electromechanical prosthetic hip joint, fixed anterior to the socket* was chosen to be the most appropriate and beneficial.

As this work constitutes the first development around the subject, the design was restricted to the sagittal plane only. The inclusion of a spring in parallel to the actuator (compliant actuation) showed a reduction of the peak torque in almost 50 %, the RMS torque in 37 %, the peak power in 76 % and the RMS power in around 28 % compared to a direct drive configuration. The motor and gearbox selection for the torque and power requirements gave two candidates: a Maxon combination and an Harmonic Drive mini-actuator. Not considering economical constraints, the conceptual design was only feasible with the second one. The assembly proposal was based on (i) a timing belt with the proximal pulley fixed, (ii) a four-bar mechanism and (iii) a torsional spring which deserves further analysis. The chosen configuration is compact and

offers space for the placement of electronic components that were discussed in this document.

As it was presented, the design of an active hip joint exo-prosthetic module for disarticulated amputees is a work that demands careful consideration and analysis. In this thesis, the design requirements were studied with a lot of detail before actually constructing a primary proposal for the design of the joint. However, despite the author's effort, there are several elements and concepts that should be studied and integrated in the design. This means that there is still work to be done. The most important and perhaps most challenging issues are the following.

The first one is the spring placement. As presented, the parallel spring defined in the PEA of the hip joint can be achieved by different means. The most common, given the nature of the device, would be by means of a torsion spring. Even though calculations for suitable springs were made, the placement of the torsion spring in the polycentric design might condition its elastic behavior and, therefore, it is a subject that deserves further study.

The second one is the fine dimensioning. The last chapter built towards the design of the hip prosthesis and selected the main functional components; however, the dimensioning of the module was not fully achieved and still needs to be tuned. Components as the bearings, the timing belt and the chassis with the connecting segments should be carefully designed to fulfill the load and actuation requirements proposed. Moreover, many other electronic components needed for the actuation and control of the device have to be studied and selected. The motor driver seems to be the most obvious one. Nevertheless, electronics boards and controllers, the addition of other sensors as presented in Section 4.4 and the power supply in terms of the batteries have to be studied in the following steps of the design.

Finally, the testing and the integration to full leg prosthesis. The design was based on several assumptions (see Section 5.2) and even though there are designing challenges to overcome, further tests with a prototype can reveal several focus of attention. One corresponds to whether the assembly of the components allows a smooth movement comparable to the testimonies given by patients with the Helix3D joint. Another is the effect of the addition of the PEA in terms of the placement and design of the spring. If the results continue to be satisfactory, in the future the module could then be added to a full leg prosthesis to evaluate the performance.

In conclusion, the preliminary design of an active prosthetic hip joint for disarticulated amputees was done. It incorporates several principles of energy transfer and energy storage, compliance, balance and comfort that have shown to have a great impact on this amputees. Even though there is still a long way to go to reach the ambitious goal of designing this hip module that can reproduce the biomechanics of a healthy leg during daily-life locomotion, with this thesis, the author hopes to have provided valuable insights that will one day lead to attaining this admirable goal.

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Appendix A

Principal Muscles Actuating the Hip Joint

A brief overview of the muscles actuating the hip joint is illustrated in Figure A.1. As depicted, many large and powerful muscles surround the hip, producing the forces needed to move the lower extremity through a variety of planes or to propel the body in nearly any direction. Other relatively smaller muscles stabilize the hip and help fine-tune various actions of the lower extremity. Although the principal contributors among these muscles will be discussed by DoF, in reality the ultimate function of the lower extremity is absolutely dependent of the interaction between all the musculature.

