

École polytechnique de Louvain

Design of an energy-efficient active knee prosthesis through comparison of optimized kinematic chains including elastic actuator and variable transmission concepts

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Abstract

Background: Individuals with transfemoral amputation face diverse mobility deficiencies. Knee prostheses stand out as rehabilitation devices capable of improving the quality of life for amputees. While passive designs remain the most commonly used, a growing number of active devices are emerging. Active prostheses hold potential to restore users' freedom of movement through their versatility in common lower limb tasks. However, their development must address critical challenges, such as ensuring sufficient autonomy and remaining lightweight, despite the embedded motor and battery.

Objectives: The aim is to design a lightweight active knee prosthesis with high energetic efficiency by comparing different kinematic chains from motor to knee joint in terms of motor energy consumption and peak power.

Simulation: The kinematic chains are grouped into two families of solutions: elastic actuators and variable transmissions. Optimizations are conducted on the simulation models of these different kinematic chains for walking exclusively and for walking and stairs ascent combined. Three small electrical motors are considered as candidate to actuate the prosthesis, with particular attention paid to their mechanical and thermal limitations.

Results: The double parallel elastic actuator (2PEA) and the clutchable series elastic actuator (CSEA) stand out in the first solutions category. The 2PEA excels in terms of walking energy consumption but is less efficient for stairs ascent. The CSEA provides inferior but more consistent results for various lower limb tasks. Similarly, two variable transmissions stand out from the optimizations: the 2-stage planetary gear (2PG) and the Utah bionic leg mechanism, both allowing the use of the smaller candidate motor to drive the kinematic chain. As the beneficial effects of the two approaches are distinct, combining them proves advantageous. After evaluating the 4 possible combinations, the CSEA-2PG combination is chosen. A draft mechanism for implementing this combination is proposed.

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Nomenclature

Symbols

α	Constant angle in the mechanism of Utah Bionic Leg	[rad]
β_{disen}	Clutch disengagement moment as a percentage	[%]
β_{en}	Clutch engagement moment as a percentage	[%]
β	Angle of the R4 joint in the mechanism of Utah Bionic Leg	[rad]
Δl	Linear spring deformation	[m]
Δr	Difference between the outer and inner radius	[m]
$\Delta T_{s,\infty}$	T° difference between housing and ambient at thermal equilibrium	[K]
$\Delta T_{w,\infty}$	T° difference between winding and ambient at thermal equilibrium	[K]
$\Delta T_{w,max}$	Difference between the winding steady-state and starting T°	[K]
$\Delta \theta_{k_p}$	Parallel spring deformation	[rad]
$\Delta \theta_{k_s}$	Series spring deformation	[rad]
$\Delta \theta_{k_{p2}}$	Parallel spring 2 deformation (2PEA)	[rad]
δ_0	Free length of the spring in the mechanism of Utah Bionic Leg	[m]
δ_1	Point A stroke in the mechanism of Utah Bionic Leg	[m]
δ_2	Point B stroke in the mechanism of Utah Bionic Leg	[m]
δ_3	Length of the spring in the mechanism of Utah Bionic Leg	[m]
$\delta_{2,h}$	Upper limit of δ_2 translation	[m]
$\delta_{2,l}$	Lower limit of δ_2 translation	[m]
η_e	Motor electrical efficiency	[-]
η_g	Gearbox efficiency	[-]
η_{bs}	Ball screw efficiency	[-]
η_{mec}	Transmission mechanical efficiency	[-]
η_{me}	Motor maximal efficiency	[-]

λ_{init}	Initial vertical coordinate of the ball screw nut (at full extension)	[m]
λ	Vertical coordinate of the ball screw nut	[m]
ω_m	Motor speed	[rad/s]
ω_{k_p}	Parallel spring speed	[rad/s]
ω_{k_s}	Series spring speed	[Nm]
$\omega_{k_{p2}}$	Parallel spring 2 speed (2PEA)	[rad/s]
$\omega_{m,max}$	Motor maximum allowed speed	[rad/s]
ω	Angle of the R2 joint in the mechanism of Utah Bionic Leg	[rad]
τ_w	Thermal time constant winding	[s]
τ_{torque}	Torque ratio	[-]
θ_1	Angle between the crank and positive x-axis	[rad]
θ_j	Knee joint position	[rad]
θ_m	Motor position	[rad]
θ_{0,k_p}	Parallel spring free-length	[rad]
$\theta_{0,k_{p2}}$	Parallel spring 2 free-length (2PEA)	[rad]
$\theta_{1,init}$	Initial angle between the crank and positive x-axis (full extension)	[rad]
θ_{CSEA}	CSEA output position	[rad]
$\theta_{m,bp}$	Fixed motor position when the clutch engages (CSEA)	[rad]
a_m	Motor acceleration	[rad/s ²]
A_x	Abscissa of point A	[m]
A_y	Ordinate of point A	[m]
A	Position of R2 and P1 joints in the mechanism of Utah Bionic Leg	[(m;m)]
a	Horizontal coordinate of the ball screw nut	[m]
B_x	Abscissa of point B	[m]
B_y	Ordinate of point B	[m]
B	Position of R3 and P2 joints in the mechanism of Utah Bionic Leg	[(m;m)]
b	Length of connecting rod in the mechanism of Utah Bionic Leg	[m]
C_w	Score for the estimated weight criterion	[-]
C_{E_e}	Score for the walking electrical energy consumption criterion	[-]

C_{imp}	Score for the implementation complexity criterion	[-]
C_{th}	Score for the thermal behavior during stairs ascent criterion	[-]
C_{tot}	Total weighted cost function	[-]
d	Length of the 'd' bar in the mechanism of Utah Bionic Leg	[m]
E_e	Motor electrical energy consumption over one cycle	[J]
e	Length of the 'e' bar in the mechanism of Utah Bionic Leg	[m]
F_1	1 st objective function (motor peak electrical power)	[W]
F_2	2 nd objective function (motor electrical energy consumption)	[J]
F_3	3 rd objective function (scalar kinematic agreement for CSEA)	[rad ² s]
F_b	Action force of the connecting rod in the mechanism of Utah Bionic Leg	[N]
F_c	Linear spring compressive force	[N]
F_a	Parallel force to the ballscrew in the mechanism of Utah Bionic Leg	[N]
F_{clamp}	Axial clamping force of the 2PG braking system	[N]
F_{ES}	Parallel force to the P2 joint in the mechanism of Utah Bionic Leg	[N]
F_{net}	Net force acting on the CVT ring	[N]
F_{P1}	Perpendicular force to the P1 joint in the mechanism of Utah Bionic Leg	[N]
F_{P2}	Perpendicular force to the P2 joint in the mechanism of Utah Bionic Leg	[N]
F_{s0}	Spring preload force in the mechanism of Utah Bionic Leg	[N]
F_s	Spring force in the mechanism of Utah Bionic Leg	[N]
f	Dynamic friction coefficient	[-]
I_m	Motor's rotor moment of inertia	[kg m ²]
i_m	Motor current	[A]
i_N	Motor nominal current	[A]
I_{chain}	Kinematic chain moment of inertia (excluding motor)	[kg m ²]
$i_{m,RMS}$	Root-mean square value of the motor current	[A]
I	Total moment of inertia	[kg m ²]
K_m	Motor torque constant	[Nm/A]
k_p	Parallel spring stiffness	[Nm/rad]
k_s	Series spring stiffness	[Nm/rad]

k_{p2}	Parallel spring 2 stiffness (2PEA)	[Nm/rad]
k	Linear spring stiffness	[N/m]
l_0	Linear spring free-length	[m]
l_{min}	Linear spring length at minimal position of the CVT ring	[m]
L	Connecting rod length	[m]
M_m	Motor mass	[kg]
N_g	Gearhead reduction ratio	[-]
$N_{2pg,h}$	2PG high reduction ratio	[-]
$N_{2pg,l}$	2PG low reduction ratio	[-]
N_{2pg}	2PG reduction ratio	[-]
N_{bs}	Ball screw reduction ratio	[-]
N_{crc}	CRC reduction ratio	[-]
N_{cvt}	CVT reduction ratio	[-]
N_{red}	Global reduction ratio	[-]
N	Number of friction interfaces	[-]
P_e	Motor electrical power	[W]
P_h	Ball screw lead	[m]
P_J	Motor Joule's power losses	[W]
P_j	Knee joint power	[W]
P_N	Motor rated power	[W]
$P_{e,pos}$	Positive part of the motor electrical power	[W]
P_{kp}	Parallel spring power	[W]
P_{ks}	Series spring mechanical power	[W]
P_{kp2}	Parallel spring 2 power (2PEA)	[W]
P_{mech}	Motor mechanical power	[W]
R_b	Reaction force of F_b	[N]
r_i	Inner radius	[m]
R_m	Motor winding resistance	[Ω]
r_o	Outer radius	[m]

$\mathbf{R}_a$	Reaction force of F_a	[N]
$\mathbf{R}_{ES}$	Reaction force of F_{ES}	[N]
$\mathbf{R}_{P1}$	Reaction force of F_{P1}	[N]
$\mathbf{R}_{P2}$	Reaction force of F_{P2}	[N]
$\mathbf{R}_s$	Reaction force of F_s	[N]
$\mathbf{R}_{th1}$	Thermal resistance winding-housing	[K/W]
$\mathbf{R}_{th2}$	Thermal resistance housing-ambient	[K/W]
r	crank length	[m]
$\mathbf{S}_x$	Abscissa of point S	[m]
$\mathbf{S}_y$	Ordinate of point S	[m]
$\mathbf{S}$	Position of R4 and P3 joints in the mechanism of Utah Bionic Leg	[(m;m)]
s	Ball screw nut vertical displacement	[m]
T_a	Ambient temperature	[K]
T_j	Knee joint torque	[Nm]
T_l	Load torque	[Nm]
T_m	Motor torque	[Nm]
T_N	Motor nominal torque	[Nm]
T_w	Winding temperature	[K]
$t_{\beta_{disen}}$	Clutch disengagement instant within the cycle duration	[s]
$t_{\beta_{en}}$	Clutch engagement instant within the cycle duration	[s]
T_{k_p}	Parallel spring torque	[Nm]
T_{k_s}	Series spring torque	[Nm]
$T_{k_{p2}}$	Parallel spring 2 torque (2PEA)	[Nm]
T_{lock}	Torque capacity of the 2PG braking system	[Nm]
$T_{m,p}$	Motor stall torque	[Nm]
$T_{w,0}$	Winding starting temperature	[K]
$T_{w,\infty}$	Winding steady-state temperature	[K]
$T_{w,max}$	Winding maximum allowed temperature	[K]
u_m	Motor voltage	[V]

w_w	Weight for the estimated weight criterion	[-]
w_{E_e}	Weight for the walking electrical energy consumption criterion	[-]
w_{imp}	Weight for the implementation complexity criterion	[-]
w_{th}	Weight for the thermal behavior during stairs ascent criterion	[-]
x_{max}	Maximal axial position of the CVT ring	[m]
x_{min}	Minimal axial position of the CVT ring	[m]
x	Axial position of the CVT ring	[m]
Z_r	Number of teeth of the common ring gear	[-]
Z_{s_1}	Number of teeth of the first sun gear	[-]
Z_{s_2}	Number of teeth of the second sun gear	[-]

Abbreviations

2PEA	Double Parallel Elastic Actuator
2PG	2-stage Planetary Gear
CPEA	Clutchable Parallel Elastic Actuator
CRC	Connecting Rod-Crank
CSEA	Clutchable Series Elastic Actuator
CVT	Continuously Variable Transmission
DC	Direct Current
DirD	Direct Drive
EC	Electronically Commutated
ESF	Early-Stance Flexion
GRF	Ground Reaction Force
PEA	Parallel Elastic Actuator
PG	Planetary Gear
PSEA	Parallel and Series Elastic Actuator
RMS	Root-Mean Square
SA	Stairs Ascent
SEA	Series Elastic Actuator
vGRF	Vertical component of the Ground Reaction Force
W	Walking

Chapter 1

Introduction

There are 9.36 million transfemoral amputees in the world. The causes of amputation are varied: diabetes, cancer, congenital malformations, infections, or accidents. This has many consequences for amputees' lives. The simplest, most instinctive everyday tasks suddenly become real challenges, requiring adaptations. Mobility is greatly reduced and getting around becomes very difficult[12].

To remedy this, numerous prostheses have been developed and marketed in recent decades. They enable patients to return to an almost normal life by serving as a substitute for the lost limb and restoring the body's appearance. These prostheses try to replicate as closely as possible the kinematics of the lost knee. They have evolved with the advent of microprocessors and the development of electric motors. These two advances have made prostheses more versatile, with real-time control of the internal mechanism thanks to the insertion of motors and sensors, offering patients even more comprehensive support[12].

Today, there are various models of mechanical and microprocessor-controlled prostheses on the market, but there is only one active motorized prosthesis commercially available, the Power Knee from the company Ossur [44]. All the others are still prototypes in development.

Active knee prostheses offer a real opportunity to recover even more of the human functions lost with amputation. In fact, as well as covering so-called passive tasks, i.e. tasks that can be carried out by mechanical prostheses, such as walking or descending stairs, they also carry out so-called active tasks, requiring the assistance of a motor, such as climbing stairs. But beyond the cost, the absence of other active models is explained by the design constraints that these prostheses imply. They are heavier than passive prostheses and those controlled by microprocessors, because of the size of the on-board motor and battery needed to cope with the higher energy consumption and power peaks associated with active tasks. The latter implies limited autonomy, which is a constraint in users' daily lives [30].

A reduction in energy consumption and better management of the power peaks of these prostheses would extend their autonomy. This would also make it possible to reduce the size of the on-board motor and battery to make the prosthesis lighter. Our research question for this end-of-study project is based on this observation. There is a need to improve the energy efficiency and peak power management of active prostheses in order to improve the daily experience of its users.

To achieve this reduction objective, we looked at the kinematic transmission chain from the motor to the prosthetic knee joint with the aim of determining the best configuration capable of ensuring walking and stair climbing while reducing energy consumption and motor power peaks. Different configurations were designed, mathematically modelled and optimised to converge on the best one. We propose a mechanism concept for implementing the chosen kinematic configuration.

Our work is part of global research into prostheses for transfemoral amputees. It provides a basis for comparison between different configurations that, to our knowledge, has not yet been done in the literature. Our investigations aim to provide knowledge to guide the design of new active prosthesis mechanisms with improved energy efficiency for the benefit of all.

Our dissertation is also the fruit of a long period of work carried out throughout the 2023-2024 academic year in pairs. It marks the end of our polytechnic engineering course. It was also an opportunity for us to combine the approaches specific to our different disciplines, mechanical and electromechanical and to find out more about the application of robotics to the medical field of rehabilitation.

The document is structured as follows:

The [chapter 2](#) presents the state of the art. It gives an overview of the range of prostheses on the market and the prototypes developed through studies, focusing more on active prostheses.

The aim of the [chapter 3](#) is to gain a better understanding of the natural behaviors that prostheses attempt to replicate. The biomechanics of the knee joint is presented for the five most common locomotion tasks of walking, stairs ascent, stairs descent, stand-to-sit, and sit-to-stand. The angle, angular velocity, torque, power and ground reaction force curves for each of these tasks have been compiled by cross-referencing several different studies. The chapter concludes by looking at the performance required of the motor to perform the various tasks, based on the curves of the natural quantities.

In [chapter 4](#) a set of specifications covering the main functions and constraints associated with the kinematic chain included in a knee prosthesis is drawn up.

Based on the project specifications, two main avenues are explored in the following chapters: elastic actuators and variable transmissions.

The [chapter 5](#) focuses on the first family of solutions, that of elastic actuators. Different configurations are tested and optimized with the twin objectives of reducing energy consumption and reducing the amplitude of power peaks for walking exclusively, and walking combined with stairs ascent. Each optimization is carried out for three different motors. Their thermal limitations are also studied. The conclusion discusses the results obtained and open perspectives for the upcoming global comparison.

The [chapter 6](#) looks at the second family of solutions: variable transmissions. The opti-

mization procedure is identical to that in the previous chapter. Various mechanisms with a variable reduction ratio, sometimes taken from the literature, are tested and optimized to meet the two objectives.

chapter 7 gives a global comparison of the two approaches. It develops a final solution that combines a configuration from each family to take advantage of the best features of both, aiming to improve energy efficiency and power even further.

A mechanism concept implementing the final configuration is presented in chapter 8. The concept is evaluated according to the main functions and constraints of the specifications.

Finally, the overall conclusion of all the work carried out can be found at chapter 9. The various stages of this dissertation and the main results are summarised. Future prospects are presented, with the aim of continuing the work begun in our dissertation in order to converge on the development of a new mechanism that would enable greater energy efficiency.

The various appendices can be found at the end. They include the graphs not presented in the document for the configurations that performed less well, as well as the mathematical developments of the concepts presented. All the symbols used are included in the nomenclature table.

To write this document, we sometimes used ChatGPT, Deepl, and Grammarly to help us with syntax and spelling. All software proposals have been reviewed and sourced, and we take responsibility for the information contained in this manuscript.

Chapter 2

State of art

This chapter looks at the range of prostheses on the market or in development, with a particular focus on active prostheses.

2.1 Transfemoral amputees

2.1.1 Who ?

Transfemoral amputees are people who have had their lower limb removed above the knee. The femur is cut at different heights depending on the patient's case, so the stump is of different lengths. The reasons for amputation can be varied (accident, illness, congenital disease or injury). [48] After the operation and recovery, these people are fitted with a prosthesis to enable them to move around on a daily basis. Bionic prostheses are a field of research in their own right. The ultimate aim of these limb-replacement prostheses is to provide patients with behaviour that is as close as possible to that of their lost natural limb, so that they can continue to lead a comfortable life.

2.1.2 Patient classification

There are different levels of functional classification for transfemoral amputations. Patients are classified according to their independence in daily tasks. The table is taken as it stands from [30] :

K-level	Descriptor	Knee	Required functions
K0	This patient does not have the ability or potential to ambulate or transfer safely with or without assistance, and a prosthesis does not enhance his or her quality of life or mobility.	Not eligible for prosthesis	-
K1	This patient has the ability or potential to use a prosthesis for transfers or ambulation on level surfaces at fixed cadence—a typical limited or unlimited household ambulator.	Single-axis, constant-friction knee	Walking stability
K2	This patient has the ability or potential for ambulation with the ability to traverse lowlevel environmental barriers, such as curbs, stairs, or uneven surfaces—a typical community ambulator.	Single-axis, constant-friction knee	Walking stability and swing resistance
K3	This patient has the ability or potential for ambulation with variable cadence—a typical community ambulator with the ability to traverse most environmental barriers and may have vocational, therapeutic, or exercise activity that demands prosthetic use beyond simple locomotion.	Fluid and pneumatic control knees	Walking stability, swing resistance and early stance flexion
K4	This patient has the ability or potential for prosthetic ambulation that exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels—typical of the prosthetic demands of the child, active adult, or athlete.	Any are appropriate	Walking stability, swing resistance and early stance flexion

Swing Resistance : It controls the maximum flexion angle and determines the timing of full extension.

Early Stance Flexion : It allows a person to lower the center of mass of their body during stance, thus absorbing the striking impact force and ensuring the smooth transition from swing to stance.

2.2 Knee prosthesis

2.2.1 Global explanation

There is a wide range of prostheses on the market to meet the different needs of users, depending on their classification and also on their daily activities. Although there are different models, there are always three distinct parts: a socket, a prosthetic knee joint and a prosthetic foot.

The socket is made to measure for each patient. It connects the prosthesis to the stump. It consists of a lining for comfort and protection of the patient's stump and a suspension for attachment to the stump. The end has an adapter for connecting the prosthetic component. The knee joint provides stability and mobility for the leg. The foot prosthesis is designed to provide stable support contact. There are three main categories of prosthesis: active, passive and semi-active/quasi-passive [48]. These categories are explained in more detail below.

2.2.2 Knee prosthesis' companies

There are two main companies operating in the transfemoral prosthesis sector. These two are *Össur* et *Ottobock*.

Össur is an Icelandic company founded in 1971 by prosthetist Össur Kristinsson. It designs orthoses and prostheses for the lower and upper limbs and is present in America,

Europe, Asia and Africa. The company offers active and passive prostheses for patients of all classes, including young users and runners. The types of prosthesis are presented next.

Ottobock is the other leader. It is a German company founded in 1919 by Mr Otto Bock after the war to equip all war wounded. Ottobock offers semi-active prostheses and monocentric passive prostheses. The company has a worldwide presence.

There are some other companies providing passive and semi-active knee prosthesis : Endolite Blatchford (United Kingdom), Proteor (France), Rebocon Bionic (China and the Netherlands), Fillauer (USA) and College Park Industries (USA).

2.2.3 Passive prostheses

Passive prostheses, also known as mechanical prostheses, have no motor. The mechanisms making up the prosthesis move solely as a result of interaction with the outside world (weight loading, ground reaction force, bending angle, etc.). The advantage of these prostheses is their weight. They are much lighter than active prostheses. They are simpler and less expensive than other types. Passive prostheses mainly use hydraulic, pneumatic and friction elements.

Unfortunately, they are not very versatile. They can only be used for everyday tasks that do not absolutely require the production of energy from the knee joint, such as walking, descending stairs or ramps, or sitting down. Climbing stairs or ramps and standing up are therefore very difficult with this type of prosthesis. It is always the healthy leg that has to provide the impulsion.

In addition, the stiffness and damping of the mechanism are always constant or have a discrete number of values. They do not adapt to every passive task. They are mainly designed for walking, which is the most common task. [55]

Passive knee prosthesis on the market

Both Ossür and Ottobock offer passive prostheses. Ottobock only offers hydraulic monocentric models (Dynion Knee Joint) or friction-based models (Knee Joint with friction brake and lock).

Ossür offers pneumatic polycentric models such as the OP5 Knee, mechanical models such as the Balance Knee OFM1 and hydraulic models such as the Mauch Knee. There is also a children's model (Total Knee Junior).

The two brands also offer a sporty model, the Ottobock's Sport Knee Join the Ossür's Cheetah Knee



Figure 2.1: Ottobock's mechanic monocentric knee prosthesis : (A) Dynion K.J. with rotation hydraulics, (B) K.J. with friction brake and lock ; Ossur's knee prosthesis : (C) OP5 Knee, (D) Balance Knee OFM1, (E) Mauch Knee [47] [41].

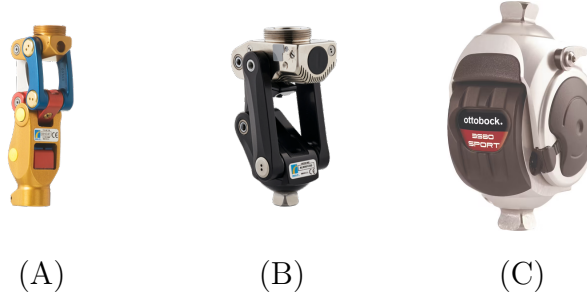


Figure 2.2: (A) Ossur's Total Knee Junior, (B) Ossur's Cheeta Knee, (C) Ottobock's Sport Knee Joint

2.2.4 Semi-active prostheses

To compensate for this lack of adaptability to the environment and to different tasks, semi-active prostheses have been developed. This type of prosthesis is also known as microprocessor-controlled knees (MPKs) or quasi-passive knees. They are equipped with microprocessors and sensors. Based on the data collected by the sensors, the microprocessor controls small actuators to modify the stiffness and/or damping of the mechanism. They offer greater adaptability than passive prostheses, but like passive prostheses, they cannot supply energy. They allow better adaptation to walking speed and reduce metabolic costs because they react better to the patient's movements. These prostheses are a little heavier than passive prostheses because they include additional components, in particular actuators and a battery [30].