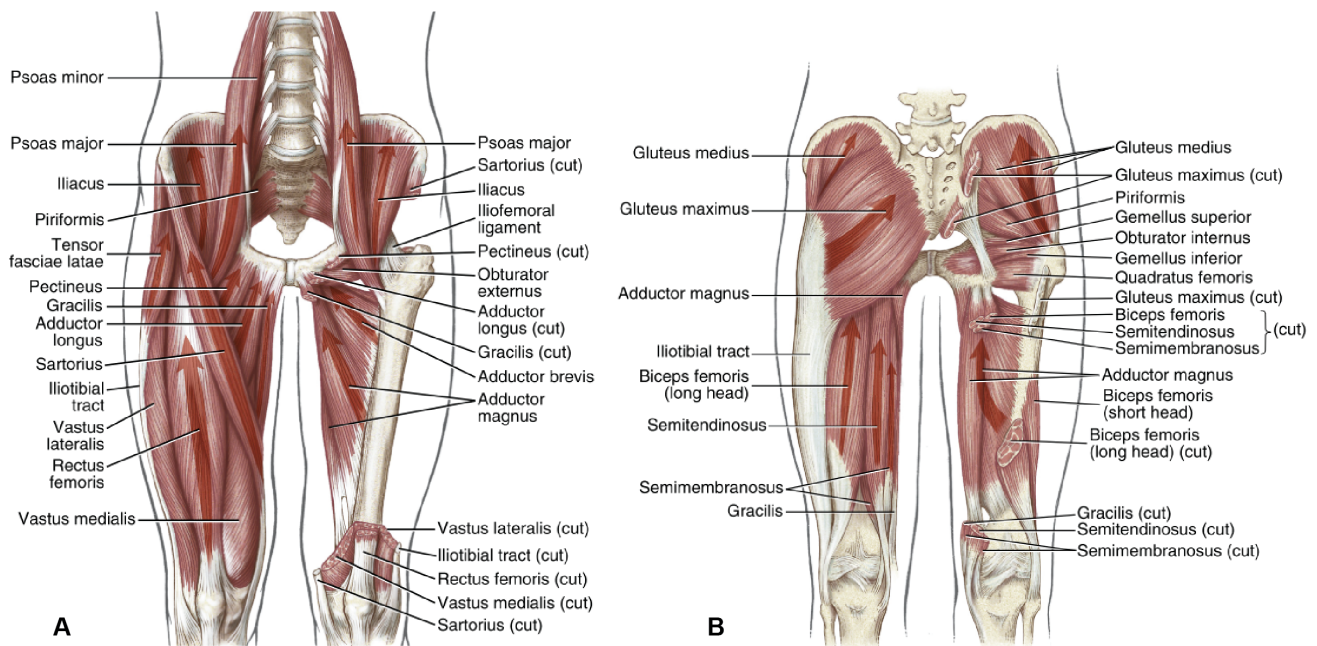


Figure A.1: A. Many muscles of the anterior hip region. The right side of the body shows the flexor and adductor muscles. On the left side of the body the adductor brevis and adductor magnus are exposed. B. Posterior view of the muscles of the hip. Images from [22].

Flexors and Extensors

The primary hip flexors are the iliopsoas, rectus femoris, sartorius, and tensor fasciae latae. When analyzed from the anatomic position, most of the hip adductor muscles also flex the hip. All hip flexor muscles have a line of pull that is anterior to the medial-lateral axis of rotation of the hip. The The sartorius is the longest muscle in the body and contributes to the hip flexion,

external rotation, and abduction, as well as flexion of the knee. The primary hip extensors are the gluteus maximus and the hamstrings (i.e., the long head of the biceps femoris, the semitendinosus, and the semimembranosus). The extensor head of the adductor magnus is also considered a primary hip extensor. This powerful group of hip extensors is used for functional activities involving upward and forward propulsion of the body, such as jumping, running, stair climbing, and transitioning from sitting to standing [22].

Abductors and Adductors

The primary hip abductor muscles include the gluteus medius, gluteus minimus, and tensor fasciae latae; the piriformis, sartorius, and superior fibers of the gluteus maximus are considered secondary hip abductors. The most frequent demands placed on the hip abductors occur while walking. The right hip abductors must supply an adequate contraction force to keep the pelvis from “falling into the space” created by the advancing left leg (known as the swing face of walking). Most studies indicate a total hip joint reaction force of three times body weight when standing on one limb. This total force is created by (1) the contraction force of the hip abductor muscles (which generates a force two times body weight), and (2) body weight itself. The primary hip adductors are the pectineus, adductor longus, gracilis, adductor brevis, and adductor magnus. The primary function of this muscle group is, of course, to create adduction torque, bringing the lower extremity toward the midline. From the anatomic position, the adductors are also considered hip flexor muscles [22].

Rotators

The primary external rotators of the hip are the gluteus maximus, the sartorius, and the six short external rotators: piriformis, gemellus superior, obturator internus, gemellus inferior, obturator externus, and quadratus femoris. The primary internal rotators of the hip are the gluteus medius (anterior fibers), the gluteus minimus and the tensor fasciae latae. Contrary to the external rotators, most internal rotator muscles have a near-vertical orientation, at least from the anatomic position. For this reason, no individual muscle is capable of generating a very strong internal rotation torque [22].

Appendix B

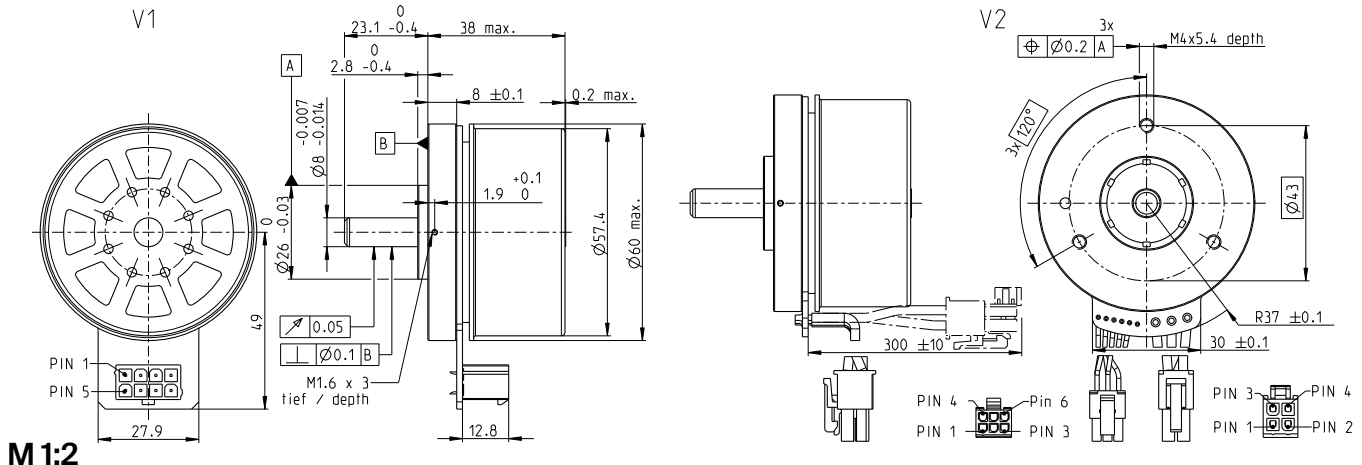
Data-sheets

This Appendix contains the data-sheets of the components used or referenced in the design. The following is the order in which they are presented:

- Motor data-sheet: Maxon EC 60 flat brushless motor
- Gearhead data-sheet: Maxon GP 52 C
- Motor data-sheet: Harmonic Drive FHA-14-C-100-E200E
- Bevel Gear data-sheet: J Series miter gear catalog sheet

EC 60 flat Ø60 mm, brushless, 150 Watt

Open Rotor



M 1:2

- Stock program
- Standard program
- Special program (on request)

Part Numbers

V1 with Hall sensors	625857	625858	625859
V2 with Hall sensors and cables	647693	647694	647695

Motor Data

Values at nominal voltage

1 Nominal voltage	V	12	24	48
2 No load speed	rpm	3760	4300	4020
3 No load current	mA	815	497	224
4 Nominal speed	rpm	2990	3480	3230
5 Nominal torque (max. continuous torque)	mNm	378	401	437
6 Nominal current (max. continuous current)	A	12*	7.25	3.63
7 Stall torque ¹	mNm	3340	4300	4870
8 Stall current	A	111	81.9	43.2
9 Max. efficiency	%	83.8	85.2	86.3

Characteristics

10 Terminal resistance phase to phase	Ω	0.108	0.293	1.11
11 Terminal inductance phase to phase	mH	0.0911	0.279	1.28
12 Torque constant	mNm/A	30	52.5	113
13 Speed constant	rpm/V	318	182	84.8
14 Speed/torque gradient	rpm/mNm	1.14	1.01	0.837
15 Mechanical time constant	ms	9.68	8.6	9.1
16 Rotor inertia	gcm ²	810	810	810

Specifications

Thermal data

17 Thermal resistance housing-ambient	1.94 K/W
18 Thermal resistance winding-housing	1.48 K/W
19 Thermal time constant winding	16.1 s
20 Thermal time constant motor	69.9 s
21 Ambient temperature	-40...+100°C
22 Max. winding temperature	+125°C

Mechanical data (preloaded ball bearings)

23 Max. speed	6000 rpm
24 Axial play at axial load < 12.0 N	0 mm
24 Axial play at axial load > 12.0 N	0.14 mm
25 Radial play	preloaded
26 Max. axial load (dynamic)	12 N
27 Max. force for press fits (static) (static, shaft supported)	170 N
27 Max. force for press fits (static) (static, shaft supported)	800 N
28 Max. radial load, 5 mm from flange	112 N

Other specifications

29 Number of pole pairs	7
30 Number of phases	3
31 Weight of motor	350 g

Values listed in the table are nominal.

Connection V1 V2 (sensors, AWG 24)

Pin 1	Hall sensor1	Hall sensor1
Pin 2	Hall sensor 2	Hall sensor 2
Pin 3	V _{Hall} 4.5...24 VDC	Hall sensor 3
Pin 4	Motor winding 3	GND
Pin 5	Hall sensor 3	V _{Hall} 4.5...24 VDC
Pin 6	GND	N.C.
Pin 7	Motor winding 1	N.C.
Pin 8	Motor winding 2	N.C.

Pin 1	V2 (Motor, AWG 16)
Pin 2	Motor winding 1
Pin 3	Motor winding 2
Pin 4	Motor winding 3
Pin 5	N.C.

Wiring diagram for Hall sensors see p. 49

Connector	Part number
Molex	46015-0806
Molex	43025-0600
	39-01-2040

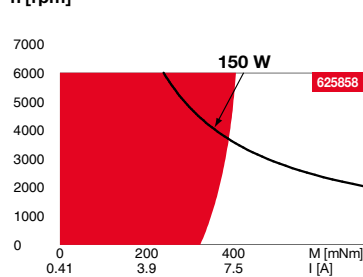
Connection cable for V1

Universal, L = 500 mm	339380
to EPOS4, L = 500 mm	354045

¹Calculation does not include saturation effect (p. 61/168)

Operating Range

n [rpm]



Comments

Continuous operation

In observation of above listed thermal resistance (lines 17 and 18) the maximum permissible winding temperature will be reached during continuous operation at 25°C ambient.
= Thermal limit.

Short term operation

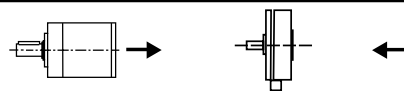
The motor may be briefly overloaded (recurring).

Assigned power rating

maxon Modular System

Planetary Gearhead

Ø52 mm
4 - 30 Nm
Page 402



Details on catalog page 38

Recommended Electronics:

Notes Page 38

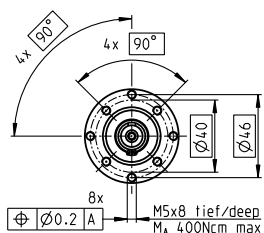
ESCON Module 50/5	487
ESCON Mod. 50/8 (HE)	488
ESCON 50/5	489
ESCON 70/10	489
DEC Module 50/5	491
EPOS4 Mod./Comp. 50/5	496
EPOS4 Mod./Comp. 50/8	497
EPOS4 Mod./Comp. 50/15	500
EPOS4 50/5	501
EPOS4 70/15	501
EPOS2 P 24/5	504

Encoder MILE
512 - 4096 CPT,
2 channels
Page 447

*In combination with EPOS4 positioning controllers, the connector technology limits the nominal current (max. continuous current load) is limited to 11 A.

Ceramic Version

gear



M 1:4

Planetary Gearhead	straight teeth			
Output shaft	stainless steel			
Bearing at output	preloaded ball bearings			
Radial play, 12 mm from flange	max. 0.06 mm			
Axial play at axial load	< 5 N	0 mm		
	> 5 N	max. 0.3 mm		
Max. axial load (dynamic)				200 N
Max. force for press fits				500 N
Direction of rotation, drive to output				=
Max. continuous input speed				6000 rpm
Recommended temperature range				-15...+80°C
Extended range as option				-40...+100°C
Number of stages	1	2	3	4
Max. radial load, 12 mm from flange	420 N	630 N	900 N	900 N

maxon Modular System												
+ Motor	Page	+ Sensor	Page	Brake	Page	Overall length [mm] = Motor length + gearhead length + (sensor/brake) + assembly parts						
EC 45, 250 W	231			AB 28	520	200.5	216.5	230.0	230.0	243.5	243.5	243.5
EC 45, 250 W	231	HEDL 9140	478	AB 28	520	217.5	233.5	247.0	247.0	260.5	260.5	260.5
EC-max 40, 120 W	243					137.1	153.1	166.6	166.6	180.1	180.1	180.1
EC-max 40, 120 W	243	MR	464			153.0	169.0	182.5	182.5	196.0	196.0	196.0
EC-max 40, 120 W	243	HEDL 5540	475			160.5	176.5	190.0	190.0	203.5	203.5	203.5
EC-max 40, 120 W	243			AB 28	518	171.5	187.5	201.0	201.0	214.5	214.5	214.5
EC-max 40, 120 W	243	HEDL 5540	475	AB 28	518	189.8	205.8	219.3	219.3	232.8	232.8	232.8
EC-i 52, 180 W	268					129.1	145.1	158.6	158.6	172.1	172.1	172.1
EC-i 52, 180 W	268	16 EASY/Abs.	449/453			142.8	158.8	172.3	172.3	185.8	185.8	185.8
EC-i 52, 180 W	268	16 RIO	466			142.8	158.8	172.3	172.3	185.8	185.8	185.8
EC-i 52, 180 W	268	AEDL 5810	469/470			151.9	168.9	181.4	181.4	194.9	194.9	194.9
EC-i 52, 180 W	268	HEDL 5540	473-477			151.9	168.9	181.4	181.4	194.9	194.9	194.9
EC-i 52, 200 W	269					159.1	175.1	188.6	188.6	202.1	202.1	202.1
EC-i 52, 200 W	269	16 EASY/XT/Abs.	450-454			172.8	188.8	202.3	202.3	215.8	215.8	215.8
EC-i 52, 200 W	269	16 EASY Abs. XT	456			173.3	189.3	202.8	202.8	216.3	216.3	216.3
EC-i 52, 200 W	269	16 RIO	467			172.8	188.8	202.3	202.3	215.8	215.8	215.8
EC-i 52, 200 W	269	AEDL 5810	469/470			181.9	198.9	211.4	211.4	224.9	224.9	224.9
EC-i 52, 200 W	269	HEDL 5540	473-477			181.9	198.9	211.4	211.4	224.9	224.9	224.9
EC 60 flat, 100 W	294					89.8	105.8	119.3	119.3	132.8	132.8	132.8
EC 60 flat, 100 W	294	MILE	446			90.8	106.8	120.3	120.3	133.8	133.8	133.8
EC 60 flat, 150 W	295					89.8	105.8	119.3	119.3	132.8	132.8	132.8
EC 60 flat, 150 W	295	MILE	446			90.8	106.8	120.3	120.3	133.8	133.8	133.8
EC 60 flat, 200 W	296					97.6	113.6	127.1	127.1	140.6	140.6	140.6
EC 60 flat, 200 W	296	MILE	446			98.6	114.6	128.1	128.1	141.6	141.6	141.6

FHA Mini Series

Servo Actuators



Harmonic Drive[®] actuator

P r e c i s i o n G e a r i n g & M o t i o n C o n t r o l

1-2 Ordering Code

■ Ordering Code (with Incremental Encoder)

FHA	-	8	C	-	50	-	US200	-	C	E	K	-	SP
1		2	3		4		5		6	7	8		9

1	Model		FHA-C Mini Series
2	Size		8, 11, 14
3	Design Version		C
4	Gear Ratio		30, 50, 100
5	Encoder Type and Resolution		US200 – Incremental Encoder with 14 wire 2000p/rev E200 – Incremental Encoder with 4 wire serial 2000p/rev
6	Connector (leads only)		Blank – Leads without connector C – Leads with connector for extension cable
7	Power Supply		E – 24VDC
8	Options	Leads	Blank – Side exit leads, right (of output side) K – Rear exit leads
		Cable	M1 – 1M Length Cable (motor & encoder) KM1 – Rear Exit 1M Length Cable (motor & encoder)
		Panel Mount	PM1 – Panel Mount Connectors, rear exit PM2 – Panel Mount Connectors, right exit PM3 – Panel Mount Connectors, left exit PM4 – Panel Mount Connectors, top exit
9	Special Specification		Blank – Standard Product SP ____ – Special Specification Code

■ Ordering Code (with Absolute Encoder or Dual Absolute Encoders)

FHA

-

8

C

-

50

-

12S17b

E

-

PM1

-

SP

1

2

3

4

5

6

7

8

1	Model		FHA-C Mini Series
2	Size		8, 11, 14
3	Design Version		C
4	Gear Ratio		30, 50, 100
5	Encoder Type and Resolution		12S17b - Single Absolute Encoder, Nikon A Single Turn (17 bit) & Multi-Turn (16 bit) SB14bSB14b - Dual Absolute Encoders, BiSS-C, bi-directional Motor Input Encoder (14 bit) & Gear Output Encoder (14 bit) NEW!
6	Power Supply		E – 24VDC
7	Options	Leads*	Blank – Leads without connector C – Leads with connector for extension cable
		Cable*	M1 – 1M Length Cable (motor & encoder)
		Panel Mount	PM1 – Panel Mount Connectors, rear exit PM2 – Panel Mount Connectors, right exit PM3 – Panel Mount Connectors, left exit PM4 – Panel Mount Connectors, top exit NEW!
8	Special Specification		Blank – Standard Product SP _____ – Special Specification Code

* only available with single absolute encoder.