Semi-active knee prosthesis on the market

Ottobock's C-Leg 4 prosthesis is the most widely used semi-active prosthesis in the world. Ottobock also offers the Genium prosthesis and its Genium X3 evolution. There is also the Kenovo prosthesis. [46] [43].

Ossur's Rheo Knee and Rheo Knee XC are suitable for K3 and K4 patients. The technology is based on the use of magnetic fluid to control flexion and extension using frictional resistance. It uses artificial intelligence to improve its response with repeated use. [42].

Other well-known companies include Proteor's Freedom Plié 3 and Rebocon Bionic's Intuy Knee.



Figure 2.3: Ottobock's MPKs [47] : (A) C-Leg 4, (B) Genium X3, (C) Ossur's MKP Rheo Knee [41], (D) Proteor MKP Plié 3 [60] (E) Rebocon Bionic MKP Intuy Knee [40]

2.2.5 Active prostheses

In an attempt to get even closer to the knee's natural behaviour, active knee prostheses (APKs) have been developed. They are also known as the powered knee prostheses. The joint inside the mechanism is activated by a motor controlled by a microprocessor based on information from on-board sensors. APKs can carry out active tasks thanks to their motor, but require a larger battery to be on board to ensure operation, as they consume more energy. This makes the final product considerably heavier, and autonomy is the main limitation. To improve all this, progress is needed on motors (reducing their size and increasing their power), on batteries (reducing their size and increasing their storage capacity) and also on mechanisms.

Another shortcoming is the absence on some models of a passive mode if the battery is empty. The prosthesis is then no longer useful.

Active prostheses are more expensive and more sensitive to their environment. Motors and batteries can overheat if they are used too much and without a good cooling system.

There is another type of knee prosthesis : there are also knee prostheses coupled with ankle prostheses. The advantage of this pairing lies in the opposite energy balances of these two joints in the human body. Walking is a mainly dissipative task for the knee but an active one for the ankle.

When walking, the knee brakes during the swing phase to prevent over-extension, absorbs the shock when the heel strikes and brakes to prevent flexion during the stance phase. It constantly absorbs and dissipates energy. The ankle provides an impulse to enable the transition from the stance phase to the swing phase by lifting the toes off the ground. The energy is stored in the knee and transmitted to the ankle for the push-off [30].



Actives knee prosthesis on the market

The only active prosthesis on the market is offered by Ossur: the Power Knee. It uses an Serie Elastic Actuator and is equipped with a high power motor to provide active energy. Few people are fitted with it because it is more expensive than passive prostheses and is also heavier [44], [10].

Figure 2.4: Ossur Power Knee [44]

Utah Bionic Leg

This complete leg prosthesis was designed by the Department of the mechanical Engineering and Robotics Center of the University of Utah in the USA. It is a lightweight robotic leg prosthesis replicating the biomechanics of the knee, ankle and toe joint in the sagittal plane. There are two motorised modules: one for the knee and the other for the ankle and toes. Although it is an active prosthesis, its weight, size and battery life match those of semi-active prostheses on the market such as the Ottobock C-Leg. Clinical tests have shown that the prosthesis approximates the natural kinetics and kinematics of the leg [62], [61].



Figure 2.5: Utah Bionic Leg. Picture from [62]

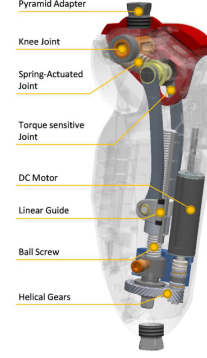


Figure 2.6: Knee mechanism of the Utah Bionic Leg. Picture from [62]

The CSEA Knee

The knee joint is based on a torque-sensitive mechanism that uses two approaches to reduce motor demand: variable transmission and elastic actuators. This mechanism enables the torque ratio to be adapted passively and continuously. So during the support phase, the torque ratio is high because the needed torque is high and during the swing phase the torque ratio is low because the natural knee torque is low. The on-board sensors and control system enable the prosthesis to be adapted at any time to suit the task in hand (walking, climbing stairs, etc.). The whole prosthesis can also operate passively if the battery runs out, which is not the case with other prostheses. In addition, this prosthesis takes advantage of the dissipative energy aspect of the knee joint while walking to regenerate energy in the battery via the joint motor, which acts as a generator at certain times in the cycle. This increases autonomy [55].

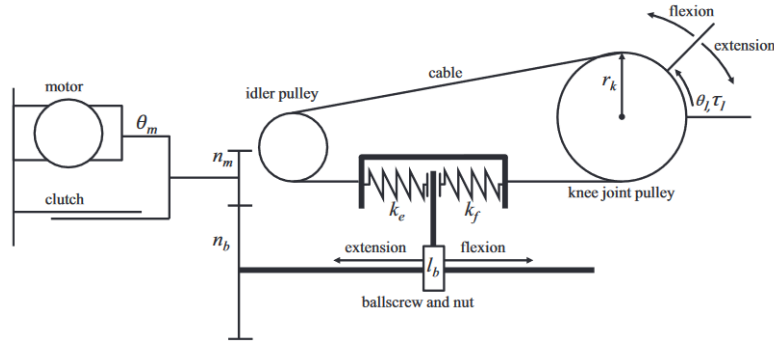


Figure 2.7: From [55] : The geometric configuration of the clutchable series-elastic actuator knee is depicted. The parallel motor and clutch were connected to a drive pulley with n_m teeth that drive the ballscrew pulley with n_b teeth. A ballscrew, with lead l_b , drives a ball nut between the extension and flexion compression springs with respective stiffnesses of k_e and k_f . The series-elastic housing drives the knee joint pulley, with radius r_k , via a drive cable.

The CyberLeg Beta-Prosthesis

This is a complete active transfemoral prosthesis consisting of a joint for the ankle and another for the knee. It can function passively for walking, but can also provide energy for power-consuming tasks [15].

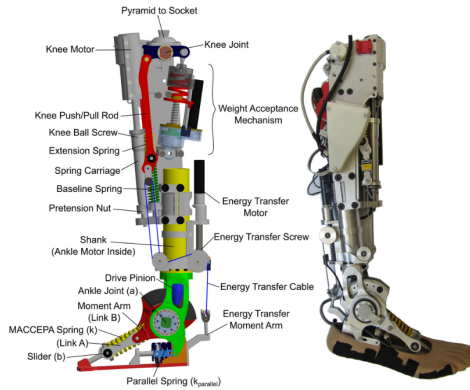


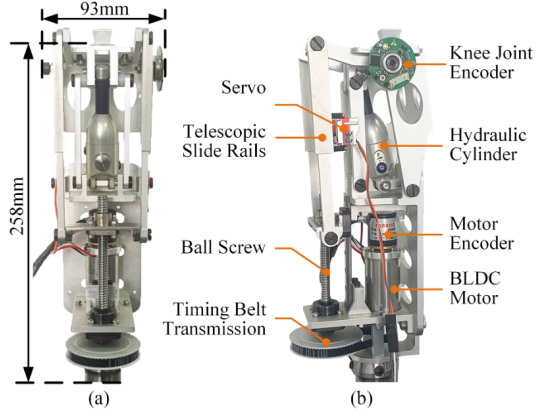
Figure 2.8: CyberLeg Beta Prosthesis [15]

It consists of 4 main mechanisms that interact with each other. The first one is a powered ankle. The second is the Weight Acceptance system, which takes advantage of the elasticity of the support phase to reduce energy consumption. The third one is the Knee Drive Baseline Actuator (KD). This is the system that provides the positive energy needed to perform the tasks such as stairs ascent or sit-to-stand task. The fourth system is the Energy Transfer system. It transfers some of the energy dissipated by the knee during the walking cycle to the ankle to reduce ankle torque.

Tested under the ideal conditions of its design on the test bench, the prosthesis works very well, but in reality, users seem to need time to adapt to it. They are unable to directly adopt the ideal behaviour that would enable the prosthesis to function optimally and save energy as intended. Moreover the Weight Acceptance and the Energy Transfer systems still need to be improved to really save energy. The mechanism is too sensitive to deviations from reality. Amputee patients have a different biomechanical behaviour to that of a non-amputee [15].

Polycentric Hybrid Knee Prosthesis

It is a novel polycentric hybrid knee prosthesis (PHKP). It is called hybrid because it can switch between passive mode for tasks as ground-walking and stairs descent to active modes for tasks as stairs ascent and sit-to-stand. Passive mode is provided by a 4-bar mechanism and active mode is provided by a 6-bar mechanism. The two modes are decoupled. The advantages of a polycentric prosthesis compare to a monocentric one is a better geometric stability during the stance phase and an increase in toe clearance during the swing phase thanks to an optimal designed instantaneous centre of rotation trajectory. In active mode, an impedance control regulates the joint torque based on the knee motion. They tried to make the prosthesis as compact and lightweight as possible. The mode transition is fast and is controlled by electromagnets. It allows to avoid mechanical interference between both modes. [65] [66]



(a) From [66] : (a) rear-view and (b) side-view picture of the PHKP prototype marking the main components are displayed, excluding the embedded system, motor driver, and battery

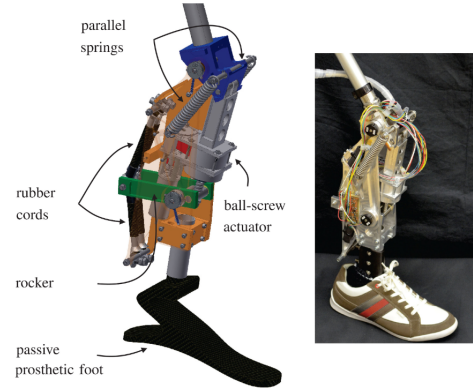


Fig. 4. CAD Drawing of the ANGELAA-leg (left) and picture of finished prototype (right).

(b) From [49]: CAD Drawing of the ANGELAA-leg (left) and picture of finished prototype (right)

Figure 2.9: Deux images côte à côte

Actuator With Angle-Dependent Elasticity (ANGELAA)

ANGELAA stands for ANGLE-dependent ELastic Actuator. This prototype uses parallel springs and viscoelastic rubber cords in series with the motor. The aim was to replicate the torque and stiffness of the knee joint. Previous studies enabled them to estimate the stiffness of the knee during walking. A Series Visco-Elastic Actuator actuate the mechanism. stiffness changes as a function of knee angle via a non-linear transmission. The position of the parallel springs has been optimised to replicate the moment-angle profile of a natural knee. Weight and power consumption have not been optimised. [49]

RIC Hybrid Knee Prosthesis

This prototype prosthesis is a hybrid that can operate in both active mode (for tasks such as stairs ascent or sit-to-stand) and passive mode (for tasks such as walking or stand-to-sit). The transition is made using a clutch. This reduces power consumption by deactivating the motor for passive tasks. Passive mode is based on the use of springs and dampers. It is lighter than the Ossur's Power Knee and the MIT CSEA. The model tested was made by recovering and modifying a passive knee prosthesis, the Ottobock 3R95. [28]

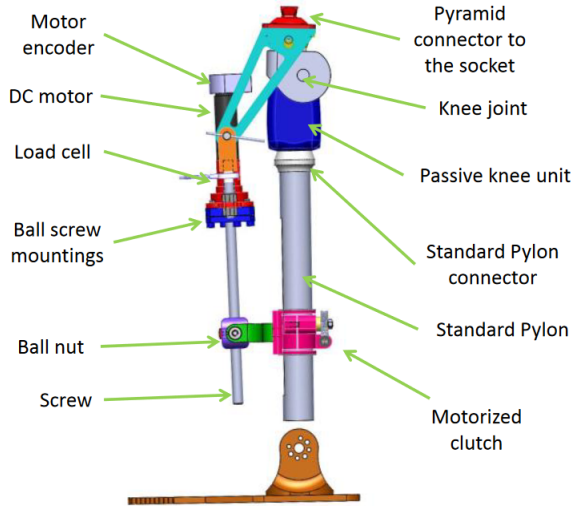


Figure 2.10: From [28] : CAD model of the RIC Hybrid Knee Prosthesis

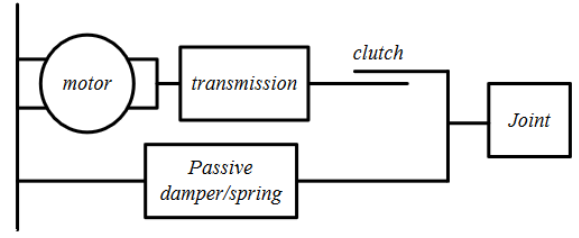


Figure 2.11: From [28] : Conceptual representation of the RIC Hybrid Knee Prosthesis

2.3 Biomechanical challenges

The use of a prosthesis brings with it certain biomechanical challenges.

The first concerns stability during the weight-bearing phase, i.e. when the patient leans on the prosthesis. The risk of falling is higher than for a healthy person (i.e. someone without a transfemoral amputee). According to [30], the probability of an amputee falling over the course of a year is 82% and the average number of falls per year per person is 3.9 (although these figures vary depending on the category of patient). The two main reasons are collapse during flexion in the stance phase and stumbling in the pre-stance phase.

A second problem is the absorption of the impact when the heel strikes the ground (start of the stance phase). In the case of a healthy leg, the Early Stance Flexion takes place just after the heel strike. This flexion absorbs the shock and maintains walking speed. In the case of amputees, this flexion does not always occur. Either because the prosthesis does not allow it or because the patient is not comfortable enough to use this function for fear of falling. As a result, the force is directed by the prosthesis towards the natural hip and pelvic joints. Cyclic shock can lead to long-term fatigue and damage to the stump. It can also lead to osteoarthritis in the healthy leg. This is due to the difference in movement between the leg with the prosthesis and the sound leg, with the former having no ESF and the latter having ESF. The absence of ESF is compensated for by the intact leg. It absorbs the impact twice per cycle and its stance phase is lengthened by 20 % to support the amputated leg in its task [30].

The third problem also stems from the asymmetry of the walking cycle (or any other task). Energy expenditure is higher because the healthy leg is put under greater strain and because the hip of the amputated limb works harder (higher torque) to stabilise the prosthesis during the stance phase. Notably due to the absence of ESF but some patients do not feel confident to use it [30].

Chapter 3

Common tasks of human lower-limb

This chapter presents the mechanical behaviour of the **knee** for the five most common human locomotion tasks: ground-level walking, stairs ascent and descent, stand-to-sit and sit-to-stand . Its aim is to provide a better understanding of the mechanical challenges of transfemoral prostheses.

3.1 Introduction

The study of human movements distinguishes three main planes characterizing the body: the sagittal plane, the frontal plane and the transverse plane:

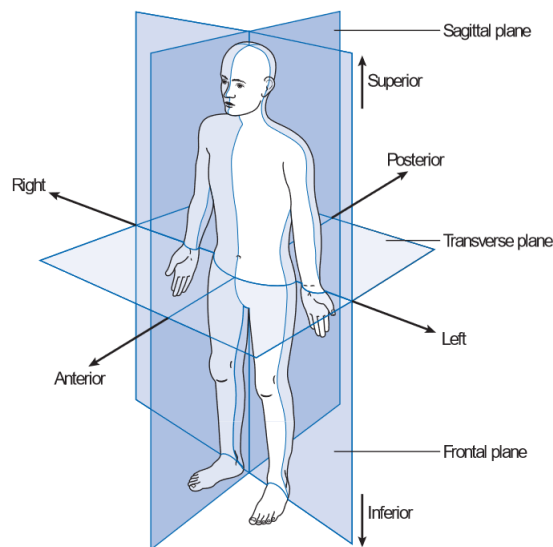


Figure 3.1: Anatomical position, with three reference planes and six fundamental directions. From [67]

These planes are used to characterise the movements of the various limbs of the human body.

To carry out an analysis of different prosthesis configurations, it is first necessary to understand the different tasks they will have to perform. These different tasks involve movements in all three planes, but it is above all movements in the sagittal plane that are most relevant to the development of mechanisms for prostheses.

Two types of task can be distinguished: passive and active tasks. Passive tasks are those that can be performed with a mechanical prosthesis. They do not necessarily require the delivery of motor energy. Active tasks require the motor to provide energy and carry out movement. The classification of a task is specific to each joint. For example, walking is a passive task for the knee joint, but an active task for the ankle and hip joints. Moreover, this classification is based on an average assessment during each task. This does not mean that there is never any motor work for passive tasks. This study focuses solely on the knee joint. For this joint, the passive tasks are walking, stand-to-sit and the stairs descent, and the active tasks are going sit-to-stand and the stairs ascent.

Walking is presented in detail. This is the most essential function that prostheses attempt to replicate for amputees. It is the most studied task in the literature ([67], [69]). Various graphs characterising the variables of the knee joint are presented: angle, torque, speed, vertical ground reaction force and the angle-torque relationship.

Then, the four other tasks are presented with the same graphs based on what has been found in the literature. Stairs Ascent is presented first followed by Stairs Descent and Stand-to-Sit and Sit-to-Stand tasks are presented together.

With the exception of reaction force graphs, the graphs presented in this section are based on a database initially created by H. Laloyaux and supplemented by J. Guiaux Brinon and L. Devillez. The database is based on data from various studies that have investigated the different activities of hip, knee and ankle joints. The various studies are :

Reference	Tasks	Reference	Tasks
[2]	Walking	[51]	Walking
[3]	Sit to Stand to Sit	[52]	Sit to Stand
[8]	Sit to Stand	[53]	Stairs Ascent and Descent
[14]	Sit to Stand	[57]	Walking
[22]	Sit to Stand	[58]	Sit to Stand
[23]	Sit to Stand	[63]	Walking
[27]	Walking	[64]	Walking
[31]	Walking	[71]	Walking
[38]	Walking	[39]	Walking

Table 3.1: References-Tasks of the database

For reaction forces, the source documents are referenced in the sections directly concerned. It was not possible to find the decomposition of the force in the three reference planes (sagittal, transverse and frontal) for each of the tasks. In order to present the five tasks in the same way, we have concentrated on the vertical component (sagittal plan), which is the most representative of the resultant reaction force in term of amplitude because it is much higher than the other two components.

Here are the conventions used in the following sections to describe each task. For ground-level walking, stairs ascent and descent, the start and

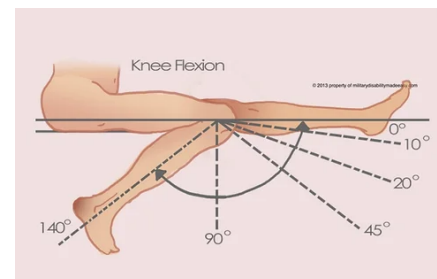


Figure 3.2: Convention used to describe the knee angle. From [16]

end of the cycle is set when the knee or the sole of the foot in question strike the ground. When the leg is in line with the thigh, the angle described by the knee joint is 0° . The leg is said to be in extension. This knee angle increases as the knee is bent. The leg is said to be in flexion. The convention chosen is that torque and speed are considered positive in extension and negative in flexion.

3.2 Ground-level Walking

Walking is the most basic form of locomotion, which is why it is the primary objective of any prosthesis. Walking involves a large number of joints (hip, knee, ankle, toes).

The gait cycle is defined as the time elapsed between two heel-to-floor contacts of the same foot. It lasts between 1 and 1.2 seconds depending on the person [2]. As shown in Figure 3.3, walking is made up of two main phases: the stance phase and the swing phase (third band in the illustration). These can be subdivided into sub-phases: Loading, Mid-Stance, Terminal Stance, Pre-Swing, Initiale Swing, Mid-Swing and Terminal Swing (first band in the illustration). The behaviour of the knee joint during these sub-sections is detailed below. During the stance phase, the person rests on their leg, which represents around 65 % of the cycle. During the swing phase, the person brings their leg from the back to the front, with the foot no longer in contact with the ground. This represents 35 % of the cycle. During a cycle, moments of double support alternate with moments of single support when one of the two legs is in swing phase.

The term "internal torque" used below is the torque of the knee whether in extension or flexion. The abbreviation vGRF stands for vertical Ground Reaction Force.

The reader is invited to consult the graphs in Figure 3.10 alongside the explanations and diagrams below for a complete understanding of the gait cycle. The cycle is shown for the right leg.

Loading phase (0-15%) : (Figure 3.4) This phase begins with the heel strike. A transfer of weight begins to take place from the left leg to the right leg. At the beginning of this phase, the knee generates an internal torque in the direction of flexion to avoid hyperextension at the end of the swing phase of the previous cycle. This torque opposes the moment of extension generated by the reaction force, so the vector passes in front of the knee joint. At the end of the phase, the reaction force vector passes behind the joint, generating a bending moment which the knee opposes with an internal extension torque that reaches the maximum value of the cycle (Figure 3.10b). The knee angle increases to reach a local maximum at the end of the phase. This is the Early Stance Flexion. It is initiated to allow continuity of movement and to absorb the shock (Figure 3.10a).

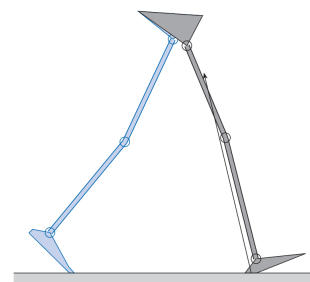


Figure 3.4: Loading Response. Right leg in grey, left leg in blue. [67]

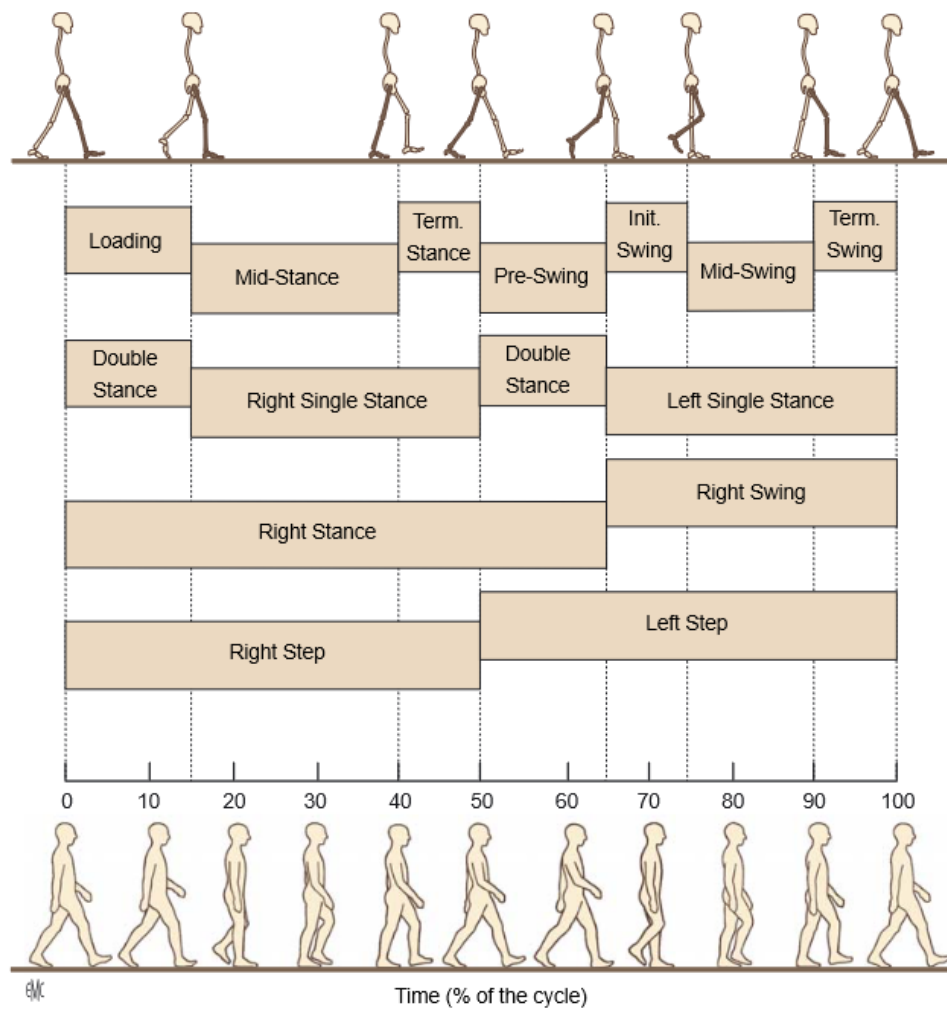


Figure 3.3: Gait cycle division. From [10] inspired by [69]. The gait cycle is presented for the right leg.