1-4-2 Supply voltage 24VDC

Item			Size	FHA-8C			FHA-11C			FHA-14C		
				30	50	100	30	50	100	30	50	100
Maximum torque (Note) 2			Nm	1.8	3.3	4.8	4.5	8.3	11	9.0	18	28
Maximum speed			rpm	200	120	60	200	120	60	200	120	60
Torque constant			Nm/A _{rms}	0.8	1.3	2.7	0.8	1.3	2.6	0.8	1.4	2.9
Maximum current (Note) 2			A _{rms}	3.0	3.3	2.4	7.8	8.2	5.6	14.8	16.4	12.3
Allowable continuous current (Note) 2			A _{rms}	1.6	1.7	1.3	3.7	3.5	2.8	6	5.4	4.4
Power supply voltage			V	24VDC								
EMF constant			V/(rpm)	0.10	0.16	0.32	0.09	0.15	0.31	0.10	0.17	0.34
Phase resistance			Ω(20°C)	0.54			0.19			0.07		
Phase inductance			mH	0.22			0.11			0.06		
Number of poles				10								
Moment of inertia (Note) 3	INC	GD ² /4	kgm ²	0.0026	0.0074	0.029	0.0060	0.017	0.067	0.018	0.050	0.20
	ABS	GD ² /4	kgm ²	0.0026	0.0073	0.029	0.0062	0.017	0.069	0.019	0.054	0.215
	Dual ABS	GD ² /4	kgm ²	0.0036	0.0100	0.0399	0.007	0.019	0.077	0.019	0.052	0.206
Allowable moment load		Nm		15			40			75		
Moment Stiffness		Nm/rad		2 x 10 ⁴			4 x 10 ⁴			8 x 10 ⁴		
Output shaft resolution (Note) 4	INC		counts/rev	240,000	400,000	800,000	240,000	400,000	800,000	240,000	400,000	800,000
	Single ABS			3,932,160	6,553,600	13,107,200	3,932,160	6,553,600	13,107,200	3,932,160	6,553,600	13,107,200
	Dual ABS Motor			491,520	819,200	1,638,400	491,520	819,200	1,638,400	491,520	819,200	1,638,400
	Dual ABS Output			16,384	16,384	16,384	16,384	16,384	16,384	16,384	16,384	16,384
One-way positioning accuracy		arc sec		150	120	120	120	90	90	120	90	90
Input power supply voltage		V		24VDC								
Mass	INC	kg	0.40			0.62			1.2			
	ABS		0.50			0.75			1.3			
	Dual ABS		0.40			0.62			1.2			
Enclosure				Totally enclosed self-cooled (IP44)								
Ambient environment specification				Use temperature: 0 to 40°C/Storage temperature: -20 to +60°C, Use humidity and storage humidity: 20 to 80%RH (non-condensing) Free from dust, dirt, metallic powder , corrosive gas, flammable gas, oil mist, and others. Avoid outdoor use or direct sunlight. Altitude: 1,000 m or less. Motor insulation: 100 MΩ (500 VDC) or higher. Dielectric strength: 1500 VAC1 min. Insulation class: B								
Mounting direction				Can be installed in any direction								
Servo drive	Incremental (14 wire)	24VDC	DCJ-055-09, DDP-090-09, DEP-090-09				DCJ-055-18, DDP-090-18, DEP-090-18				DDP-090-36, DEP-090-36	
			HA-680-4-24						HA-680-6-24			
	Incremental (4 wire serial)	24VDC	DEP-090-09				DEP-090-18				DEP-090-36	
			ABS	24VDC	DEP-090-09				DEP-090-18			
	Dual ABS	24VDC	BEL or BPL-090-06				BEL or BPL-090-14				BEL or BPL-090-30	

Note 1: The table shows typical output values of actuators.

Note 2: Values when using HA-680 drive.

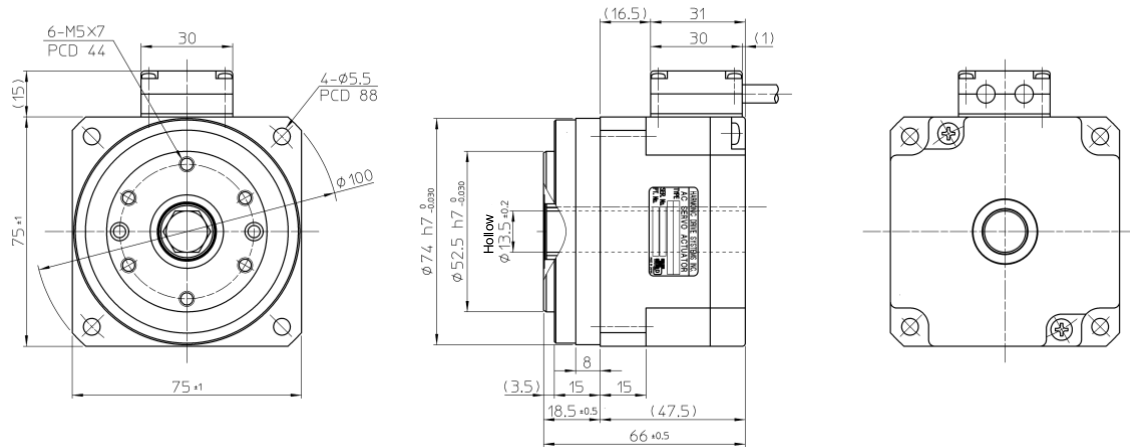
Note 3: All values are typical.

Note 4: Incremental encoder resolutions are obtained by [motor encoder resolution] x 4 x [reduction ratio]. Motor ABS encoder resolutions are obtained by [motor encoder resolution] x [reduction ratio].

Note 5: Refer to the HA-680 manual for details.

Note 6: All parts, except the rotary sliding parts (oil seal) and C option connectors, are protected against solid bodies of superior dimensions to 1mm, and against the water sprays. Panel mount connectors are IP68.

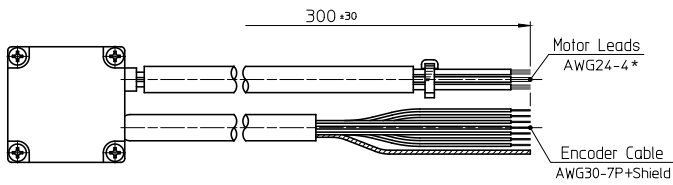
■FHA-14C-xxx-US200-□ □ K



■Cables - Common

(except size 14, 24VDC version

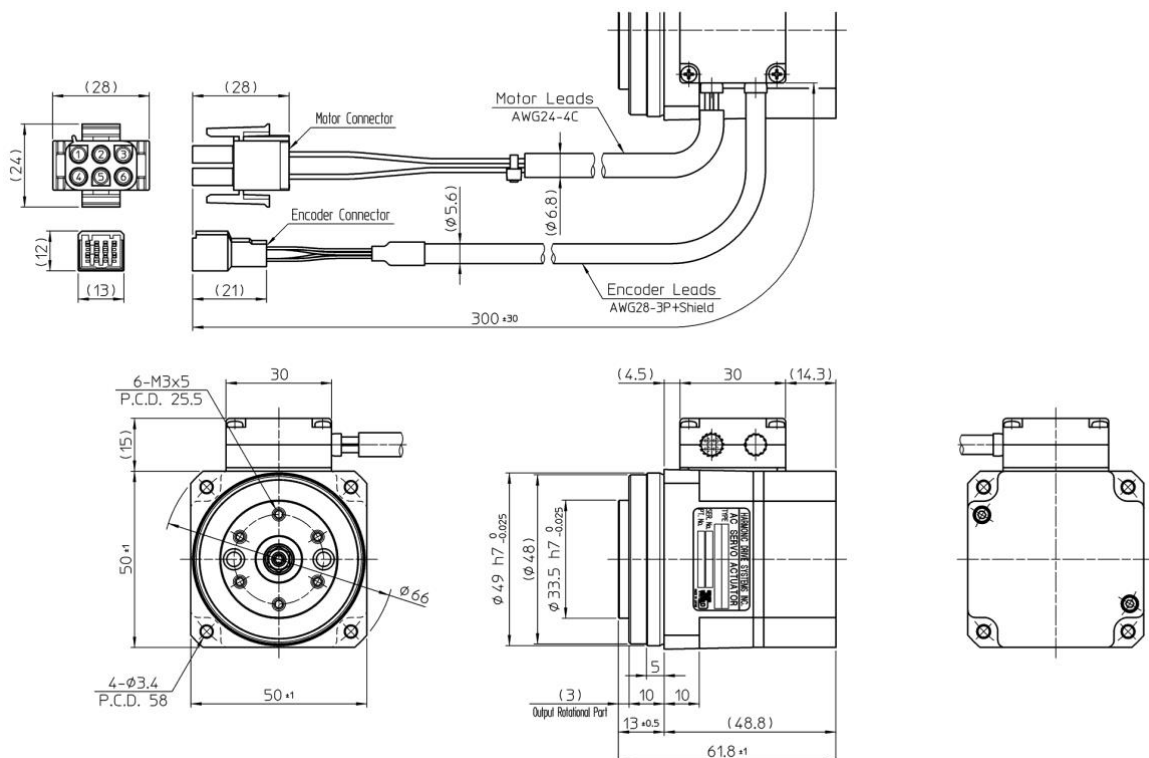
*AWG20-4 for FHA-14C-xxx-US200-E)