Mid-Stance phase (15-40%) : (Figure 3.5) This phase begins with the opposite foot (left) being lifted off the ground. The weight of the body is now supported solely by the right foot. Initially, the knee provides an extension torque to counterbalance the bending moment due to the ground reaction force vector passing behind the joint. Early Stance Flexion is maximal at the beginning of the phase and then decreases as the leg extends. In a second phase, when the swinging left leg moves back in front of the right, the force vector moves back in front of the joint causing an extension moment which the knee opposes with an internal flexion torque (Figure 3.10b).

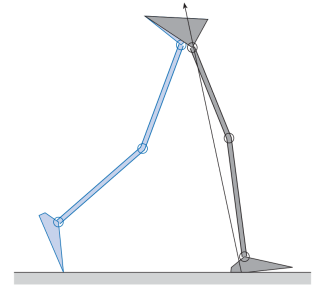


Figure 3.5: Mid-Stance - Opposite toe off. Right leg in grey, left leg in blue. [67]

Terminal Stance phase (40-50%) : (Figure 3.6) The swinging left leg is back in front of the right. The joint is in front of the GRF vector. The force vector generates a bending moment which the joint opposes with an internal extension torque (Figure 3.10b).

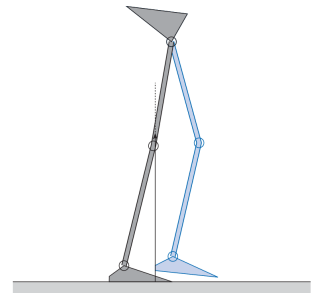


Figure 3.6: Terminal Stance. Right leg in grey, left leg in blue. [67]

Pre-swing phase (50-65%) : (Figure 3.7) This phase begins with an attack from the heel of the other foot (the left in this case). This is the resumption of the double contact. The leg begins to flex (Figure 3.10a). The joint always produces an extension torque to slow down the natural flexion transferred from the hip joint, which lowers. The reaction force is maximal at the beginning of the phase when the ankle and toes give the impulse to lift the foot off the ground, then falls to zero at the end of the phase (Figure 3.10e).

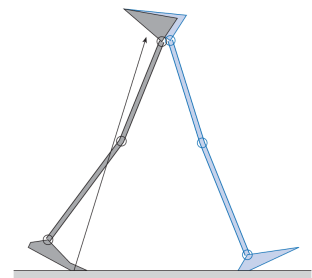


Figure 3.7: Pre-swing phase begins at opposite initial contact. Right leg in grey, left leg in blue. [67]

Initiate Swing Phase (65-75%) : (Figure 3.8) This is the start of the swing phase, which begins as the toes lift off the ground and the reaction force disappears. The hip raises the leg and brings it forward while the knee flexes to lift the foot off the ground. The angle is maximal during this phase (Figure 3.10a).

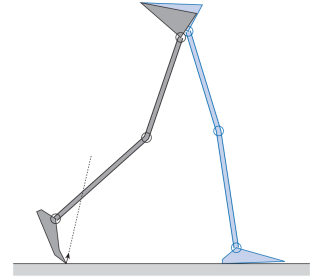


Figure 3.8: Initiate Swing Phase - Toe off. Right leg in grey, left leg in blue. [67]

Mid-Swing (75-90%) : (Figure 3.9) This phase begins when the leg crosses the supporting leg to return to the front. During this phase, the internal torque goes into flexion to progressively slow down the extension of the leg. With the leg in extension, the angle decreases rapidly and the speed is positive (Figure 3.10a, Figure 3.10c).

Terminal Swing (90-100%) : This phase begins when the speed of extension reaches its maximum and decreases until the end of the cycle under the action of the internal torque of the joint, which is flexed to slow down the extension of the leg (Figure 3.10b, Figure 3.10c).

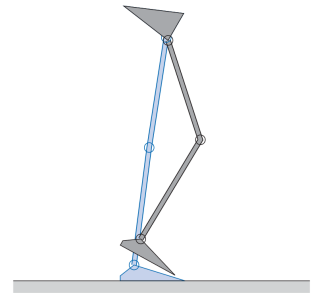


Figure 3.9: Mid-Swing - Toe off. Right leg in grey, left leg in blue. [67]

The curves of the physical quantities characterising the behaviour of the **knee** during a walking cycle are shown in Figure 3.10.

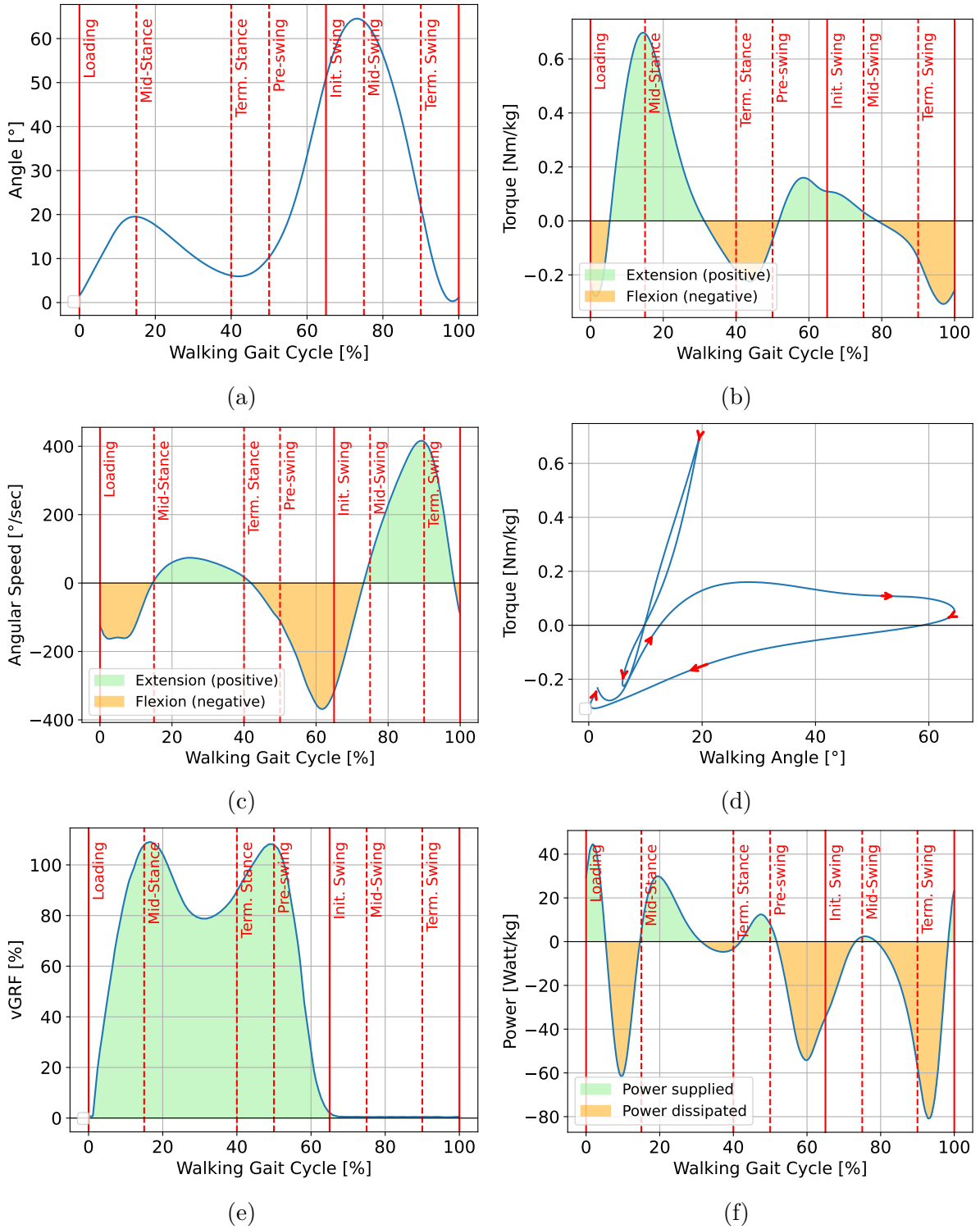


Figure 3.10: Profiles of **knee** variables during Walking Gait Cycle (a) Angle (b) Torque (c) Angular Speed (d) Angle-Torque Relation (e) vertical Ground Reaction Force (vGRF) (f) Power. Each profile is expressed as a percentage of the gait cycle, with torque and power normalised to weight and vGRF expressed as a percentage of $9.81 \times \text{weight}$. (d) The angle-torque relationship highlights the elastic aspect of the stance phase. The relationship between the two is linear during this phase. This characteristic is often exploited in prostheses by placing a spring to reduce energy consumption.

3.3 Stairs Ascent

Stairs ascent differs from walking in its energy balance. Walking is a dissipative task for the knee, whereas stairs ascent requires the joint to produce energy. It is also characterised by greater torque and angular speed. The complete cycle is described in stages below. It is also divided into a stance phase and a swing phase, which are subdivided into the following sections: Weight Acceptance, Pull-up, Forward Continuance, Foot Clearance and Foot Placement [50], [26]. Figure Figure 3.11 provides an overview of these stages and Figure Figure 3.13 presents the various characteristic graphs (angle, torque, angular speed, the angle-torque relationship and the vertical component of the ground reaction force) obtained from the database of J. Guiaux Brinon and L. Devillez. It should also be noted that the ascent and descent of stairs are influenced by the length and height of the steps, although the general shape of their curves is preserved [53].

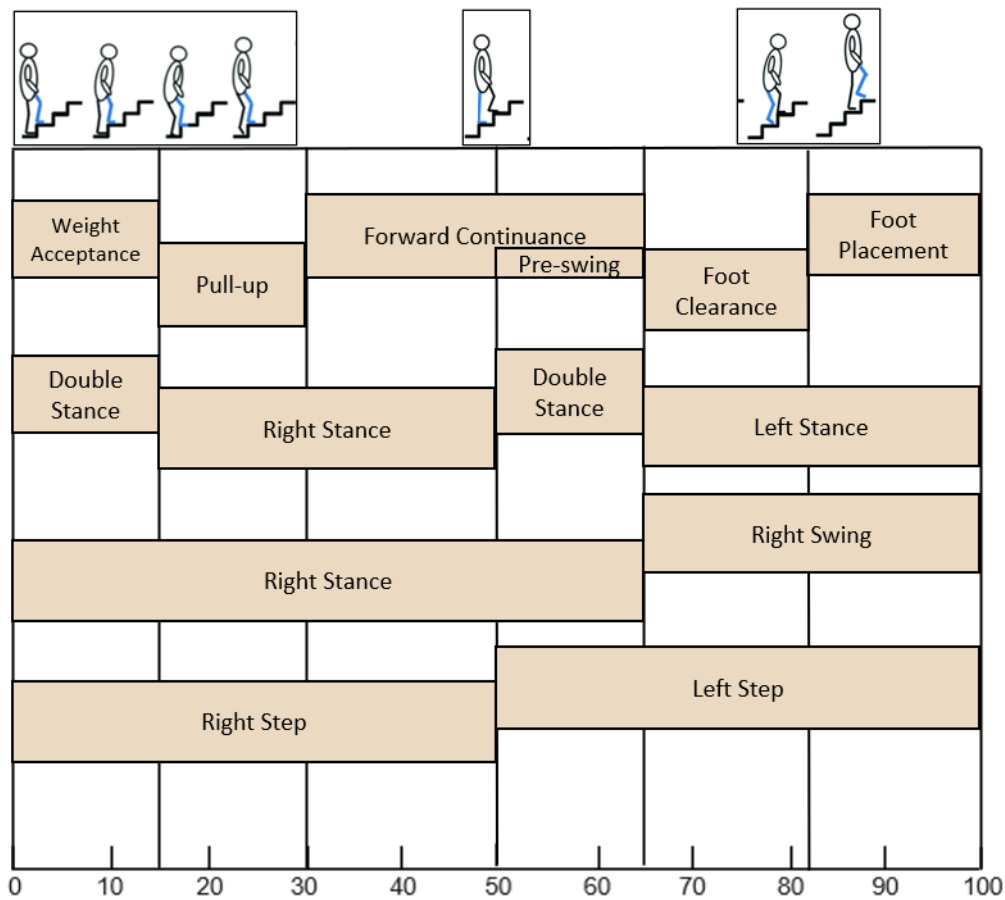


Figure 3.11: Detailed Stairs Ascent Cycle. Diagram inspired by [10] and [69]. Illustrations from [50]

The reader is invited to consult the graphs in Figure 3.13 alongside the explanations and diagrams below for a complete understanding of the stairs ascent cycle.

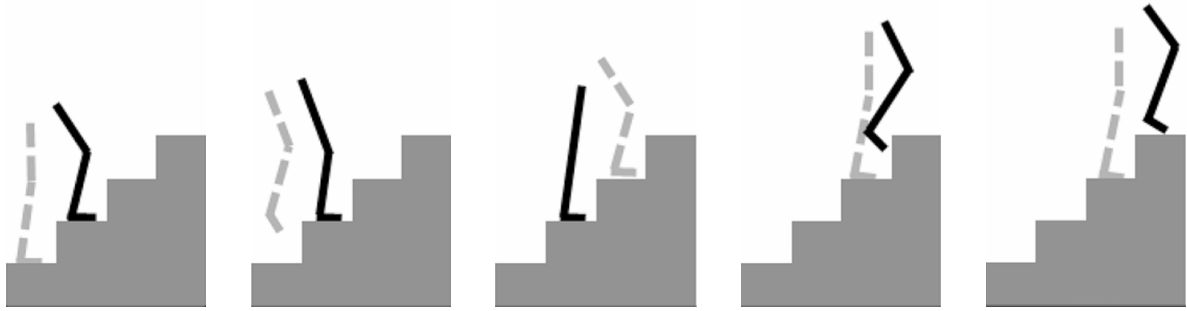


Figure 3.12: Illustration of the phases of stairs ascent : (a) Weight Acceptance (b) Pull Up (c) Forward Continuance (d) Foot Clearance (e) Foot Placement. From [26]

- **Weight Acceptance (0-15%)** : (Figure 3.12(a)) A leg cycle lasts between 1.3 and 1.5 seconds, depending on the individual. It begins with the sole of the foot coming into contact with the step. This is the first part of the stance phase, the weight acceptance. The body weight is transferred from one leg to the other. The knee produces a significant extension torque to oppose the natural bending torque due to the load and to initiate the lifting of the body. The speed is also in extension. Peak power is high and energy is consumed.
- **Pull Up (15-32.5%)** : (Figure 3.12(b)) This phase begins with the foot lifting off the left leg (the dashed one in the illustration). All the weight is now carried by the right leg. Torque is now at its maximum (Figure 3.13b). The torque and speed (Figure 3.13c) are still in extension to lift the body. This is the impulse phase where the power is at its highest. The reaction force reaches a peak(Figure 3.13e).
- **Forward Continuance (32.5-50%)** : (Figure 3.12(c)) During this phase, the knee torque changes sign and goes into flexion to slow the leg and avoid hyperextension. Speed decreases.
Forward Continuance - Pre-swing(50-65%) : This second part of the Forward Continuance phase begins with the resumption of double contact as the sole of the opposite foot makes contact with the step. The torque and speed of the knee enters flexion to initiate the swing phase and lift the leg above the step. At the start of this phase, the angle is minimal (Figure 3.13a). At the beginning of the phase, the foot pushes on the ground to take off, the vertical reaction force is at its maximum and then falls to zero at the end of the phase (Figure 3.13e).
- **Foot Clearance (65-82.5%)** : (Figure 3.12(d)) This phase begins when the foot is no longer in contact with the ground. The leg enters the swing phase. The angular speed of the joint is at its maximum in flexion to pass the step. During this phase, the torque changes sign and returns to extension to slow down the flexion and initiate extension of the leg (Figure 3.13b).
- **Foot Placement (82.5-100%)** : (Figure 3.12(e)) This phase begins when the angle is maximum (Figure 3.13a). This is the leg extension phase, the speed becomes positive again (Figure 3.13c). This is the second half of the swing phase. The torque return to flexion to brake the leg in order to avoid a too great extension in relation to the next step and to avoid abrupt contact with the ground. Both parts of the swing phase have the same duration.

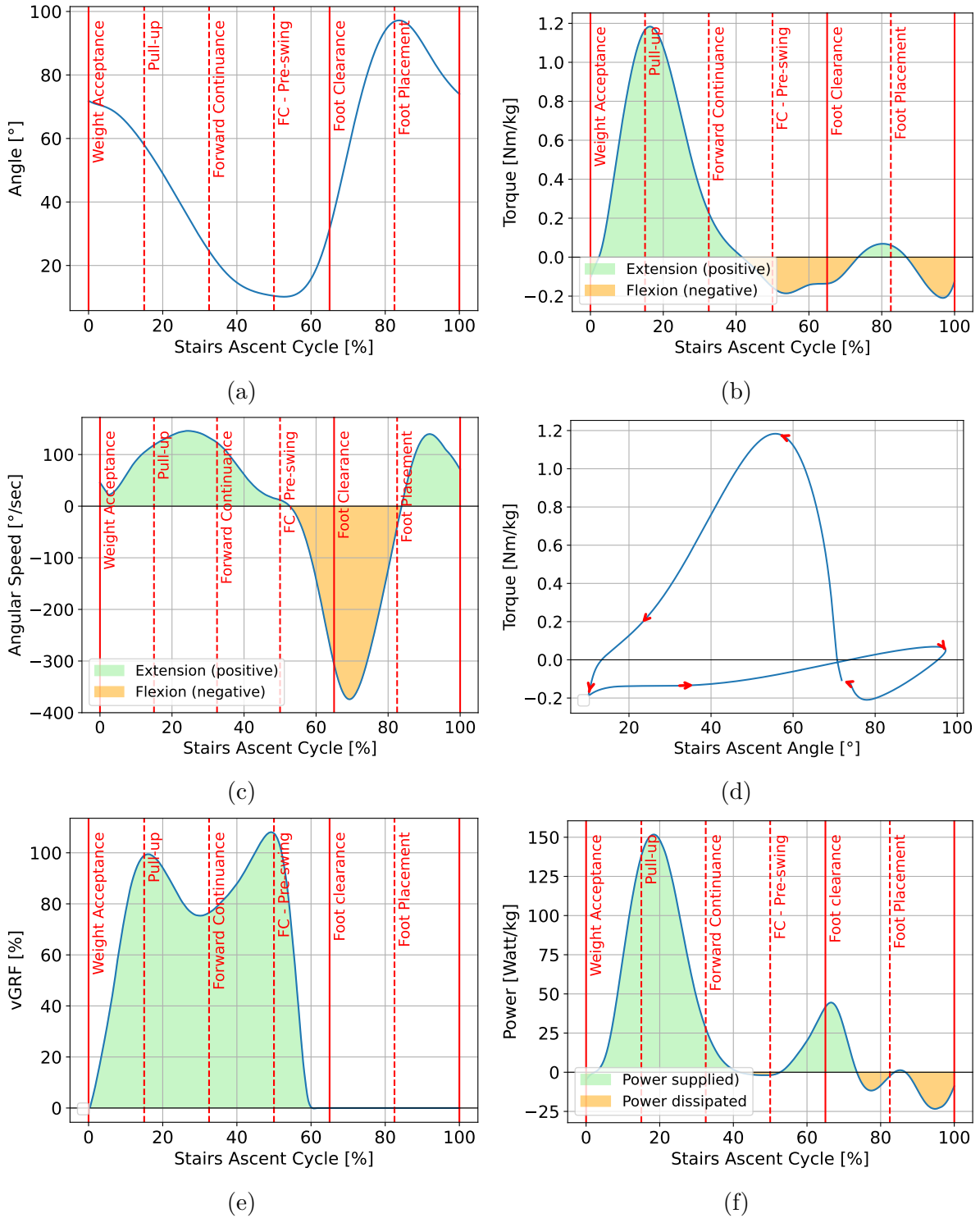


Figure 3.13: Profiles of **knee** variables during Stairs Ascent Cycle (a) Angle (b) Torque (c) Angular Speed (d) Angle-Torque Relation (e) vertical Ground Reaction Force (vGRF) [1], [59] (f) Power. Each profile is expressed as a percentage of the gait cycle, with torque and power normalised to weight and vGRF expressed as a percentage of $9.81 \times \text{weight}$. (e) shows that unlike walking, there is no linear relationship returning to the same point between torque and angle in stairs ascent that could be replicated by a single spring.

3.4 Stairs Descent

The single-leg stair descent cycle is also broken down into a stance phase and a swing phase. Within the stance phase, there are 3 parts: Weight Acceptance, Forward Continuance and Controlled Lowering. Within the swing phase, there are 2 parts: Leg Pull-Through and Foot Placement. The stair descent cycle lasts an average of 1.2 seconds. Here too, the stairs descent can be performed step by step with both feet touching each step, but this is not a natural behaviour. The reader is invited to consult the graphs in Figure 3.15 alongside the explanations and the sketch below for a complete understanding of the gait cycle.

- **Weight Acceptance (0-15 %)** : This phase begins with the toe touching the step below. The leg is stretched to make contact with the step and then flexes to absorb the impact. The speed is conventionally negative as the knee enters flexion (Figure 3.15e). The weight is gradually transferred from one leg to the other. Throughout the stance phase, the joint produces an internal extension torque to slow the descent and ensure stable contact at the end of the stance phase of the previous cycle (Figure 3.15b). The ground reaction force reaches its maximum. This maximum is higher than for walking and stairs ascent because the body is accelerated by gravity (Figure 3.15e).
- **Forward Continuance (15-30 %)** : This phase begins when the foot of the opposite leg takes off. The body is now in single stance as the opposite leg is in swing phase.
- **Controlled Lowering (30-50-65 %)** : The beginning of this phase is marked by an increase in flexion. The internal torque opposes the speed to control the descent. It reaches its maximum value when the double contact is resumed and the opposite leg enters the stance phase (50%). This event marks the start of the pre-swing phase. The weight is transferred on the other leg and the torque decreases.
- **Leg Pull-Through (65-82.5 %)** : This phase marks the start of the swing phase when the foot leaves the top step. During this phase the speed changes from flexion to extension to bring the leg forward and allow the leg to pick up the next step. The opposite leg is in single stance.
- **Foot Placement (82.5-100 %)** : During this phase, the speed of extension is reduced by the internal torque of flexion to avoid hyperextension and to look for the next step smoothly (Figure 3.15c, Figure 3.13b). The leg straightens, the angle of the joint returns to zero (Figure 3.15a).

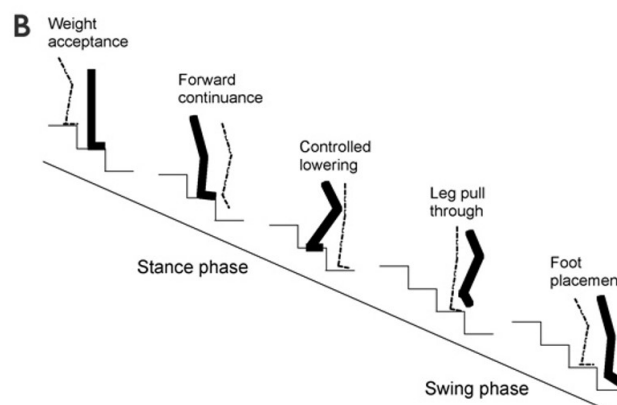


Figure 3.14: Stairs Descent Sktech. From [extremityreview]

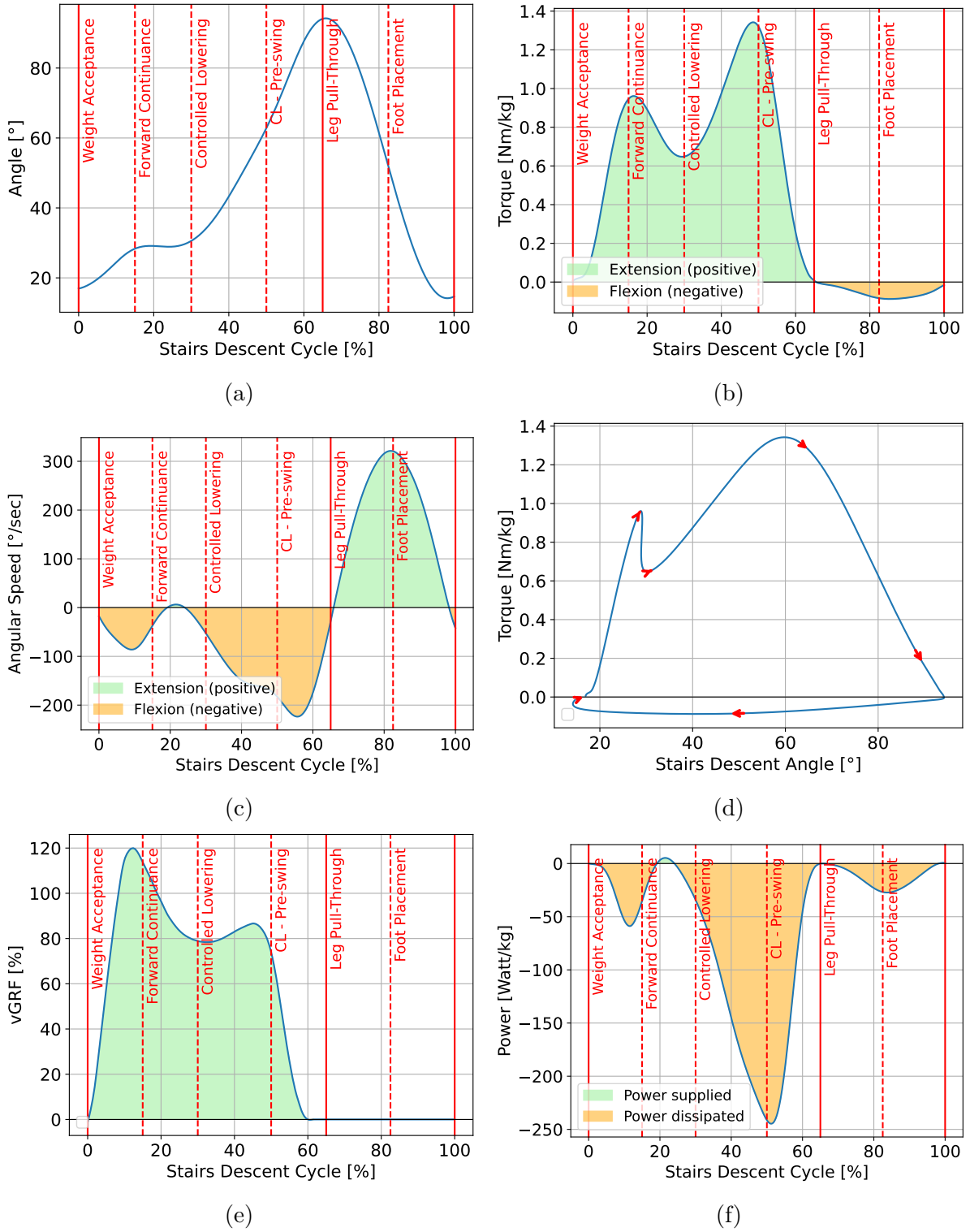


Figure 3.15: Profiles of **knee** variables during Stairs Descent Cycle (a) Angle (b) Torque (c) Angular Speed (d) Angle-Torque Relation (e) Vertical Ground Reaction Force (vGRF) [1], [59] (f) Power. Each profile is expressed as a percentage of the gait cycle, with torque and power normalised to weight and vGRF expressed as a percentage of $9.81 \times \text{weight}$.

3.5 Sit-to-Stand & Stand-to-Sit

The curves of the physical quantities characterising the knee for sit-to-stand and stand-to-sit are presented in figure Figure 3.16. These tasks last on average 1.2 seconds. These are two non-cyclic tasks. In fact, they could be seen as half-periods of the same cycle.

The angle curves are highly symmetrical (Figure 3.16a). The red dotted line at 50% of the cycle helps to visualise this. The angle-torque relationships of Sit-to-Stand and Stand-to-Sit follow the same pattern (Figure 3.16d). In both cases, the internal torque is in extension (Figure 3.16b) and is always high when the bending angle is large (Figure 3.16d). This is to oppose the bending moment generated by the weight of the body. As the movements are opposite, so are the speeds (Figure 3.16c).

Sit-to-Stand speed and torque have the same sign, which corresponds to an active task where power is supplied by the joint. Stand-to-Sit speed and torque have opposite sign, which corresponds to a passive task where power is dissipated by the joint.

Concerning the vertical component of the reaction force, only data for standing could be found. No data was found for Stand-to-Sit, but given the high correlation with standing, we can assume that the curve follows the inverse pattern (Figure 3.16e). To obtain the vertical reaction force experimentally, the procedure asks participants to keep their arms crossed over their torso. The movement is initiated when the force decreases and ends when the force is equal to the body weight. The maximum reaction force per body weight reflects the maximum downward force that pushes the body upwards [56], [5], [4].

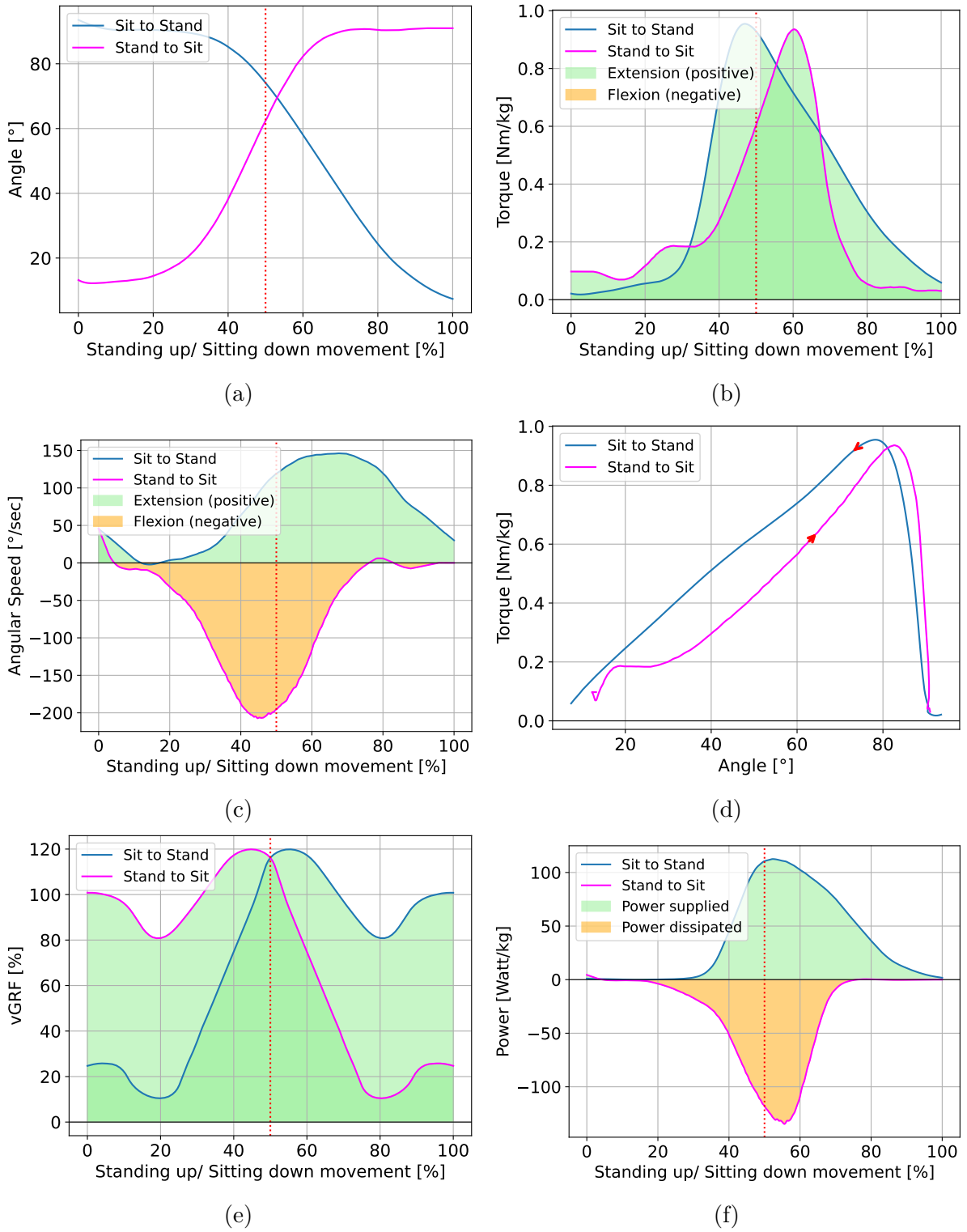


Figure 3.16: Profiles of **knee** variables during Sit-to-Stand and Stand-to-Sit Cycle (a) Angle (b) Torque (c) Angular Speed (d) Angle-Torque Relation (e) vertical Ground Reaction Force (vGRF) (f) Power. Each profile is expressed as a percentage of the gait cycle, with torque and power normalised to weight and vGRF expressed as a percentage of $9.81 \times \text{weight}$.

3.6 Tasks comparison and prioritization for prosthesis design

The previous sections presented each of the tasks. In this section, the aim is to put these analyses into perspective for active prostheses. A short comparison is made in order to evaluate the maximum ranges of each of the physical quantities required to produce a prosthesis that meets the five tasks.

3.6.1 Angle curves comparison

Figure 3.17(a) repeats the angle curves presented earlier in Figures Figure 3.10a, Figure 3.13a, Figure 3.15a and Figure 3.16a for ease of comparison. The angular amplitudes of each task are fairly similar if walking is excluded. The prosthesis should therefore be able to cover an amplitude of $100[^\circ]$.

3.6.2 Angular speed curves comparison

The speed curves for the five tasks (Figure 3.10c, Figure 3.13c, Figure 3.15c and Figure 3.16c) are plotted together in Figure Figure 3.17(b).

In terms of angular velocity of the joint, walking and going up and down stairs have much higher speeds than the other two tasks.

Sit-to-stand and stand-to-sit are slower because the body starts at rest and has no inertia. The stand-to-sit speed is higher than the sit-to-stand speed due to gravitational acceleration. The human starts by letting himself fall and then brakes before making contact with the seat.

3.6.3 Torque curves comparison

The graph in figure Figure 3.17(c) shows the torque curves for the five tasks already presented in the figures (Figure 3.10b, Figure 3.13b, Figure 3.15b and Figure 3.16b). Torque is an important factor to ensure for every task. It guides the choice of motor and reduction mechanism. Stairs descent and ascent require the highest torques.

3.6.4 Power curves comparison

Descending the stairs is the task that dissipates the most power followed by stand-to-sit. These two tasks require very little positive power compared with the other 3 tasks. Stairs ascent has the highest positive power peak, while stairs descent presents the highest negative power peak.

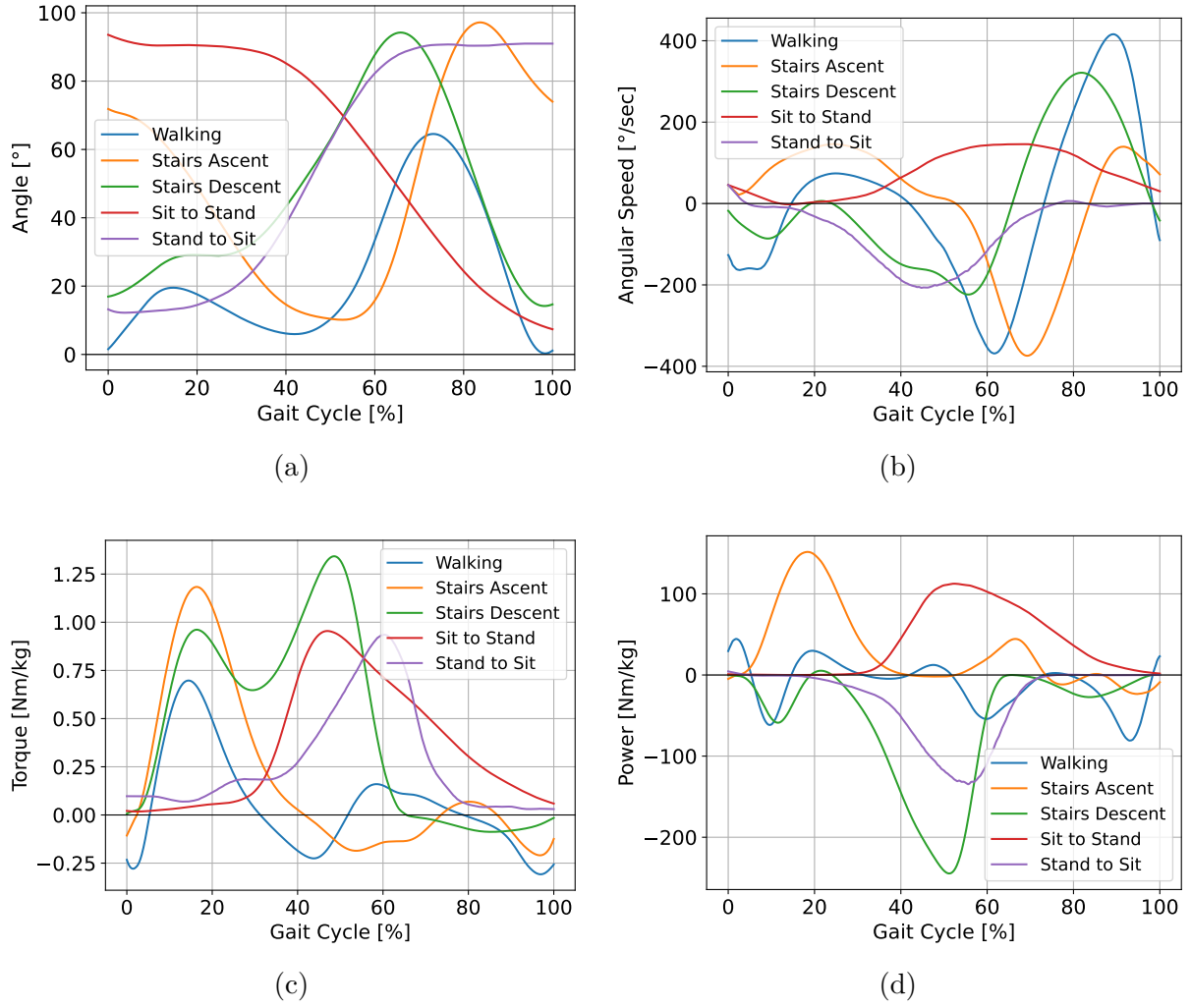


Figure 3.17: (a) Angle curves (b) Angular speed curves (c) Torque curves (d) Power curves

Figure 3.17(d) shows the power curves for the various tasks (for the conventions of the graph, a positive power corresponds to an input of energy from the motor and a negative power corresponds to dissipation or operation of the motor in generator mode). The power curve is an important datum, as it guides the choice of motor for the prosthesis. Motor power is the product of motor torque and motor speed.

3.6.5 Conclusion

The analysis to date is summarised in the table Table 3.2 below. This table shows the extreme values for angle, torque, speed and positive and negative power for five tasks. The cell with the maximum value has been highlighted with a pink background.

Task	Angle [°]	Speed [°/sec]	Torque [Nm/kg]	Power Max [W/kg]	Power Min [W/kg]
Walking	64.5	416.1	0.69	44.5	-80.8
StairsAsc	97.1	374.1	1.18	151.7	-23.4
StairsDsc	94.2	321.4	1.34	5.29	-244.6
SiTSt	93.5	146.1	0.95	112.55	-0.09
StTSi	90.9	207.2	0.93	4.44	-134.68

Table 3.2: Comparison of the five tasks' biomechanical data

The comparison table shows that the study of walking and stairs ascent is sufficient for the study of energy consumption and the choice of motor for a prosthesis.

In fact, as far as active tasks are concerned, if the prosthesis can be used for stairs ascent, then sit-to-stand will be possible too. The maximum torque and speed of the prosthesis are lower than those for stairs ascent. Furthermore, neither the power dissipated nor the power supplied by sit-to-stand exceeds the active power of stairs ascent.

Furthermore, sit-to-stand is not a repetitive task. It is performed without consecutive repetition, unlike stairs ascent. There are often several steps to ascend, whether at home, at work or in a public place. For ten or so steps to be ascended, the prosthesis of a unilateral amputee will have to cover five of them consecutively. This is a more demanding task than sit-to-stand in terms of design, particularly in terms of the risks of the motor overheating and autonomy.

As far as passive tasks are concerned, if you can walk and ascend stairs, stand-to-sit will be too problem-free. The speed of the latter is lower than that of the other two, and the torque is lower than that of stairs ascent. The same goes for passive power, which remains lower than active power for stairs ascent.

The situation is more delicate for stair descent because the torque is the highest of the five tasks and the negative power is higher than the active power for stair ascent. As this is dissipated power, it does not affect the prosthesis battery consumption. On the other hand, it can lead to overheating of the motor-generator as the cycle is repeated. It is important to reduce the peak power dissipation and ensure that the motor is properly cooled. In terms of torque, the difference with stairs ascent is fairly small. Despite its limitations, descending stairs remains a passive task that can also be performed passively without the intervention of a motor. This may be preferable to increase independence. Motorization provides assistance, which may be comfortable but is not essential.

Chapter 4

Project Specifications

The first chapter provided a clearer picture of the active prosthesis landscape. It highlighted the main aspects of each prototype. The second chapter focused on the different locomotion tasks. These were presented and compared to understand them better and identify the biomechanical challenges involved. Building on the first two chapters and addressing the research question, this chapter establishes the graph of objectives and the specifications (main functions and constraints) for developing a kinematic chain that could be incorporated into active knee prostheses.

4.1 Graph of Objectives

Figure 4.1 displays the graph of objectives for the design of an energy-efficient active knee prosthesis. Numerical ranges for the weight and autonomy are based on what is currently reached by state-of-art active prosthesis described in subsection 2.2.5.

To achieve the crucial design objective of energetic efficiency, the inclusion of (a) passively activated mechanism(s) will be investigated. This approach entails that any mechanism integrated into the kinematic chain to reduce energy consumption and enhance autonomy must operate without additional actuators, even if it requires to change from one state to another during a certain fraction of the cycle defined by triggering and releasing points. The only actuator present in the prosthesis is the main motor which initially drives the kinematic chain.

One of the ultimate objectives is to reach an autonomy of 5-8 hours of continuous operation while maintaining a lightweight design (2.5-2.7 [kg]). Other objectives about safety, quality, and ergonomics are considered, as well as average ranges associated with transfemoral amputees.

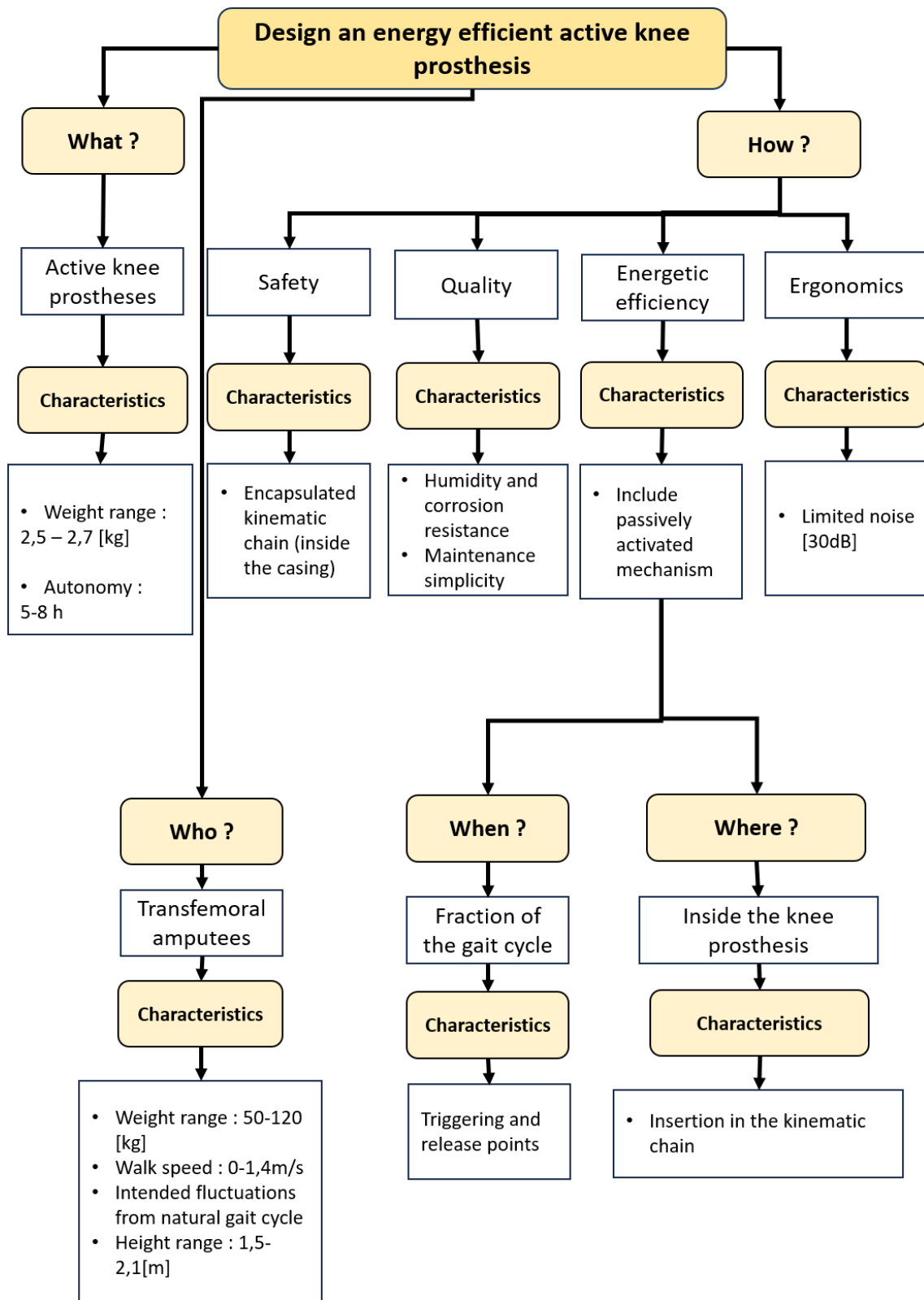


Figure 4.1: Graph of objectives

4.2 Specifications

Table 4.1 presents the main functions of any passively activated mechanism incorporated into the kinematic chain with the objective of modifying it to improve the energetic efficiency of the prosthesis. Performance criteria include the mechanism activation time, reliability, and margin as well as its capacity to be reversible.