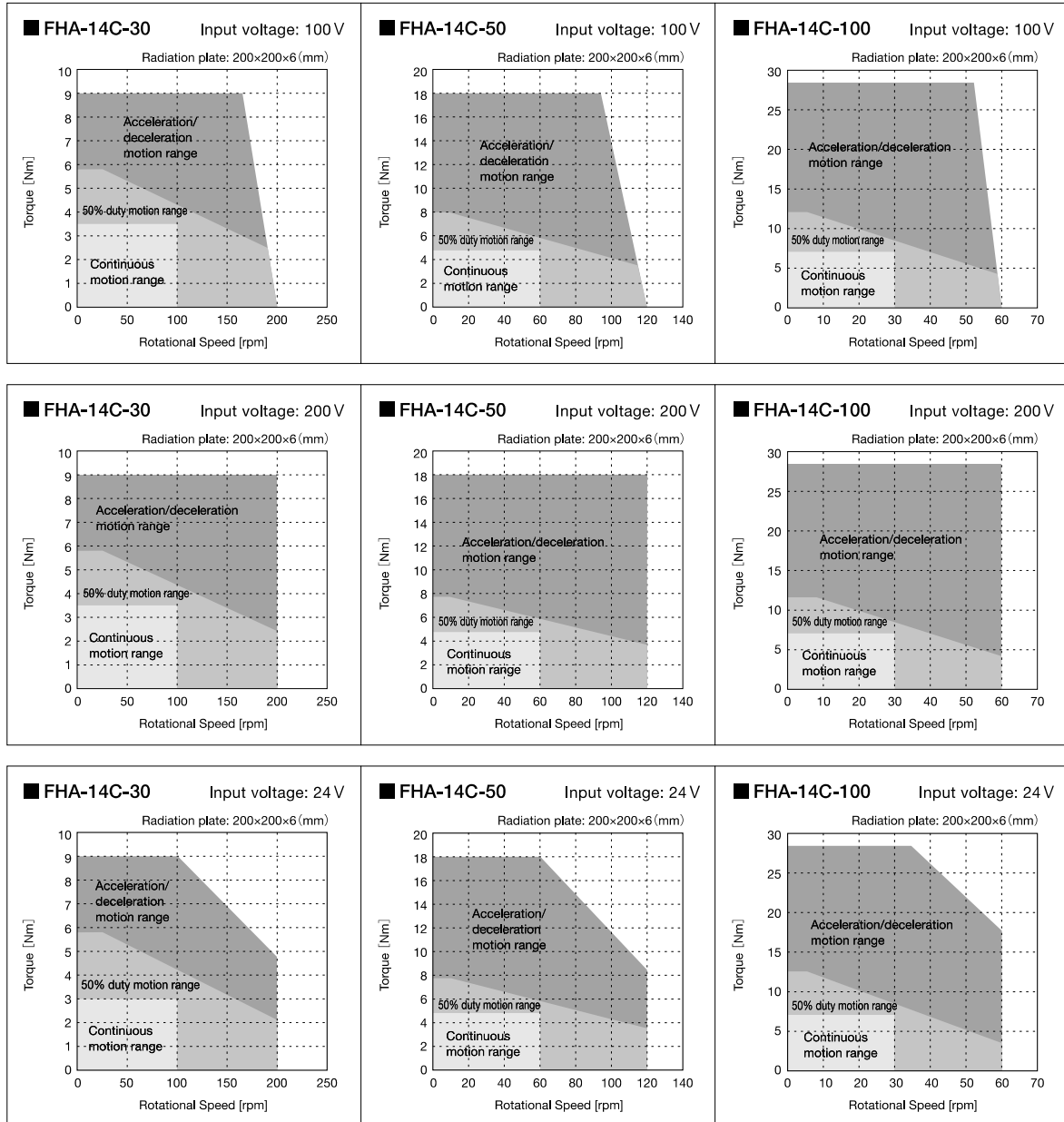
1-5-3 Actuators with Absolute Encoder

■FHA-8C-xxx-12S17b □-C

Unit: mm (third angle projection)



FHA-14C





Specifications												
Precision grade		JIS B 1704: 1st grade 1 *		A _{h7}		Bore						
Gear teeth		Gleason		B		Hub dia.						
Pressure angle		20°		C		Pitch dia.						
Helix angle		35°		E		Mounting distance						
Material		SCM415		F		Total length						
Heat treatment		Carburizing		G		Crown to back						
Tooth hardness		55 ~ 60HRC		H		Hub width						
				I		Length of bore						
				J		Face width						
				K		Holding surface dia.						

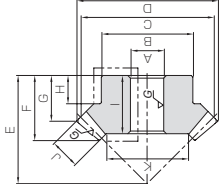
* The precision grade of J Series products is equivalent to the value shown in the table.

Catalog No.	Gear ratio	No. of teeth	Shape	Avr	B	C	D	E	F	G	H	I	J	K	Allowable torque (N·m)	Bending strength	Surface durability	Allowable torque (kg·m)	Backlash (mm)	Weight (kg)
MMMSG2-20R MMMSG2-20L MMMSG2.5-20R MMMSG2.5-20L	1	20	B3	12	35	40	42.7	35	21.98	16.35	12.5	20	9	24.54	17.0	23.5	1.73	2.40	0.04~0.10	0.14
				14	42	50	53.2	45	28.63	21.6	16	26	11	30.89	32.7	46.1	3.33	4.70	0.05~0.11	0.27
				16	52	60	63.99	50	30.78	21.99	16	27	14	34.4	58.5	83.7	5.97	8.54	0.06~0.12	0.43
				20	50	70	74.53	55	32.45	22.26	14	29	16	42.75	91.8	133	9.36	13.6	0.07~0.13	0.51
MMMSG3-20R MMMSG3-20L MMMSG3.5-20R MMMSG3.5-20L	1	25	B3	20	55	80	84.99	65	39.13	27.5	17	35	18	49.08	136	199	13.8	20.3	0.09~0.15	0.80
				12	38	50	52.5	40	23.43	16.25	11	21	11	30.89	27.5	47.0	2.80	4.79	0.04~0.10	0.21
				16	45	62.5	65.54	50	29.57	20.27	14	26	14	37.4	54.3	94.5	5.54	9.64	0.05~0.11	0.37
				20	55	75	78.78	60	35.6	24.39	17	31	17	43.92	94.5	167	9.64	17.0	0.06~0.12	0.65
MMMSG3-25R MMMSG3-25L MMMSG3.5-25R MMMSG3.5-25L	1	25	B4	25	65	87.5	91.81	70	41.65	28.41	19	37	20	52.43	151	270	15.4	27.5	0.07~0.13	1.04
				28	75	100	104.7	80	47.8	32.35	22	42	23	58.95	216	392	22.1	40.0	0.09~0.15	1.57
				14	45	60	62.42	50	29.27	21.21	15	26	12	38.06	38.5	78.6	3.93	8.02	0.04~0.10	0.36
				16	55	75	78.04	60	34.08	24.02	16	30	15	47.57	75.3	156	7.68	16.0	0.05~0.11	0.66
MMMSG2-30R MMMSG2-30L MMMSG2.5-30R MMMSG2.5-30L	1	30	B4	20	65	90	93.61	70	40.25	26.8	18	36	20	55.43	139	294	14.2	30.0	0.06~0.12	1.11
				25	80	105	109.21	80	44.4	29.6	20	40	22	67.77	204	436	20.8	44.5	0.07~0.13	1.75
				28	90	120	124.7	90	49.27	32.35	22	44	25	77.29	303	657	30.9	67.0	0.09~0.15	2.49