In contrast, Table 4.2 details the main constraints imposed on the kinematic chain of the active knee prosthesis, with numerical ranges inspired again by state-of-the-art devices. These constraints are crucial for ensuring the prosthesis's performance and reliability, taking into account a broad range of performance criteria. For constraint C6, a lifespan of 5 years is considered to compute the approximate number of walking and stairs ascent cycles the prosthesis must be able to achieve.

Main Functions	
Performance criteria	Performance level
MF1: Modify passively the kinematic chain	
1. Activation time 2. Activation reliability 3. Activation margin 4. Reversibility	1. Rapid 2. The mechanism follows the 6σ method. It can not fail more than 3.4 cycles <i>per million</i> [68]. 3. The mechanism is activated close to the triggering point. 4. The mechanism can operate in both directions. The triggering point can become the release point and vice-versa.
MF2: Re-establish passively the kinematic chain	
1. Activation time 2. Activation reliability 3. Activation margin 4. Reversibility	1. Rapid 2. The mechanism follows the 6σ method. It can not fail more than 3.4 cycles <i>per million</i> . 3. The mechanism is activated close to the release point. 4. The mechanism can operate in both directions. The release point can become the triggering point and vice-versa.

Table 4.1: Main Functions of the passively activated mechanism(s)

Main Constraints	
Performance criteria	Performance level
C1 : The kinematic chain must be lightweight	
Weight range	• 50-60 % of prosthesis weight (For comparison, the CSEA weighs 2.7 [kg] with a kinematic chain of 1.5[kg] (55%) [55])
C2 : The kinematic chain must have a low encumbrance	
Max. vertical length	• 220 [mm] corresponding to 50% of the average shank's length, i.e. 12.3% of amputee's average height of 1.8 [m] [70] (For comparison, vertical length including casing is 285 [mm] for the CSEA [55], 277 [mm] for the Power Knee [44], 298 [mm] for the Utah Bionic Leg [61], 289 [mm] for the Ottobock C-Leg [43])
C3 : The kinematic chain must be quiet	
Noise limit	• 30 [dB] (Equivalent to a whisper [20], ensuring discretion for the patient)
C4 : The kinematic chain must be compliant	
Shock absorption	• Allow a certain degree of flexibility in the kinematic chain to absorb energy from shocks and impact
C5 : The kinematic chain must be able to work normally at different temperatures	
Temperature range	• [-10°C ; 40°C] (Enabling use of the prosthesis in a broad range of environments and conditions)
C6 : Lifetime of the kinematic chain	
Number of cycles	• Walking¹: $\frac{1500}{2} \left[\frac{\text{cycle}}{\text{day}} \right] \cdot 365 \left[\frac{\text{day}}{\text{year}} \right] \cdot 5 [\text{year}] = 1.4 \cdot 10^6 [\text{cycle}]$ • Stairs ascent²: $50 \left[\frac{\text{cycle}}{\text{day}} \right] \cdot 365 \left[\frac{\text{day}}{\text{year}} \right] \cdot 5 [\text{year}] = 92\,000 [\text{cycle}]$
C7 : The kinematic chain must be robust	
Max. acceptable force	• $9.81 \cdot 120 \cdot 1.5 = 1766 [\text{N}]$ (Support the max. weight of 120 [kg] with a safety factor of 1.5³)
C8 : The kinematic chain must be protected from the environment	
Weatherproof and dust-proof	• IP value of 54, i.e. protection against dust deposits (solid) and protection against splashing water (liquid) [6]. (For comparison, Ottobock's C-Leg 4 and Genium have an IP value of 67 but Ottobock's Kenovo has an IP value of 22 [45])
C9 : The kinematic chain must be able to meet the maximum values identified on the biological curves of the lower limb tasks. (based on Table 3.2).	
• Flexion angle range • Max. torque • Max. speed	• [0° ; 100°] • 1.4 [N/kg] • 420 [°/sec]

Table 4.2: Main Constraints on the kinematic chain

¹ An amputee takes on average 1500 steps per day [19], corresponding to $\frac{1500}{2} \left[\frac{\text{cycle}}{\text{day}} \right]$

² $50 \left[\frac{\text{cycle}}{\text{day}} \right]$, corresponding to 100 staircase steps per day, are considered in this work

³ SF = 1.25 to 1.5 for exceptionally reliable materials used under controllable conditions and subjected to loads and stresses that can be determined with certainty - used almost invariably when a low weight is particularly important [25].

Chapter 5

Elastic actuators - Kinematic chains modeling and optimization

One common way to improve the energetic efficiency of active transfemoral prostheses is through the use of elastic actuators. This usually consists of integrating series or parallel elastic elements with an electrical motor. By incorporating springs capable of storing energy and returning it later, the power flows in the motor can be reduced for various lower limb tasks, thereby enhancing the efficiency and autonomy of the active prosthetic device.

Elastic actuators offer a diverse range of possible configurations. In this research, configurations with a maximum of 2 elastic elements were considered, some in which a spring or even the motor can be engaged or disengaged at strategic instants of the gait cycle. Indeed, situations where an elastic actuator would require more energy than a direct drive actuator can be avoided by disengaging a spring or by preserving simply the elastic contribution and bypassing the motor. The 6 configurations investigated are as follows:

- Series Elastic Actuator (SEA)
- Clutchable Series Elastic Actuator (CSEA)
- Parallel Elastic Actuator (PEA)
- Clutchable Parallel Elastic Actuator (CPEA)
- Double Parallel Elastic Actuator (2PEA)
- Parallel and Series Elastic Actuator (PSEA)

For each of these configurations, a simulation model is developed to optimize the motor's electrical performance. Once established, the performance of each actuator configuration is first optimized for ground-level walking alone. Subsequently, a combined optimization is conducted for both walking and stairs ascent. Hence, the simulation only focuses on two tasks among the five presented in chapter 3. The motivation underlying this choice is detailed in section 3.6.

To recall, ground-level walking can be quite well achieved with passive devices but can't be entirely replicated without energy input. For instance, during the early-stance flexion phase which is important for shock absorption at heel strike and maintaining walking

speed, passive mechanisms can mimic the knee flexion but fail to reproduce the torque profile of a biological knee [30].

The second task, stairs ascent, presents a more significant challenge due to the high torque and power peaks required during the weight acceptance and pull-up phases (see Figure 3.13b). These demands highlight the necessity for an active device to enable an individual to stand up and climb steps effectively.

This chapter concludes with a detailed discussion and comparison of the results for each configuration, providing insights into their potential to reduce the energy consumption of transfemoral knee prostheses.

5.1 Biomechanical data

To perform simulations, the biomechanical data for walking and stairs ascent were obtained by extracting averaged angle, torque, and power curves from the database initially created by H. Laloyaux, supplemented by L. Devillez and J. Guiaux Brinon. It should be noted that values for the torque and power were given as normalized values (respectively $\left[\frac{Nm}{kg\ user}\right]$ and $\left[\frac{W}{kg\ user}\right]$). To infer the motor dynamics in absolute values, data were multiplied by $80\ [kg]$, which was chosen as a weight close to the average of the weight range targetted in the specifications ($50 - 100\ [kg]$). This led to the obtention of θ_j , T_j , and P_j , respectively the biological knee joint angle, torque, and power in the sagittal plane for a specific task. The database also allowed to extract a normal-speed cycle time of $1.0\ [s]$ for walking and $1.4\ [s]$ for stairs ascent.

The authors are aware that this choice represents a solid basis but is not ideal, since the deviation from the averaged weight or normal gait speed could impact the biomechanical curves. Additionally, to compensate for the loss of their lower limb, transfemoral, amputees may develop gait asymmetry and intended fluctuations from biological behaviors, as explained in section 2.3.

5.2 Series Elastic Actuators

As demonstrated in chapter 3, the knee torque-angle relationship in the sagittal plane during walking exhibits a spring-like behavior over certain phases of the cycle, especially the early-stance flexion (ESF) phase (see Figure 3.10d). A torsional spring could thus be used to render this quasi-linear relationship. Connecting a torsional spring in series with an electrical motor is commonly referred as a Series Elastic Actuator (SEA), whose kinematic diagram is shown in Figure 5.1.

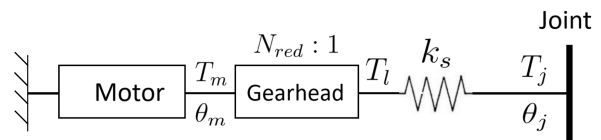


Figure 5.1: Series Elastic actuators (SEA) kinematic diagram

5.2.1 SEA model

To model a SEA configuration, the following equations are used:

$$\Delta\theta_{k_s} = -\frac{T_j}{k_s} \quad (5.1)$$

$$\theta_m = N_{red} \cdot (\theta_j - \Delta\theta_{k_s}) \quad (5.2)$$

$$\omega_{k_s} = \frac{d\Delta\theta_{k_s}}{dt} \quad (5.3)$$

$$P_{k_s} = \omega_{k_s} \cdot T_j \quad (5.4)$$

$$T_l = T_j \quad (5.5)$$

with $\Delta\theta_{k_s}$ the series spring deformation defined by its stiffness k_s and its torque which is equal to the torque applied at the joint ($T_{k_s} = T_j$). The motor position along the cycle depends on the joint position θ_j , the series spring deformation, and the global reduction ratio N_{red} which, in this chapter, is simply equal to the reduction ratio of the gearhead¹. Equation 5.3 gives the series spring deformation speed ω_{k_s} useful to compute its mechanical power P_{k_s} through Equation 5.4. T_l is the torque load (also called the resisting torque), namely the torque applied by the load at the gearhead output shaft that must be overcome by the motor to drive the knee joint. In the case of a SEA, T_l is simply equal to the joint torque.

The motor speed ω_m and acceleration a_m are obtained by the first and second-time derivatives of the motor position:

$$\omega_m = \frac{d\theta_m}{dt} \quad (5.6)$$

$$a_m = \frac{d\omega_m}{dt} = \frac{d^2\theta_m}{dt^2} \quad (5.7)$$

As one can deduce from Equation 5.2, the deformation of the spring induces a reduction of the required motor speed, but the motor still has to provide reactionary torque even when the spring alone provides the joint torque-angle linear relationship. Since it generates this torque at a lower speed than it would have in a direct drive configuration, the motor mechanical power is reduced; nevertheless, the motor electrical power is not reduced as much as the mechanical one.

Indeed, as depicted in Figure 5.2, the electrical power P_e is the mechanical power P_{mech} increased by the Joule's power losses P_J ². These losses are proportional to the square of the current, itself proportional to the motor torque. Therefore, the high reactionary torque, typically needed during the stance phase, will lead to high Joule power losses.

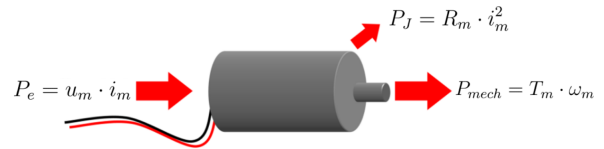


Figure 5.2: Energy conversion in an electrical Direct Current (DC) motor. Adapted from [32].

¹Let's note that any reduction ratio appearing in this report is defined as the input speed divided by the output speed.

²It refers to the electrical energy lost as heat due to the resistance in the motor's windings.

5.2.2 CSEA model

One way to avoid the high Joule's power losses during the stance phase is by adding a clutch in the SEA kinematic chain to form a so-called Clutchable Series Elastic Actuator (CSEA), whose kinematic diagram is displayed in Figure 5.3. The principle is to include a clutching element in parallel to the motor and engage it during the ESF phase to provide the reaction torque on the motor shaft and let the series spring render the conservative torque-angle relation while bypassing the motor.

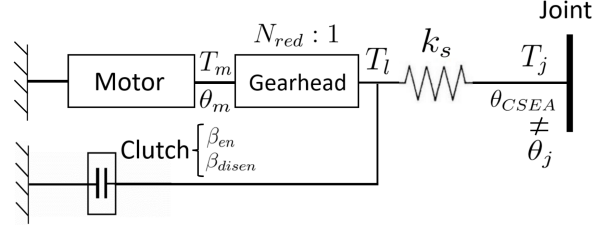


Figure 5.3: Clutchable Series Elastic actuators (CSEA) kinematic diagram

This configuration has been implemented in previous researches (e.g. [55]) and showed a substantial reduction in electrical energy consumption for walking, thanks to the fact that their electromagnetic clutch was far more energy efficient for providing the reactionary torque than the low-speed motor (electric motors are known to be inefficient at low speeds). A possible improvement to their work would be to engage and disengage the clutch passively.

The mathematical equations used to model the CSEA configuration are the same as the ones derived for the SEA, except Equation 5.2 that becomes:

$$\theta_m = \begin{cases} N_{red} \cdot (\theta_j - \Delta\theta_{k_s}) & \text{if clutch disengaged} \\ \theta_{m,bp} & \text{if clutch engaged} \end{cases} \quad (5.8)$$

where $\theta_{m,bp}$ represents the fixed motor position at the moment when the clutch engages and bypasses the motor. In the simulation, the moments of engagement β_{en} and disengagement β_{disen} were encoded as a percentage of the task's cycle. A new variable is specifically introduced for this configuration: θ_{CSEA} representing the position of the CSEA output which, conversely to all the other elastic actuators concepts, will be different from the biological knee joint curve θ_j since the motor is bypassed over a certain interval of time.

5.3 Parallel elastic actuators

Another widely studied family of elastic actuators is Parallel Elastic Actuators (PEA). As shown in the kinematic diagram (Figure 5.4), an elastic element is connected in parallel to the electrical motor. Previous work such as [29, 18, 17] demonstrated that a parallel spring can be used to decrease the power peak and energy consumption of the motor by mimicking the spring-like behavior observed over some phases of lower limb tasks. If well tuned, it can also have a significant positive impact on the motor peak torque since, conversely to a SEA, a PEA

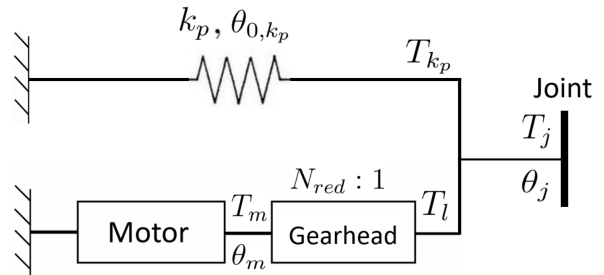


Figure 5.4: Parallel Elastic actuators (PEA) kinematic diagram

reduces the torque demand on the motor but doesn't influence its speed, allowing the motor to rotate rapidly and efficiently actuate the knee joint.

5.3.1 PEA model

To model the PEA configuration, the following equations are used:

$$\Delta\theta_{k_p} = \theta_j - \theta_{0,k_p} \quad (5.9)$$

$$\theta_m = N_{red} \cdot \theta_j \quad (5.10)$$

$$T_{k_p} = -k_p \cdot \Delta\theta_{k_p} \quad (5.11)$$

$$\omega_{k_p} = \frac{d\Delta\theta_{k_p}}{dt} \quad (5.12)$$

$$P_{k_p} = \omega_{k_p} \cdot T_{k_p} \quad (5.13)$$

$$T_l = T_j - T_{k_p} \quad (5.14)$$

Equation 5.9 and Equation 5.10 differ from the SEA configuration. Indeed, the parallel spring deformation $\Delta\theta_{k_p}$ isn't proportional to the joint torque but to its position reduced by the spring free length θ_{0,k_p} . The motor position is also directly proportional to the joint position through the reduction ratio. The parallel spring torque T_{k_p} is computed using its stiffness k_p and the spring speed ω_{k_p} and mechanical power P_{k_p} are calculated in the usual manner. Equation 5.14 expresses the reduction of the load torque T_l thanks to the parallel spring. Equation 5.6 and Equation 5.7 are again used to derive the motor speed and acceleration.

Despite its above-mentioned advantages, a classical PEA configuration presents a dilemma when it comes to the choice of parallel spring stiffness. It should be high to provide sufficient torque during the high torque-demanding phases of the tasks (typically the stance phase). On the other hand, whatever the stiffness value, the spring will deform a lot during the high-speed phases (typically the swing phase) often associated with low torque demand. If the stiffness gets too high, the spring could generate too much torque during these high-speed/low-torque phases and work against the motor.

5.3.2 CPEA model

Similarly to section 5.2, a solution to cancel this trade-off on the stiffness k_p could be to add a clutch to engage and disengage the parallel spring from the kinematic chain at strategic moments, forming a Clutchable Parallel Elastic Actuator (CPEA) configuration (kinematic diagram in Figure 5.5). Disengaging the parallel spring with the clutching element offers the opportunity to choose a very stiff spring to reduce torque, energy, and power peak demand on the motor during the stance phase while avoiding a counter-productive contribution of the spring in swing.

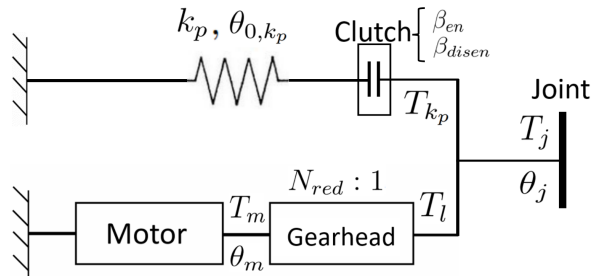


Figure 5.5: Clutchable Parallel Elastic actuators (CPEA) kinematic diagram

The mathematical model used for the CPEA configurations is identical to the one derived for the PEA, except Equation 5.9 that becomes:

$$\Delta\theta_{k_p} = \begin{cases} \theta_j - \theta_{0,k_p} & \text{if clutch engaged} \\ 0 & \text{if clutch disengaged} \end{cases} \quad (5.15)$$

For the simulation, the variables from subsection 5.2.2 β_{en} and β_{disen} are again used as the moments of engagement and disengagement of the parallel spring, still defined as a percentage of the task's cycle. As it will be later developed in this section, the two moments must correspond to close joint position values that will define the spring free length. In that way, the parallel spring is never engaged or released with a high amount of stored energy, which would have a negative impact in both cases: in the engagement situation, this would increase the actuation force needed from the engagement mechanism and in the disengagement case, the mechanical energy passively harvested in the spring would be lost.

5.3.3 2PEA model

Merging the PEA and the CPEA models (subsection 5.3.1 and subsection 5.3.2), results in a new configuration called Double Parallel Elastic Actuator (2PEA) incorporating 2 parallel springs, one of which is clutchable. The kinematic diagram can be seen in Figure 5.6 and the mathematical equations are the combination of those derived in the sections mentioned above. Adding two springs in parallel to the motor offers the advantage of selecting two different stiffness values, capturing a broader range of spring-like torque-angle relationships compared to using a single spring.

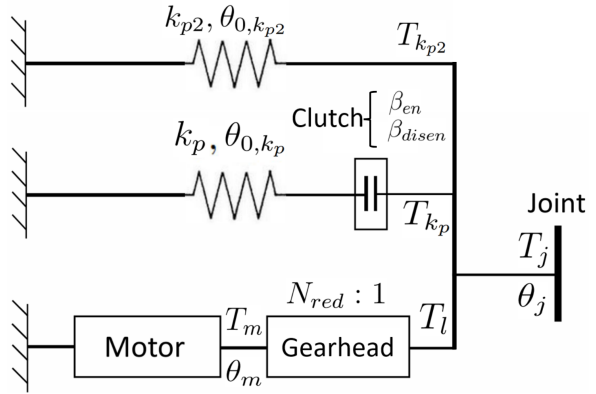


Figure 5.6: Double Parallel Elastic actuators (2PEA) kinematic diagram

5.3.4 PSEA model

Eventually, an interesting configuration is found in the merging of PEA with a SEA to form a Parallel and Series Elastic Actuator (PSEA). It has been studied in previous work, notably [21], and presents the advantages of both concepts. The mathematical model is also formulated by combining the equations derived for the PEA and SEA models. The only difference is that the series spring torque is no longer equal to the joint torque since there is now a contribution from the parallel spring:

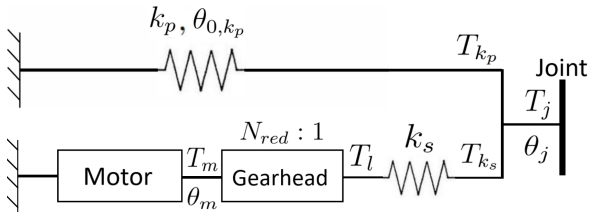


Figure 5.7: Parallel and Series Elastic actuators (PSEA) kinematic diagram

$$T_{k_s} = T_l = T_j - T_{k_p} \quad (5.16)$$

5.4 DC motor dynamic equations

The dynamic equations of a DC motor encompass both its electrical and mechanical aspects, providing a complete picture of its operation. The objective is to determine the motor's electrical power, which can then be integrated to calculate electrical consumption over one cycle.

5.4.1 Mechanical model

Neglecting viscous effects, the mechanical equation to compute the motor torque reads:

$$T_m = I \cdot a_m + \frac{T_l}{N_{red}} \cdot \eta_{mec} \quad (5.17)$$

$$\eta_{mec} = \begin{cases} \frac{1}{\eta_g} & \text{if } \frac{T_l \cdot \omega_m}{N_{red}} > 0 \\ \eta_g & \text{otherwise} \end{cases} \quad (5.18)$$

with $I = \frac{I_{chain}}{N_{red}^2} + I_m \approx I_m$ if $N_{red} \gg 1$. The moment of inertia I can effectively be approximated just by the motor's rotor inertia I_m since active prosthetic devices require a high reduction ratio, meaning that the rotational inertia of the rest of the kinematic chain I_{chain} will be negligible compared to that of the rotor, as *seen* by the motor. The only exception to Equation 5.17 is the CSEA configuration, for which $T_m = 0$ as long as the clutch is engaged and provides the reactionary torque instead of the motor that is bypassed.

As shown in previous sections, the transmission of all elastic actuator configurations is modeled by a gearhead. Since the primary concern is energy consumption, the transmission's mechanical efficiency η_{mec} needs to be considered. For that, the method described in [21] is adopted. It consists in applying linear interpolation to estimate the gearhead efficiency η_g as a function of the optimized reduction ratio N_{red} , considering a Maxon planetary gearhead [35] whose efficiency decreases quite steadily from 80% to 50% as its reduction ratio increases. As seen in Equation 5.18, this mechanical efficiency is greater than one when the power flow at the gearhead output is positive and smaller than one in the opposite situation.

5.4.2 Electrical model

For simplicity, the motor inductance is ignored as the inductance term is negligible compared to the other terms of the DC motor dynamic equation:

$$i_m = \frac{T_m}{K_m} \quad (5.19)$$

$$u_m = R_m \cdot i_m + K_m \cdot \omega_m \quad (5.20)$$

$$\Leftrightarrow u_m = T_m \cdot \frac{R_m}{K_m} + K_m \cdot \omega_m \quad (5.21)$$

where i_m , u_m , K_m , and R_m represent respectively the motor current, voltage, torque constant, and winding resistance. Based on the torque and speed, the motor mechanical power P_{mech} is computed and its electrical power consumption P_e is derived using the

current and voltage. For the same reason as above, the electrical efficiency η_e is taken into account through the motor maximal efficiency η_{me} . The Joule's power losses P_J can also be estimated as proportional to the square of the current:

$$P_{mech} = T_m \cdot \omega_m \quad (5.22)$$

$$P_e = u_m \cdot i_m \cdot \eta_e \quad (5.23)$$

$$\eta_e = \begin{cases} \frac{1}{\eta_{me}} & \text{if } u_m \cdot i_m > 0 \\ \eta_{me} & \text{otherwise} \end{cases} \quad (5.24)$$

$$P_J = R_m \cdot i_m^2 \quad (5.25)$$

Finally, assuming the negative power flow in the motor is dissipated as heat and not recovered, the electrical energy consumption E_e (positive energy flow) is computed as:

$$E_e = \int P_{e,pos} dt \quad (5.26)$$

$$\text{with } P_{e,pos} = \begin{cases} P_e & \text{if } P_e > 0 \\ 0 & \text{otherwise} \end{cases} \quad (5.27)$$

5.4.3 Thermal model

The thermal model is based on guidelines provided by [37]. Any DC motor is characterized by two zones in a speed-torque diagram: a continuous operation zone and an intermittent (or short-term) operation zone. The continuous operating range is generally limited by the maximum allowed speed $w_{m,max}$ and the maximum continuous (or nominal) torque T_N . Operating the motor at speeds and torques within this area will not lead to overheating, ensuring a reasonable motor life.

On the other hand, a DC motor can be highly overloaded, i.e., operated at a torque much higher than the maximum continuous torque. All speed-torque combinations with torque higher than the nominal value constitute the intermittent operating zone, where the motor may be run repeatedly but only for short durations. The permissible level of overload for a motor depends on the heating of its thermally weakest component, the winding. The winding temperature must stay below its maximum allowable limit, noted $T_{w,max}$. The heating process of the winding follows an exponential function (see Figure 5.8a), characterized by the winding's thermal time constant τ_w ³ until it reaches a thermal equilibrium, after several time constants.

As illustrated in Figure 5.8a, the temperature difference between the winding and ambience $\Delta T_{w,\infty}$, and between the housing (or stator) and ambience $\Delta T_{s,\infty}$ are defined as:

$$\Delta T_{w,\infty} = T_{w,\infty} - T_a \quad (5.28)$$

$$\Delta T_{s,\infty} = T_{s,\infty} - T_a \quad (5.29)$$

with T_a the ambient temperature, $T_{w,\infty}$ and $T_{s,\infty}$ the steady-state temperatures of the winding and housing, reached at thermal equilibrium. Furthermore, the temperature

³This intrinsic constant is defined as the time it takes for the winding to reach 63% of the final steady-state temperature. This definition is based on mathematical reasons and engineering conventions.

difference winding-ambient can be estimated thanks to the thermal resistances between winding and housing R_{th1} and between housing and ambient R_{th2} , and the knowledge of the Joule's power losses:

$$\Delta T_{w,\infty} = (R_{th1} + R_{th2}) \cdot P_J \quad (5.30)$$

The thermal resistances describe how well the excess heat generated inside the motor by Joule's effect can flow from the winding to the housing, and from the housing to the ambient. Their numerical values can be found in a motor datasheet. It's crucial to denote that, if R_{th1} is intrinsic to the motor design, R_{th2} can be significantly reduced compared to its attributed value. This can be achieved for example by forced air convection or by mounting the motor on a metallic frame acting as a heat sink [32, 61].

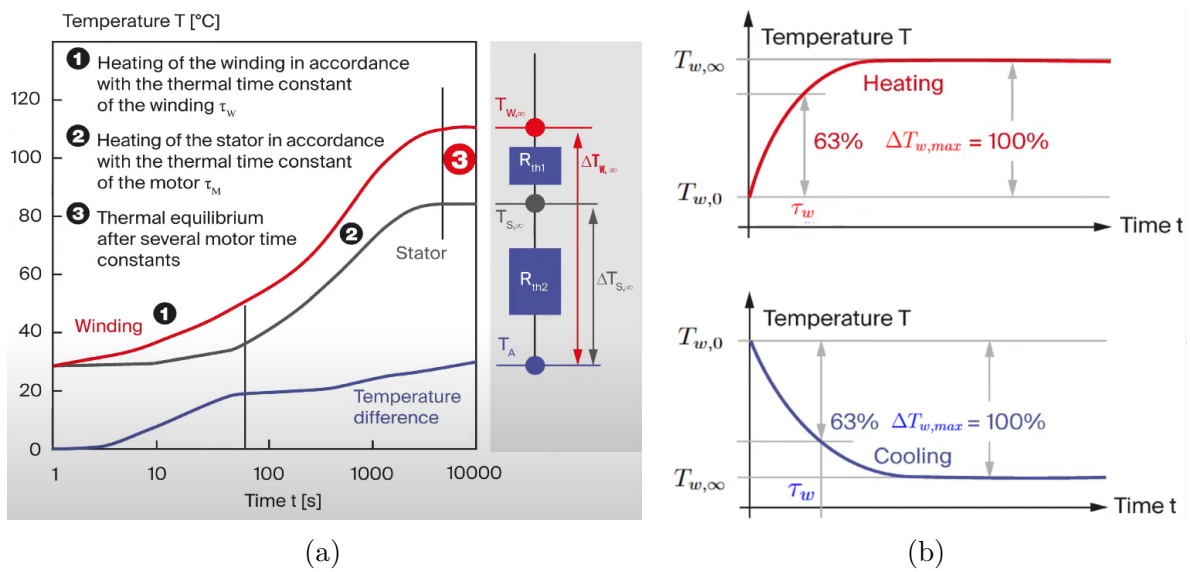


Figure 5.8: Adapted from [37]. (a) Winding and stator temperature evolution in a DC motor (b) Heating and cooling of the winding

For tasks requiring a high torque profile, such as stairs ascent, the motor will likely be operated for a certain amount of time in the intermittent zone. The question is: can the motor perform indefinitely without overheating? If not, how many stairs ascent cycles can it perform safely?

To answer this question, a simplified model assuming a uniform heat flux for the winding is used. Its graphic representation is illustrated in Figure 5.8b. The equation associated with this model is:

$$\Delta T_w(t) = T_w(t) - T_{w,0} = \Delta T_{w,max} \cdot \left(1 - e^{-\frac{t}{\tau_w}}\right) \quad (5.31)$$

$$\text{where } \Delta T_{w,max} = T_{w,\infty} - T_{w,0} \quad (5.32)$$

with T_w the current winding temperature and $T_{w,0}$ the starting winding temperature depending on the thermal history of the motor. $\Delta T_{w,max}$ is the maximum temperature difference between the steady-state winding temperature (reached at thermal equilibrium)

and the starting winding temperature. If the steady-state temperature is superior to the starting one ($\Delta T_{w,max} > 0$), the winding is heating and it is cooling the other way around ($\Delta T_{w,max} < 0$).

Within a stairs ascent cycle, there is a short but significant overload period at the beginning of the stance phase and for the rest of the cycle, the motor can typically be operated in the continuous zone, given the small motors considered in this work whose respective specifications can be found later in subsection 5.5.1.

Furthermore, Joule's power losses fluctuate continuously within the cycle, so in Equation 5.25, the motor current i_m is replaced by its RMS value $i_{m,RMS}$ (Equation 5.38) over the whole cycle to obtain the RMS value of Joule's losses over a stairs ascent cycle. This allows the use of Equation 5.30 to compute the temperature difference between winding and ambient when the thermal equilibrium is reached. From there, $\Delta T_{w,max}$ can be computed:

$$\Delta T_{w,max} = \begin{cases} \Delta T_{w,\infty} & \text{if } T_{w,0} = T_a \\ \Delta T_{w,\infty} - (T_{w,0} - T_a) & \text{otherwise} \end{cases} \quad (5.33)$$

Knowing $\Delta T_{w,max}$ (> 0 for an active task like stairs ascent), the time evolution of the winding temperature can be derived through Equation 5.31. If the winding temperature T_w can not reach its steady-state value at thermal equilibrium $T_{w,\infty}$ without surpassing the maximum allowed temperature $T_{w,max}$, the number of stairs ascent cycles that could be performed is limited. Otherwise, the motor can theoretically perform an unlimited number of cycles without risking overheating.

5.5 Optimization procedure

Figure 5.9 displays a block diagram of the optimization procedure. It can be broken down into three main steps.

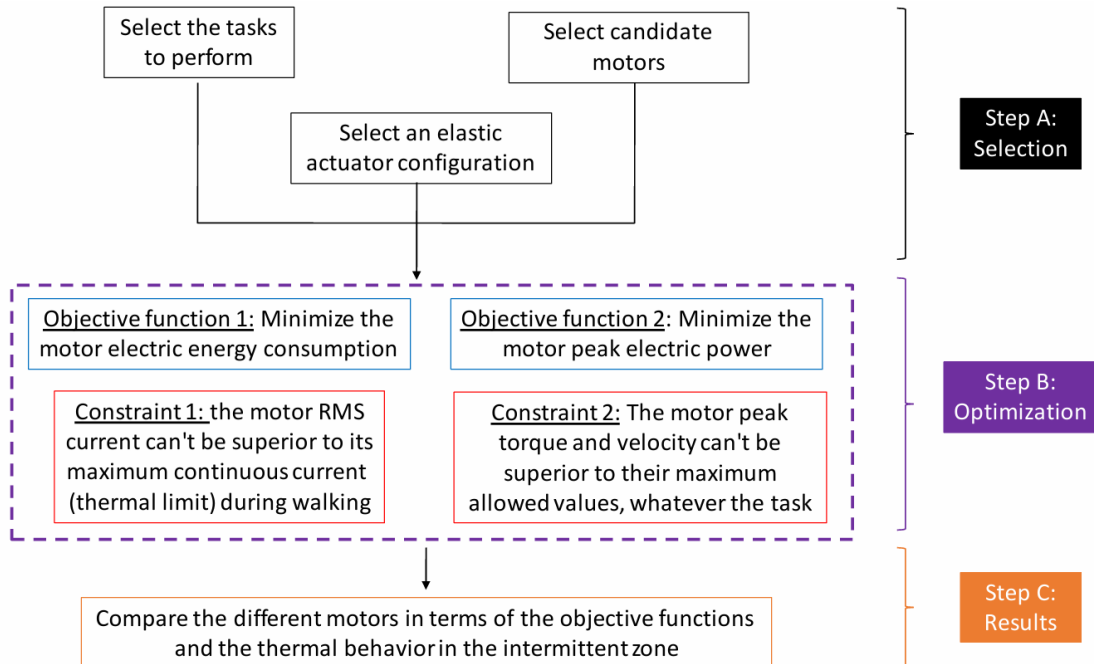


Figure 5.9: Block diagram of the optimization procedure

5.5.1 Step A: Selection

Three main decisions have to be made:

- **Tasks to perform:** As previously discussed, the tasks can either involve walking alone or walking combined with stairs ascent.
- **Elastic actuator configuration:** it can be one of the six configurations described at the beginning of this chapter
- **Candidate motors:** Inspired by [21], 3 different small EC motors⁴ are evaluated: the Maxon EC30 4 poles [34], the Maxon EC22 4 poles [33], and the Faulhaber BP3274 4 poles [13]. The main intrinsic parameters of these motors are detailed in Table 5.1.

Motor	EC30	EC22	BP3274
Rated power P_N [W]	200	120	152
Nominal torque T_N [mNm]	94.6	54.6	158
Nominal current i_N [A]	7.61	4.21	6.88
Torque constant K_m [mNm/A]	13.7	13.5	28.1
Winding resistance R_m [Ω]	0.102	0.341	0.253
Rotor inertia I_m [g cm²]	33.3	8.91	48
Maximal efficiency η_{me} [-]	0.9	0.89	0.88
Stall torque $T_{m,p}$ [mNm]	3240	954	2650
Max. speed $\omega_{m,max}$ [rpm]	25000	25000	16000
Winding max. T° $T_{w,max}$ [°C]	155	155	150
Thermal resistance			
– winding-housing R_{th1} [K/W]	0.21	0.7	1.1
– housing-ambient R_{th2} [K/W]	7.4	10.7	7.9
Thermal time winding τ_w [s]	2.11	4.66	22
Mass M_m [g]	300	175	325

Table 5.1: Main parameters of the evaluated motors. Values are given for an ambient temperature of 22 [°C] and at a nominal voltage of 24 [V]

5.5.2 Step B: Optimization

Optimization was performed using the versatile algorithm Optimeed developed by C. De Gréef, a research assistant from UCLouvain. The program takes some objective functions and constraints as well as optimization variables with a defined range and returns the optimized parameters that minimize the objective functions while satisfying all the constraints.

Objective functions

The objective functions to minimize are defined as follows:

⁴EC" motor stands for "Electronically Commutated" motors, equivalent to "Brushless DC" motor

1. **The motor peak electrical power (in absolute value)**

$$F_1 = \max(|P_e|) \quad (5.34)$$

Minimizing F_1 aims to flatten the motor power curve since the power peak of an electric motor is one of the primary factors influencing its size [10].

2. **The motor electrical energy consumption over one task's cycle**

$$F_2 = E_e \quad (5.35)$$

The electrical energy consumption is obtained via Equation 5.26 and Equation 5.27 and represents the integral of the positive part of the electrical power over the period of one cycle. One could have chosen to integrate the absolute electrical power to minimize the positive values (when the motor is consuming power) as well as the negative ones (when the motor is absorbing power). Indeed, in the absence of regeneration or energy transfer strategies, the negative power flow to the motor has to be dissipated as heat in the circuit. If the power to be absorbed gets high, this could lead the motor to overheat. However, as explained below in the *constraints* paragraph, particular attention is paid to the motor's thermal behavior to ensure that it never exceeds its limits, allowing to focus only on the positive power flow.

3. **The scalar kinematic agreement between the CSEA output and joint positions (specific to the CSEA configuration)**

$$F_3 = \int (\theta_{CSEA} - \theta_j)^2 dt \quad (5.36)$$

This third objective function only pertinent to the CSEA configuration is taken from [55]. Indeed, for this concept, the evolution of the output position (θ_{CSEA}) will differ from the biological knee joint curve (θ_j) since the motor is bypassed over a certain interval of time. This difference between both should be minimized to replicate the motion of a sound leg as faithfully as possible.

F_1 and F_2 are the main objective functions included for every optimization. F_3 is only involved in the optimization of the CSEA configuration. When optimizing for both tasks, the walking objective functions are given a weight ten times greater than those for stairs ascent. This weighting reflects the relative importance of the tasks, as walking is performed more frequently than stairs ascent (the specifications foresee 1500 walking steps and 100 staircase steps per day).

Constraints

The following feasibility constraints need to be introduced to ensure that the RMS current, maximum speed, and maximum torque remain within the motor's specifications [54].

1. **Winding thermal constraint**

$$i_{m,RMS} \leq i_N \quad (5.37)$$

$$\text{where } i_{m,RMS} = \sqrt{\frac{1}{\Delta t} \int_0^{\Delta t} i_m^2 dt} \quad (5.38)$$

The winding is the most critical part when looking at the thermal limit of a DC motor. The winding heats up as electrical current passes through its resistance, generating heat according to Joule's formula (Equation 5.25). Keeping the motor root-mean square (RMS) current $i_{m,RMS}$ inferior to its maximum continuous current i_N (also known as the nominal current) prevents the winding from overheating, even when the thermal equilibrium is reached during sustained operation. For the motors evaluated here, this constraint was applicable for simulating walking but not for stairs ascent, which is an active task for the knee requiring high motor torque and, consequently, a high RMS current. However, as stairs ascent may be performed for a limited number of cycles, it calls for further investigation of the thermal behavior of DC motors operating in their intermittent zone, as discussed later in this chapter.

2. Peak speed constraint

$$\max(|\omega_m|) \leq \omega_{m,max} \quad (5.39)$$

The maximum allowed speed $\omega_{m,max}$ serves as a guideline rather than an absolute limit and is fixed by mechanical constraints notably related to bearings life. Approaching or surpassing this speed will gradually result in increased audible noise and reduced motor life [32]. This constraint is applicable whatever the task.

3. Peak torque constraint

$$\max(|T_m|) \leq T_{m,p} \quad (5.40)$$

Whatever the task, the motor peak torque can never be superior to its peak torque rating $T_{m,p}$ (also known as the stall torque) since this could cause demagnetization of the permanent magnets [54].

4. Engagement and disengagement timing constraint

$$|\theta_j(t_{\beta_{en}}) - \theta_j(t_{\beta_{disen}})| \leq 1^\circ \quad (5.41)$$

As already discussed in subsection 5.3.2, the engagement and disengagement instants, $t_{\beta_{en}}$ and $t_{\beta_{disen}}$ must correspond within the cycle to similar joint position values. These *time* variables should not be confused with β_{en} and β_{disen} , the moments of (dis)engagement as *percentages* of the gait cycle. In the program, the difference limit between both was arbitrarily set to a low value of 1° . In the case of a CPEA or 2PEA, this avoids losing energy stored in the parallel spring during disengagement and minimizes the actuating force required during re-engagement. For the CSEA, this avoids unwanted instantaneous motor speed peaks during disengagement of the clutch (i.e. when the motor is "switched back on").

Optimization Variables

Table 5.2 presents the optimization variables assigned to each configuration, along with their corresponding ranges within which the algorithm will search for the optimized value.

Some choices warrant particular attention. First, one can see that the CPEA and 2PEA do not include the free length of the (dis)engageable parallel spring (θ_{0,k_p}) as an optimization variable. This is directly linked to the 4th constraint: the free length must be equal to the joint angle at the engagement instant $\theta_j(t_{\beta_{en}})$ or equivalently - since both values must be close - at the disengagement instant $\theta_j(t_{\beta_{disen}})$.

Additionally, for the combined optimization of walking and stairs ascent for these same two configurations, ranges were provided for the moments of engagement β_{en} and disengagement β_{disen} during walking, but not during stairs ascent. This is because the algorithm consistently converges to a value of around 10° for the joint position at (dis)engagement, which is optimal for a walking cycle. For a stairs ascent cycle, this approximates the minimal joint position that is reached only once during the forward continuance phase (see Figure 3.13a). Consequently, it is not feasible to implement a useful CPEA mechanism for stairs ascent considering that the fourth constraint must be satisfied for all tasks taken into account, thus it was simulated as a PEA.

	SEA	CSEA	PEA	CPEA	2PEA	PSEA
k_s [Nm/rad]	[50, 450]	[50, 450]				[50, 450]
k_p [Nm/rad]			[5, 50]	[5, 350]	[5, 350]	[5, 50]
k_{p2} [Nm/rad]					[5, 50]	
θ_{0,k_p} [°]			[0, 60]	-	-	[0, 60]
$\theta_{0,k_{p2}}$ [°]					[0, 60]	
W β_{en} [%]		[0, 100]		[0, 100]	[0, 100]	
W β_{disen} [%]		[0, 100]		[0, 100]	[0, 100]	
SA β_{en} [%]		[0, 100]		-	-	
SA β_{disen} [%]		[0, 100]		-	-	
N_{red} [-]	[100, 400]	[100, 400]	[100, 400]	[100, 400]	[100, 400]	[100, 400]

Table 5.2: Optimization variables ranges (W: "Walking" & SA: "Stairs Ascent")

5.5.3 Step C: Results

The following paragraphs aim to present the various results and graphs generated by the optimization program, given the example of a SEA configuration and the candidate motors described in Table 5.1. The next section will focus on comparing these results between the different elastic actuator configurations.

Walking optimization

1. Optimization results

Table 5.3 displays the optimized parameters of a SEA for each evaluated motor, as well as the value of the minimized objective functions along with their percentage reduction compared to a direct drive configuration (optimized reduction ratio but no series spring) equipped with the same motor.

	EC30	EC22	BP3274
k_s [Nm/rad]	265	265	290
N_{red} [-]	231	375	145
$F_1 = \max(P_e)$ [W]	101.8 (reduction by 27%)	81.5 (reduction by 33%)	80.1 (reduction by 23%)
$F_2 = E_e$ [J]	10.1 (reduction by 52%)	10.2 (reduction by 48%)	9.2 (reduction by 47%)

Table 5.3: Optimization results for a SEA configuration with the different motors

2. Position, torque, and power profiles (walking)

The graphs reported below in Figure 5.10 are for a SEA configuration embedding a Faulhaber BP2374 motor. This choice is arbitrary since the profiles associated with the other two motors are very similar.

Figure 5.10a and Figure 5.10b highlight the behavior of the series spring along a walking cycle. Its optimized stiffness is such that it can render the quasi-linear torque-angle relationship exhibited during the ESF phase. During this phase, the motor speed is reduced compared to a direct drive configuration, as observed in Figure 5.10c. The spring acts like a mechanical energy buffer that allows to flatten the motor mechanical power profile (green curve in Figure 5.10d). However, the high reactionary torque required from the motor during ESF induces high Joule's power losses (yellow curve), thus the motor electrical power (blue curve) isn't reduced as much as the motor mechanical power.