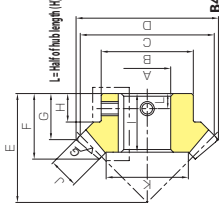
[Caution on Product Characteristics]

- ① A set of miter gears must be identical in module and number of teeth, but opposite in spiral hands.
- ② The allowable torques shown in the table are the calculated values according to the assumed usage conditions. Please see page 272 for more details.
- ③ Dimensions of the outside diameter, the overall length and crown to back length are all theoretical values, and some differences will occur due to the corner chamfering of the gear tips.
- ④ These gears produce axial thrust forces. See page 274 for more details.
- ⑤ Please read "Caution on Performing Secondary Operations" (Page 274) when performing modification and/or secondary operations for safety concerns. KHK Quick-Mod Gears, the KHK's system for quick modification of KHK stock gears is also available.
- ⑥ In the illustration, the area surrounded with - - - - - line is masked during the carburization process and can be modified. However, care should be exercised since the hardness is high (approx. HRC40, maximum).

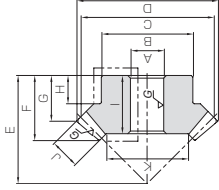
[Caution on Secondary Operations]



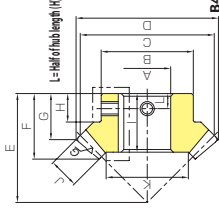
B3



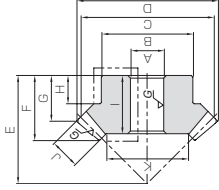
B4



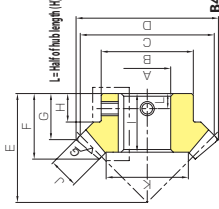
B3K



B4K



B3K



B4K

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Bore h7		12	14	15	16	17	18	19	20	22	25	28	30	32	35	40	45	50
Keyway as9		4 x 1.8		5 x 2.3		6 x 2.8		M5		M6		M8		10 x 33		12 x 33		14 x 38
Screw size		M4																M10
Catalog No.																		
MMMSG2-20R	1 BORE	B3K	B3K	B3K	B3K	B3K	B3K											
MMMSG2-20L	1 BORE	B3K	B3K	B3K	B3K	B3K	B3K											
MMMSG2.5-20R	1 BORE	B3K	B3K	B3K	B3K	B3K	B3K	B3K										
MMMSG2.5-20L	1 BORE	B3K	B3K	B3K	B3K	B3K	B3K	B3K										
MMMSG3-20R	1 BORE		B3K	B3K	B3K	B3K	B3K	B3K	B3K									
MMMSG3-20L	1 BORE		B3K	B3K	B3K	B3K	B3K	B3K	B3K									
MMMSG3.5-20R	1 BORE								B4K	B4K	B4K	B4K	B4K					
MMMSG3.5-20L	1 BORE								B4K	B4K	B4K	B4K	B4K					
MMMSG4-20R	1 BORE								B4K	B4K	B4K	B4K	B4K	B4K				
MMMSG4-20L	1 BORE								B4K	B4K	B4K	B4K	B4K	B4K	B4K			
MMMSG2-25R	1 BORE	B4K	B4K	B4K	B4K	B4K	B4K	B4K										
MMMSG2-25L	1 BORE	B4K	B4K	B4K	B4K	B4K	B4K	B4K										
MMMSG2.5-25R	1 BORE				B4K	B4K	B4K	B4K										
MMMSG2.5-25L	1 BORE				B4K	B4K	B4K	B4K										
MMMSG3-25R	1 BORE								B4K	B4K	B4K	B4K	B4K	B4K	B4K			
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MMMSG4-25L	1 BORE														B4K	B4K	B4K	
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MMMSG2-30L	1 BORE	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K						
MMMSG2.5-30R	1 BORE				B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	
MMMSG2.5-30L	1 BORE				B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K	
MMMSG3-30R	1 BORE								B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K		
MMMSG3-30L	1 BORE								B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K		
MMMSG3.5-30R	1 BORE										B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K
MMMSG3.5-30L	1 BORE										B4K	B4K	B4K	B4K	B4K	B4K	B4K	B4K
MMMSG4-30R	1 BORE																	B4K
MMMSG4-30L	1 BORE																	B4K

[Caution on J series]

- ① As available-on-request products, requires a lead-time for shipping within **2 working days (excludes the day ordered)**, after placing an order. Please allow additional shipping time to get to your local distributor.
- ② Number of products we can process for one order is **1 to 20 units**. For quantities of 21 or more pieces, we need to quote price and lead time.
- ③ Keyways are made according to JIS B1301 standards. J5.9 tolerance.
- ④ Certain products which would otherwise have a very long tapped hole are counterbored to reduce the length of the tap. (Products marked with "*" are tap size).
- ⑤ For products having a tapped hole, a set screw is included.

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Rue Archimède, 1 bte L6.11.01, 1348 Louvain-la-Neuve, Belgique | www.uclouvain.be/epl