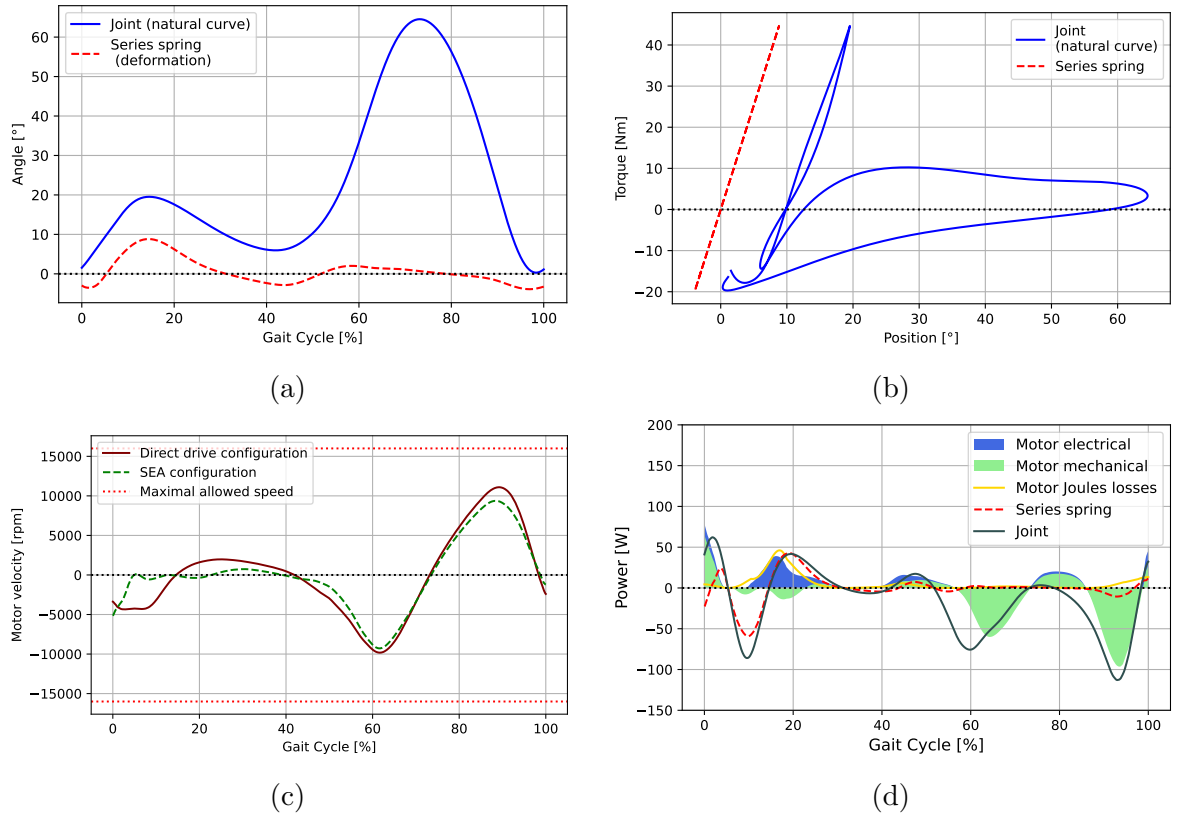


Figure 5.10: Profiles for a SEA configuration embedding a Faulhaber BP3274 motor (walking) (a) position (b) torque-angle (c) motor speed (d) powers

3. Motor current and voltage profiles (walking)

Figure 5.11a gives the visual confirmation that the 1st constraint on the winding temperature is satisfied as the motor RMS current is inferior by 32% to its maximum continuous (or nominal) current.

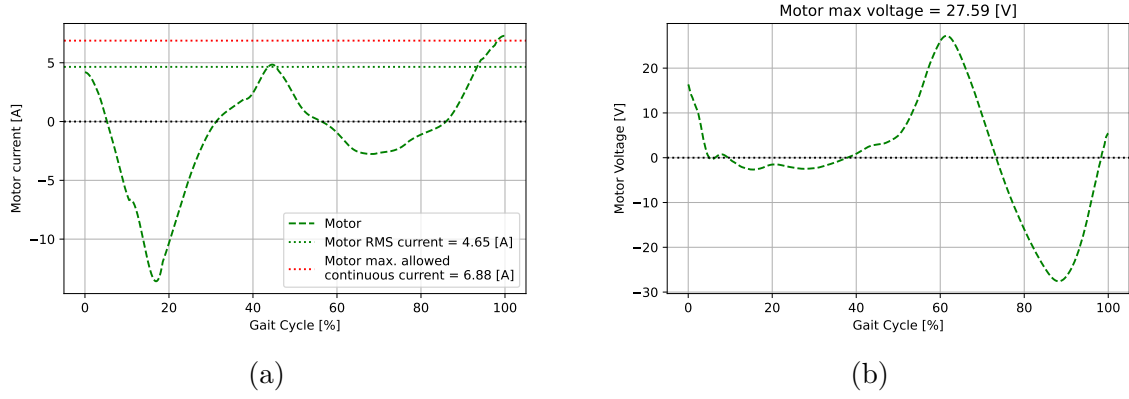


Figure 5.11: Profiles for a SEA configuration embedding a Faulhaber BP3274 motor (walking) (a) motor current (with RMS value indicated) (b) motor voltage

4. Motors speed-torque operating range (walking)

It can be observed in Figure 5.12 that for each motor, the speed and torque mainly stay in the continuous operating range extracted from the datasheets. The motor still spends a short amount of time in the intermittent zone when it has to provide the peak reactionary torque during ESF, but the overload isn't thermally critical since the 1st constraint on the RMS current is respected for all 3 motors.

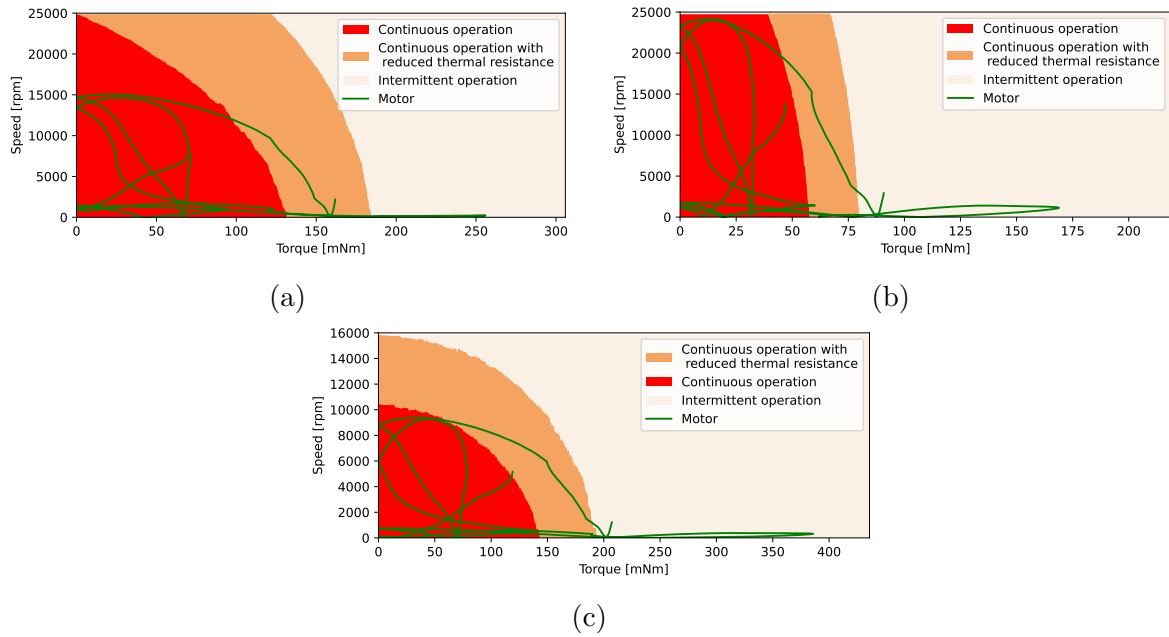


Figure 5.12: Speed-torque operating range for a SEA configuration embedding different motors (walking) (a) Maxon EC30 (b) Maxon EC22 (c) Faulhaber BP3274

Combined walking and stairs ascent optimization

1. Optimization results

Table 5.4 gives the optimized parameters of a SEA for each evaluated motor, together with the value of the minimized objective functions for both walking and stairs ascent.

Once again, the percentage gain compared to a direct drive configuration equipped with the same motor is indicated.

	EC30	EC22	BP3274
k_s [Nm/rad]	345	445	320
N_{red} [-]	230	378	180
W $F_1 = \max(P_e)$ [W]	107.1 (reduction by 23%)	90.6 (reduction by 25%)	98.9 (reduction by 5%)
W $F_2 = E_e$ [J]	12.0 (reduction by 43%)	13.8 (reduction by 30%)	10.2 (reduction by 41%)
SA $F_1 = \max(P_e)$ [W]	531.2 (no reduction)	635.9 (reduction by 9%)	518.3 (reduction by 8%)
SA $F_2 = E_e$ [J]	132.3 (reduction by 10%)	159.2 (reduction by 12%)	126.8 (reduction by 15%)

Table 5.4: Optimization results for a SEA configuration embedding the different candidate motors (W: "Walking" and SA: "Stairs Ascent")

Compared to the results obtained when the optimization is only focused on minimizing F_1 and F_2 during walking (see Table 5.3), both the electrical peak power and energy consumption have increased for each motor. This highlights a significant drawback inherent to elastic actuators: the parameters, particularly the spring(s) stiffness, are only optimal for a specific task and cadence, as already noted in [10]. When combining several tasks to optimize the parameters of an elastic actuator, trade-offs are necessary, preventing the power peak and energy consumption from reaching the theoretical minimums achievable for each task taken individually.

2. Position, torque, and power profiles (stairs ascent)

The graphs corresponding to a walking cycle are not displayed again since, despite the differences in the optimized parameters, the profiles remain analogous to those shown in Figure 5.10. Instead, Figure 5.13 depicts the profiles corresponding to a stairs ascent cycle for a SEA configuration incorporating a Faulhaber BP2374 motor. This choice remains arbitrary as the profiles associated with the other two motors still exhibit similar characteristics.

Figure 5.13b indicates that the optimized stiffness of the series spring is not perfectly suited for the stance phase of the stairs ascent cycle. Furthermore, as visible in Figure 5.13d, the joint mechanical power exhibits a very high positive peak at the beginning of the cycle that the series spring cannot effectively mitigate. This is because the spring must first store mechanical energy before it can release it, yet there are no phases of high dissipation during which the spring could store energy without increasing the motor mechanical power. Eventually, the high torque required during the stance phase to stand up and climb the step (see Figure 5.13c generates substantial Joule's power losses, significantly increasing the electrical power compared to the mechanical power.

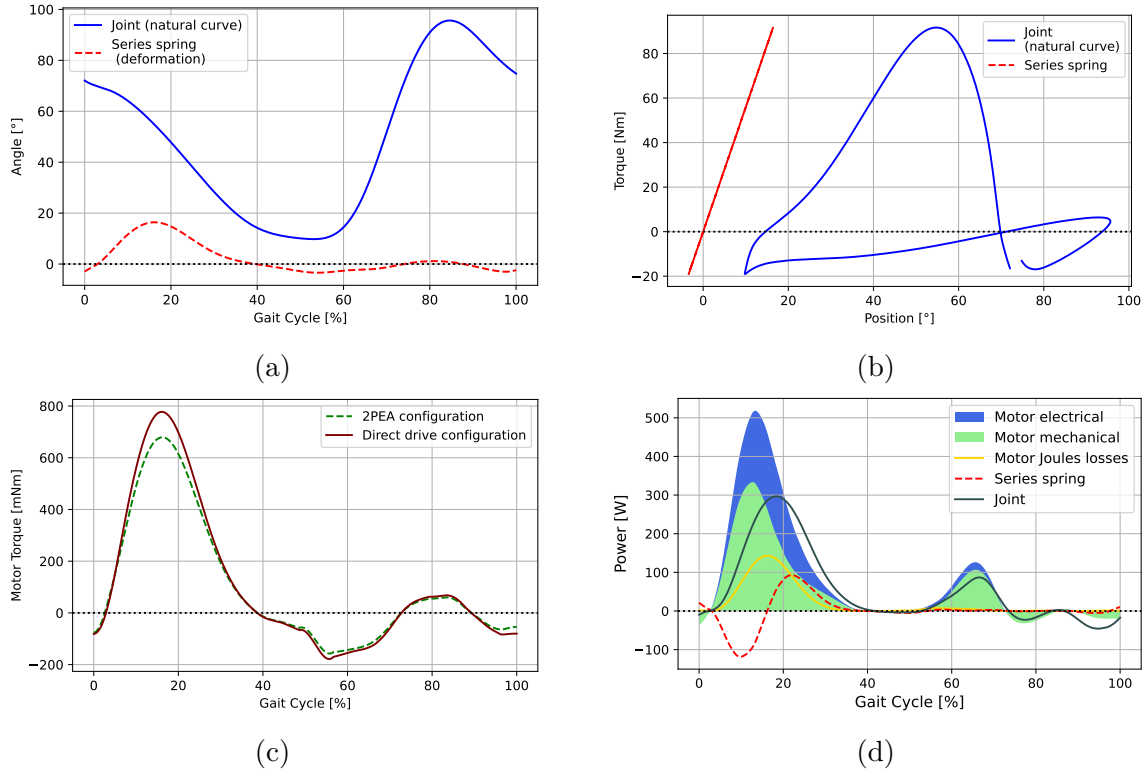


Figure 5.13: Profiles for a SEA configuration embedding a Faulhaber BP3274 motor (stairs ascent) (a) position (b) torque-angle (c) motor torque (d) powers

3. Motor current and voltage profiles (stairs ascent)

Again, the graphs below are for a SEA configuration incorporating a Faulhaber BP2374 motor.

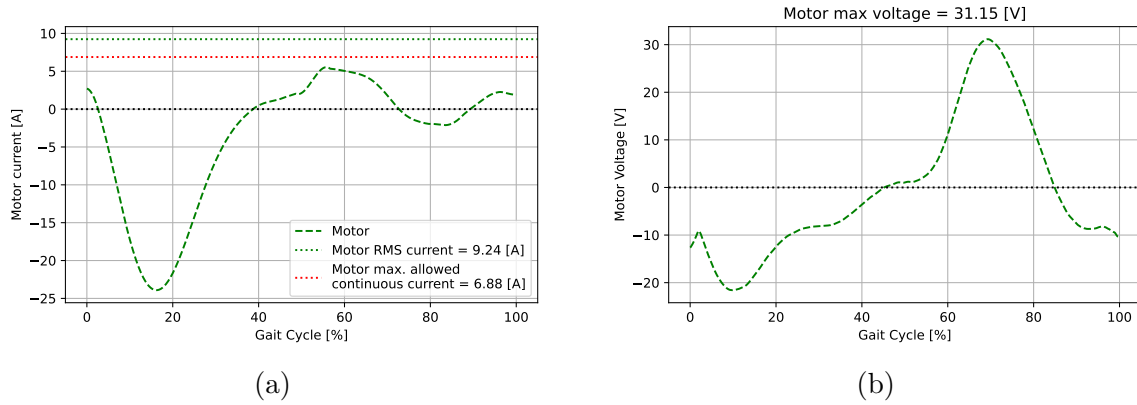


Figure 5.14: Profiles for a SEA configuration embedding a Faulhaber BP3274 motor (stairs ascent) (a) motor current (with RMS value indicated) (b) motor voltage

Unlike a walking cycle, a stairs ascent cycle requires high torque and thus high current, leading to an RMS current value that exceeds the maximum continuous (or nominal) current (see Figure 5.14a). This could pose a thermal issue for the motor, necessitating a check of the temperature evolution of the winding (the thermally weakest part of the motor).

4. Motors speed-torque operating range (stairs ascent)

As expected, Figure 5.15 shows that for each motor, even though the speed and torque primarily remain within the continuous operating range, there is a significant intrusion into the intermittent zone due to the high torque required during the weight acceptance phase. Such an overload could lead to winding overheating; hence, as mentioned in the previous paragraph, the thermal behavior needs to be studied.

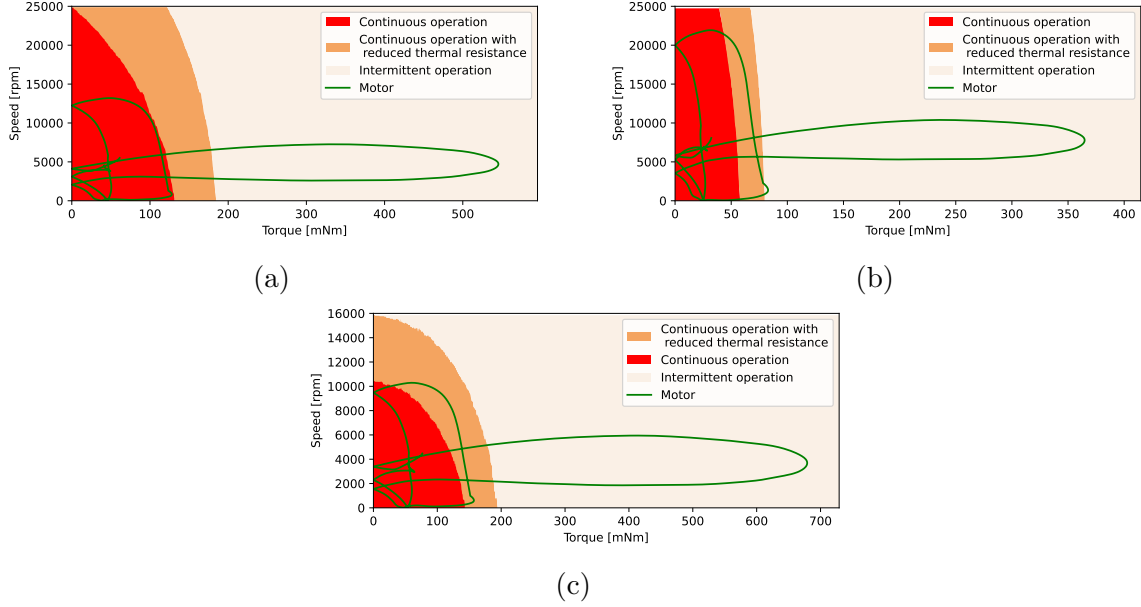


Figure 5.15: Speed-torque operating range for a SEA configuration embedding different motors (stairs ascent) (a) Maxon EC30 (b) Maxon EC22 (c) Faulhaber BP3274

5. Winding thermal behavior (stairs ascent)

To analyze the winding thermal behavior, the model described in subsection 5.4.3 is employed, with the starting winding temperature $T_{w,0}$ set as the steady-state temperature reached during sustained walking activity. This temperature is obtained by computing Joule's losses associated with a walking cycle operated at RMS current, and ranges for the SEA between 40 and 90 [°C], depending on the selected motor. The idea is that a user has probably walked over a certain period before climbing stairs, therefore the winding isn't at ambient temperature.

Figure 5.16 emphasizes what was stated previously: During a stairs ascent cycle, there is a brief but high overload period at the beginning of the stance phase, after which the motor can generally operate within the continuous zone for the remainder of the cycle. This overload induces a RMS torque that is greater than the motor maximum continuous (or nominal) torque.

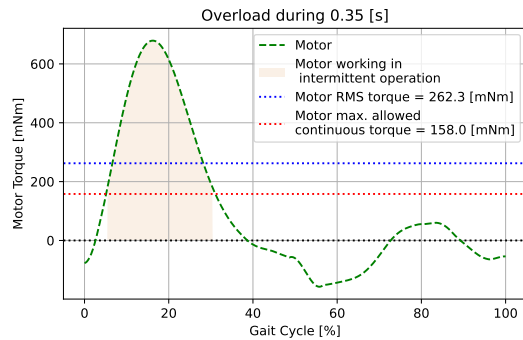


Figure 5.16: Motor torque profile highlighting the overload period for a SEA configuration embedding a Faulhaber BP3274 motor (stairs ascent)

Applying the thermal model to the three candidate motors yields the results shown in Figure 5.17. The model is applied twice: first using the housing-ambient thermal resistance value (R_{th2}) provided in the motor datasheet, and then using half of this value ($R_{th2} - 50\%$). The second simulation, which assumes the capacity to dissipate excess heat from housing to ambient is twice more effective, demonstrates significantly better thermal performance.

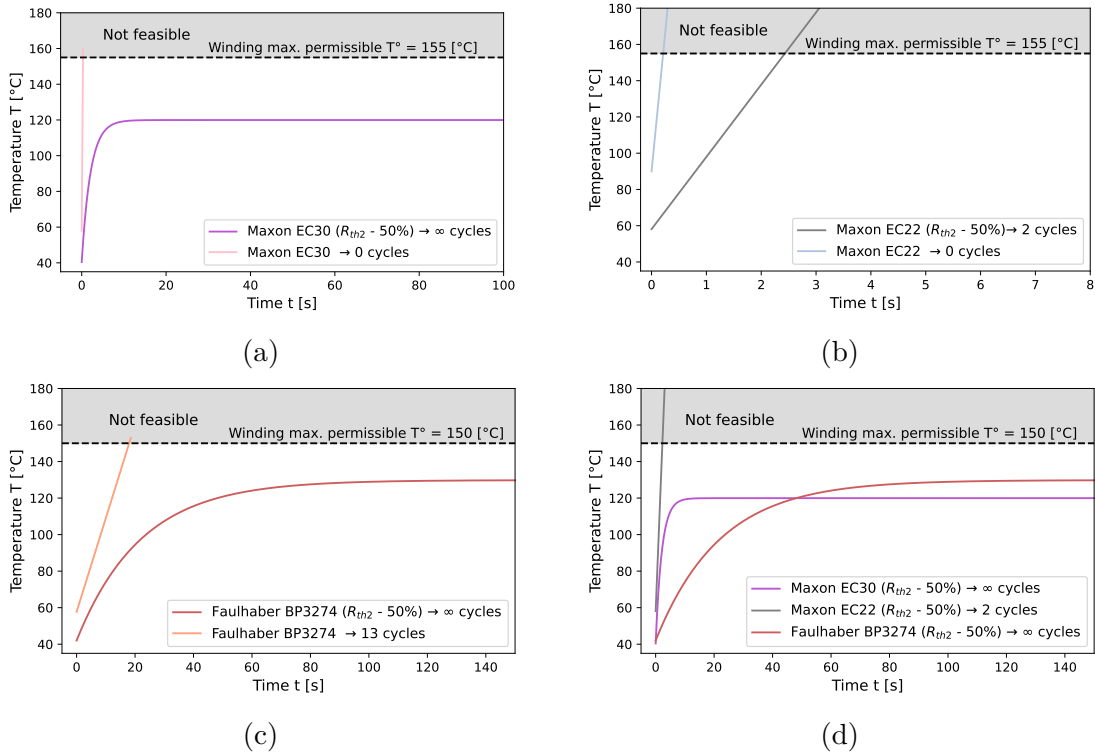


Figure 5.17: Winding temperature evolution during stairs ascent for a SEA configuration
(a) Maxon EC30 (b) Maxon EC22 (c) Faulhaber BP3274 (d) comparison

As indicated in the legend of the different graphs (Figure 5.17a, Figure 5.17b, Figure 5.17c), the thermal performance during stairs ascent for each motor is:

- Maxon EC30:
 - Unable to perform any cycle
 - if ($R_{th2} - 50\%$): able to perform indefinitely without overheating
- Maxon EC22:
 - Unable to perform any cycle
 - if ($R_{th2} - 50\%$): able to perform 2 cycles before overheating
- Faulhaber BP3274:
 - able to perform 13 cycles before overheating
 - if ($R_{th2} - 50\%$): able to perform indefinitely without overheating

Figure 5.17d displays a comparison between the three motors with reduced thermal resistance. Note that a single winding max. permissible temperature of 150 [°C] is represented despite the slight difference between the motors (150 [°C] for the Maxon

motors but 155 [°C] for the Faulhaber motor). The Maxon EC22 is unable to perform sustained stairs ascent, even with the reduced thermal resistance. This limitation holds for every elastic actuator configuration, thus the Maxon EC22 must be rejected from a kinematic chain including only an elastic actuator, as will be discussed further in the next section. Conversely, the other two motors can theoretically perform an indefinite number of cycles if their thermal resistance is reduced by 50%. This reduction can be approached by mounting the motor on a metallic frame, as done in [61], for example.

5.6 Comparison and discussion of the results

This section will be divided into two parts: a comparison of the results of an optimization focused exclusively on walking, and those of an optimization that includes both walking and stairs ascent. The results compared for all possible combinations of motors and elastic actuator configurations are the minimized objective functions, namely the electrical energy consumed by the motor over one cycle and the motor peak electrical power.

Additionally, the elastic actuator configurations are compared to a direct drive configuration, which has an optimized reduction ratio but does not include any springs. The goal is to highlight the gain that can be made by incorporating elastic elements in the kinematic chain.

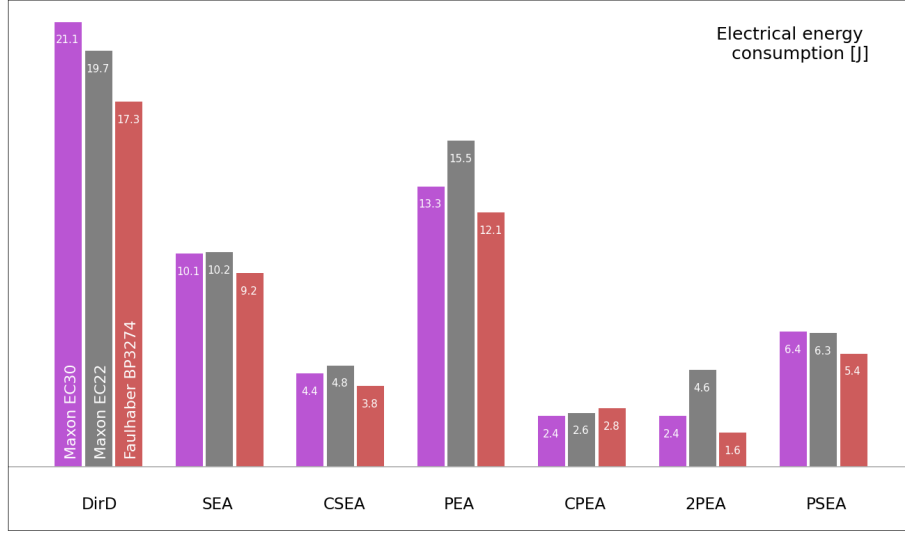
For the optimization including stairs ascent, the thermal performance will also be evaluated to reject the combinations of motors and elastic actuators that can't perform sustained stairs ascent without overheating.

It could be argued that configurations with more variables (several springs, clutch, or both) are more likely to perform better than their simpler counterparts and that the decreased energy consumption or improved thermal performance could be offset by the increased weight and complexity. Nevertheless, in the present and next chapters, the choice is made to compare the configurations based exclusively on the objective functions, as energetic efficiency is the primary design objective. Specifically, walking energy consumption is used to identify which kinematic chains offer the highest prosthesis autonomy, while less importance is granted to the peak electrical power since it allows for the selection of smaller motors, however, the candidate motors have already been selected. Chapter 7, which proposes a global comparison of both categories of kinematic chains investigated in this work, will take into account the weight and implementation complexity.

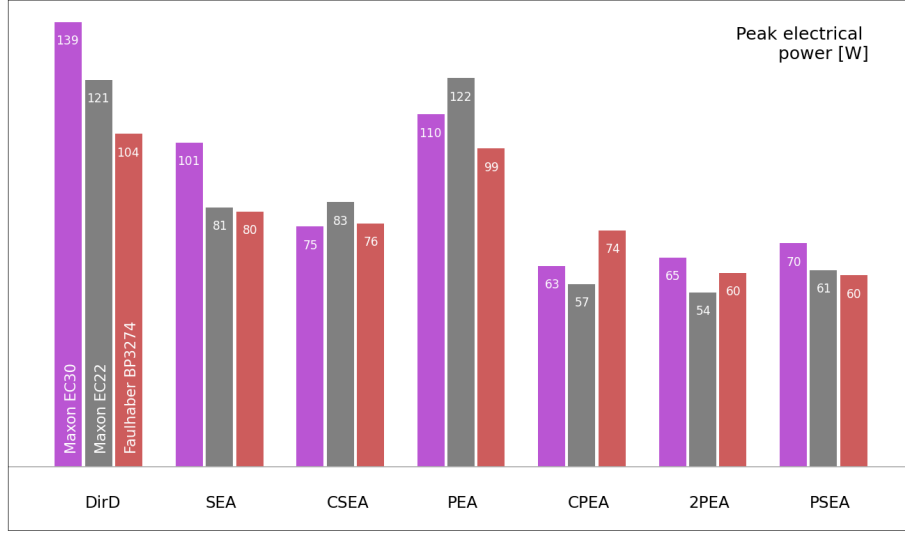
5.6.1 Walking optimization

As displayed in Figure 5.18, the best walking performance is provided by elastic actuators that include a (dis)engageable parallel spring (CPEA and 2PEA). Figure 5.18a) shows that the lowest electrical energy consumption during walking is achieved by **a double parallel elastic actuator (2PEA) equipped with a Faulhaber BP3274 motor**.

The energy consumption over one walking cycle for this combination is estimated to be around 1.6 [J], which is one order of magnitude smaller (reduction by 90.8%) than the estimated 17.3 [J] required in direct drive. The peak electrical power estimate is 60 [W], representing a reduction of 42.3% compared to the estimation of 104 [W] in direct drive.



(a)



(b)

Figure 5.18: Comparison of walking performance between direct drive and all six elastic actuators configurations optimized for walking only (a) Electrical energy consumed by the motor over one walking cycle (b) Peak motor electrical power during walking

Figure 5.19 gives a meaningful insight of the advantages inherent to the 2PEA configuration. The (dis)engageable parallel spring (yellow dotted curve) has a high stiffness corresponding to the linear torque-angle relationship characteristic of the walking ESF (see Figure 5.19b). Table 5.5 gives the numerical value of this stiffness as well as the moments of engagement and disengagement of the spring, which as previously discussed are constrained to correspond to a close joint angle, here around 9.4° .

As visible in Figure 5.19a, the high-stiffness parallel spring is engaged just before heel strike, reduces significantly the motor torque compared to a direct drive configuration during ESF (see Figure 5.19c), and eventually is disengaged during late stance to prevent it from working against the motor during the high-speed swing phase.

Figure 5.19f confirms that the high-stiffness spring captures all power fluctuations during ESF, bringing the motor's electrical energy to 1.6 [J].

The second spring which is not (dis)engageable (purple dotted curves) has a low

stiffness adapted to the flexion torque required at late swing to prevent the knee from hyperextension. Again, it allows the motor to provide less torque, therefore lowering the Joule's power losses.

Lastly, Figure 5.19d displays the speed-torque range within which the motor is operating, validating that it stays in the continuous zone where it won't risk overheating. Indeed, as seen in Figure 5.19e, the RMS current is inferior by 72% to the nominal current.

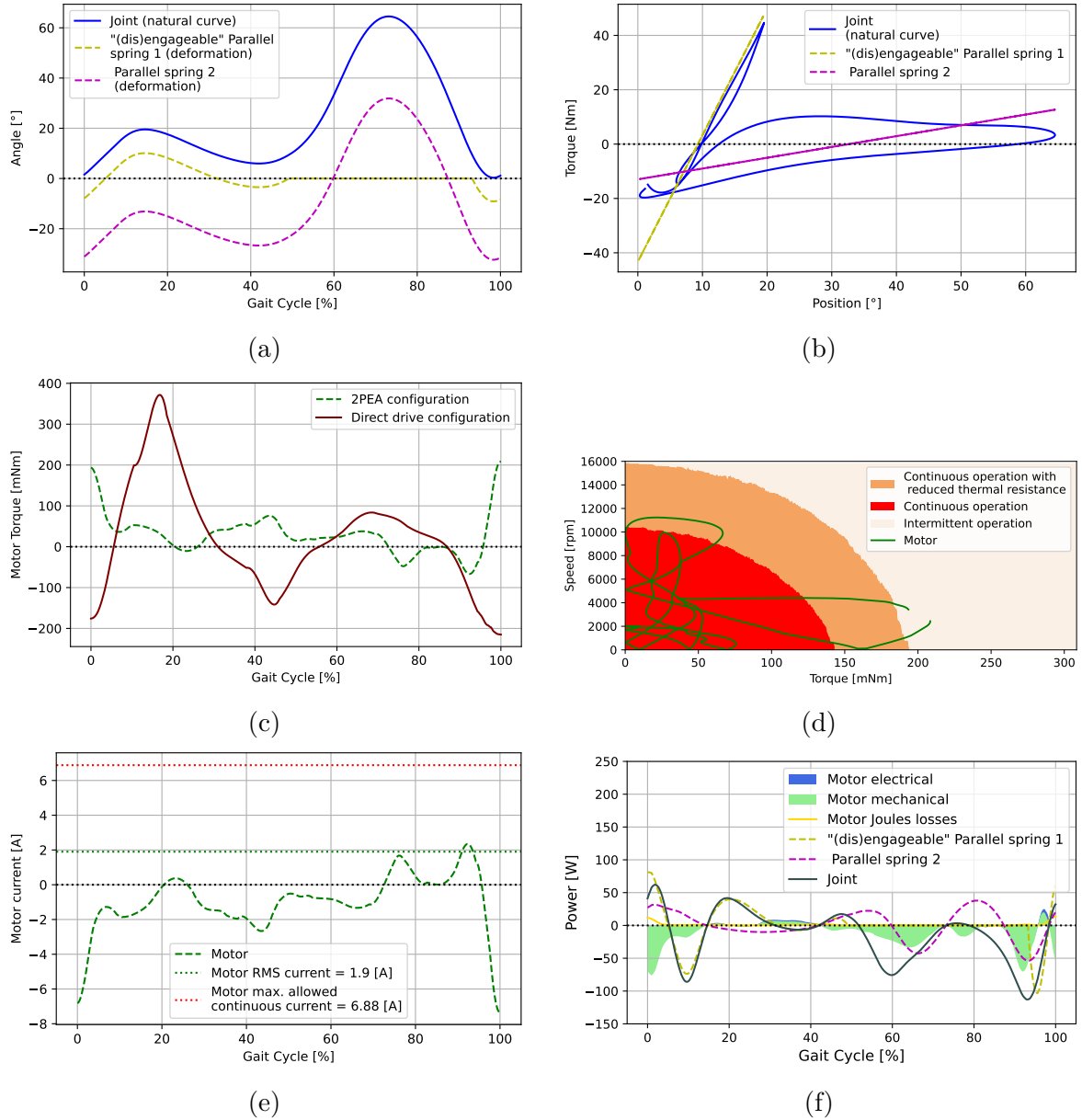


Figure 5.19: Walking profiles for the 2PEA configuration embedding a Faulhaber BP3274 motor optimized for walking only (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) motor current (f) powers

k_p [Nm/rad]	k_{p2} [Nm/rad]	θ_{0,k_p} [°]	$W \beta_{en}$ [%]	$W \beta_{disen}$ [%]	N_{red} [-]
267	23	33	93.1	49.2	162

Table 5.5: Parameters for the 2PEA configuration embedding a Faulhaber BP3274 motor optimized for walking only (W stands for "Walking")

State-of-art active knee prostheses such as [61, 55] require approximately 2 [W] of electrical power for the embedded electronic system (encoders, motherboard, sensors, etc.). For a normal-speed walking cycle time of 1 [s], this accounts for 2 [J] of electrical energy consumption. Taking this value results in a total electrical energy consumption of $1.6 + 2 = 3.6$ [J] per cycle. Given this and considering a six-cell 24[V] lithium polymer battery with a 1200 [mAh] capacity (mass around 160 [g]), the autonomy of the 2PEA configuration with a Faulhaber BP3274 motor can be roughly evaluated as follows:

1. Total energy stored in the battery:

$$24 \text{ [V]} \cdot 1.2 \text{ [Ah]} = 28.8 \text{ [Wh]} \quad (5.42)$$

$$28.8 \text{ [Wh]} \cdot 3600 \left[\frac{\text{J}}{\text{Wh}} \right] = 103\,680 \text{ [J]} \quad (5.43)$$

2. Number of walking cycles:

$$\frac{103\,680 \text{ [J]}}{3.6 \left[\frac{\text{J}}{\text{cycle}} \right]} = 28\,800 \text{ [cycle]} \quad (5.44)$$

3. Autonomy:

$$28\,800 \text{ [cycle]} \cdot 1 \left[\frac{\text{s}}{\text{cycle}} \right] = 28\,800 \text{ [s]} = 8 \text{ [h]} \quad (5.45)$$

The optimized combination could provide up to 28 800 strides and 57 600 steps which exceed by far the 1500 steps that a transfemoral amputee takes on average per day [19]. This corresponds to a battery autonomy of 8 hours that falls within the targetted range defined in the specifications (reminder: 5-8 [h]). This is consistent with previous work aiming at reducing the energy consumption of active knee prostheses with elastic actuators (e.g. the CSEA: 30 000 strides for 8.7 hours of autonomy [55])

5.6.2 Combined walking and stairs ascent optimization

First, when considering an active task like stairs ascent, the motor's thermal behavior must be evaluated to prevent winding overheating during prolonged activity. Using the housing-ambient thermal resistance value (R_{th2}) provided in the datasheet consistently results in overheating after a few cycles. Therefore, a 50% reduction in R_{th2} is assumed to compare the thermal performance between different motors and configurations. Further consideration about this reduction will be developed in chapter 7.

Figure 5.20 shows the results yielded by the thermal model. Combinations of motor and elastic configuration that exceed the thermal limit are indicated by a red cross topped by the number of cycles achievable before reaching this limit, i.e. the maximum temperature allowed in the winding which is around 150-155 [°C] for all three motors. The other combinations are theoretically able to perform stairs ascent indefinitely without overheating and are defined in the graph by their winding steady-state temperature or in other terms, the winding temperature reached at thermal equilibrium after several time constants τ_w .

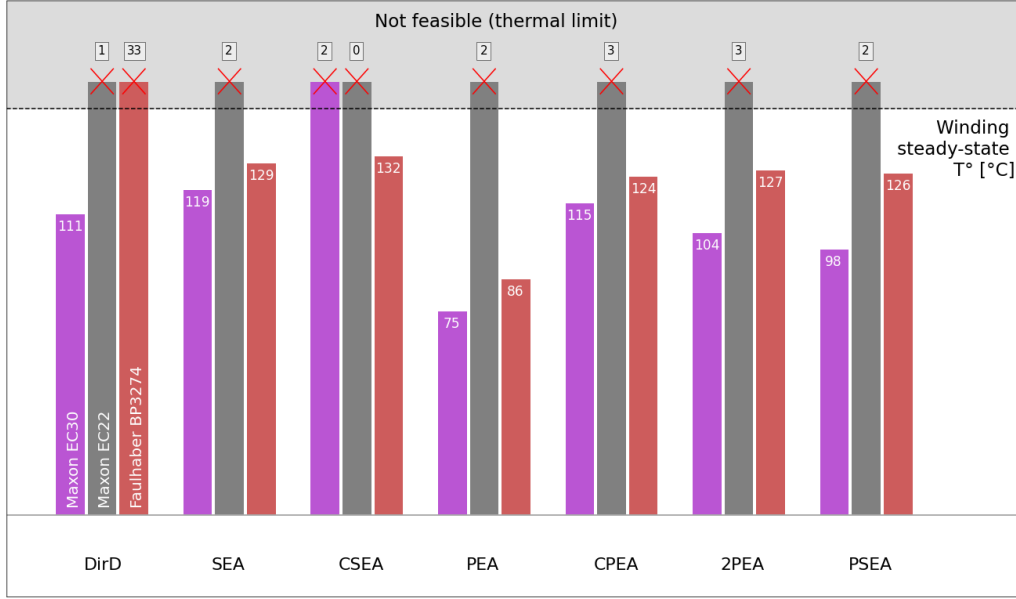


Figure 5.20: Comparison of the steady-state winding temperature reached at thermal equilibrium during sustained stairs ascent between direct drive and all six elastic actuators configurations optimized simultaneously for walking and stairs ascent. The red crosses indicate that the temperature exceeds the maximum allowed temperature in the winding, and the framed numbers above them correspond to the number of stair climb cycles that can be performed before reaching overheating.

The Maxon EC22 motor must be rejected as a candidate to drive a knee prosthesis incorporating an elastic actuator. In the best case scenario, this motor can only perform 3 cycles before overheating. This is largely due to its small size, which results in a limited heat transfer area, making it difficult to dissipate the excess heat generated inside the motor to the ambient environment. This is confirmed by the high thermal resistance values found in the datasheet (see Table 5.1).

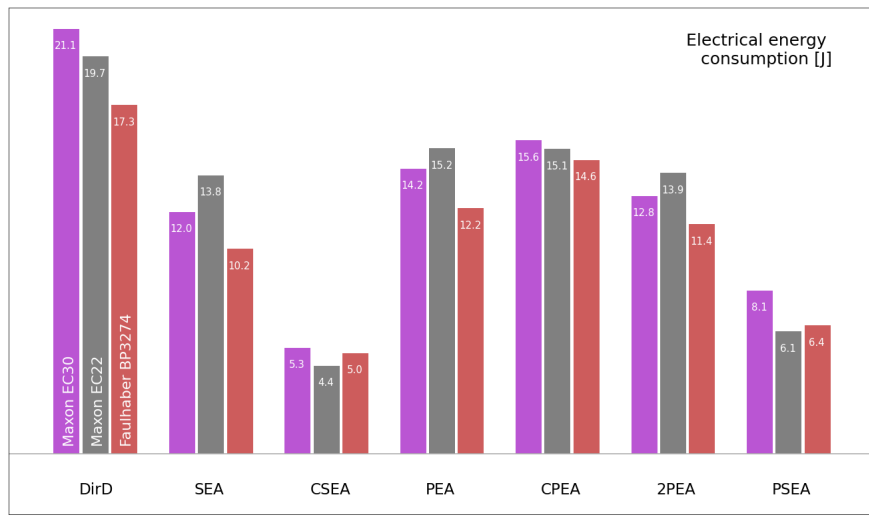
The other 2 motors, the Maxon EC30 and the Faulhaber BP3274 are capable of performing stairs ascent indefinitely whatever the configurations, except for the CSEA equipped with a Maxon EC30. The Maxon EC30 generally leads to lower winding steady-state temperatures, since its thermal resistances are lower than those of the Faulhaber BP3274.

When the optimization is performed simultaneously on walking and stairs ascent, the results in terms of the objective functions are substantially different, as it can be seen in Figure 5.21 and Figure 5.22. Neither walking nor stairs ascent performance is minimized by the configurations incorporating a (dis)engageable parallel spring, as it was the case for the optimization focused only on walking. Instead, it seems that elastic configurations including a series spring (SEA, CSEA, and PSEA) find a better trade-off in optimizing their parameters to minimize the objective functions for both tasks, given their respective weight in the optimization (see subsection 5.5.2).

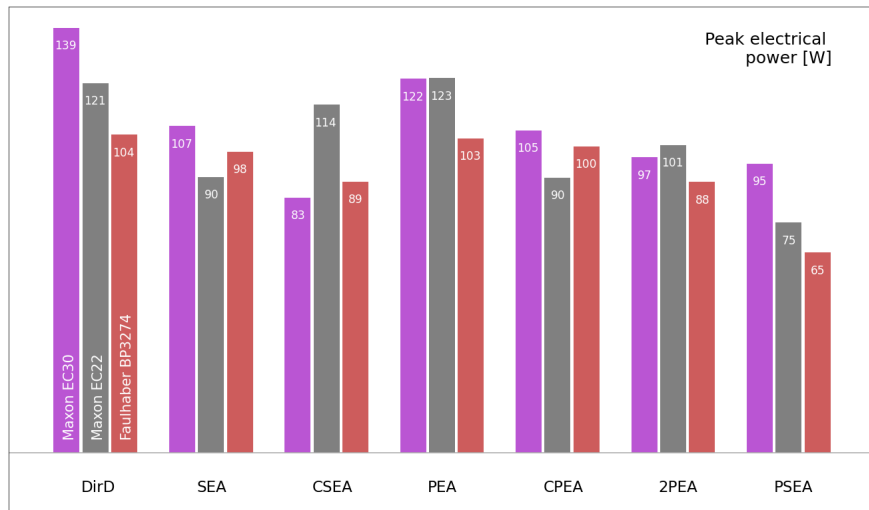
To determine the most advantageous combination of motor and elastic configuration, 4 different criteria are evaluated: the electrical energy consumption and peak electrical power for walking and stairs ascent. As explained before, the peak power is less important

to minimize since the candidate motors are already determined. The most important criterion among the remaining 2 is undoubtedly the energy consumption during walking (Figure 5.21a), as already explained.

From there, the optimal combination can be identified as **a clutchable series elastic actuator (CSEA) equipped with a Faulhaber BP3274 motor**. The 2 other motors give similar results in this configuration but must be rejected due to thermal considerations detailed above. The electrical energy consumption over one walking cycle for this combination is estimated to be around 5.0 [J], representing a reduction of 71.1% compared to the estimation of 17.3 [W] in direct drive. The walking peak electrical power estimate is 89 [W], which corresponds to a reduction of 14.4% compared to the estimation of 104 [W] in direct drive.



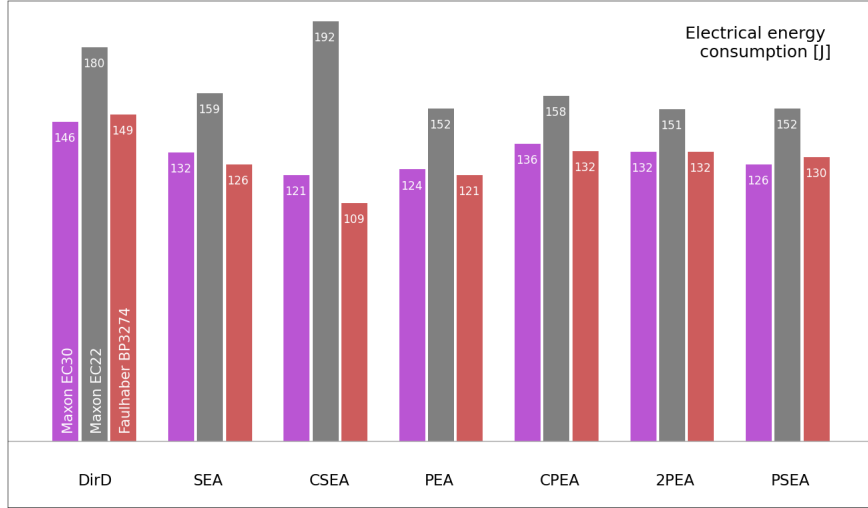
(a)



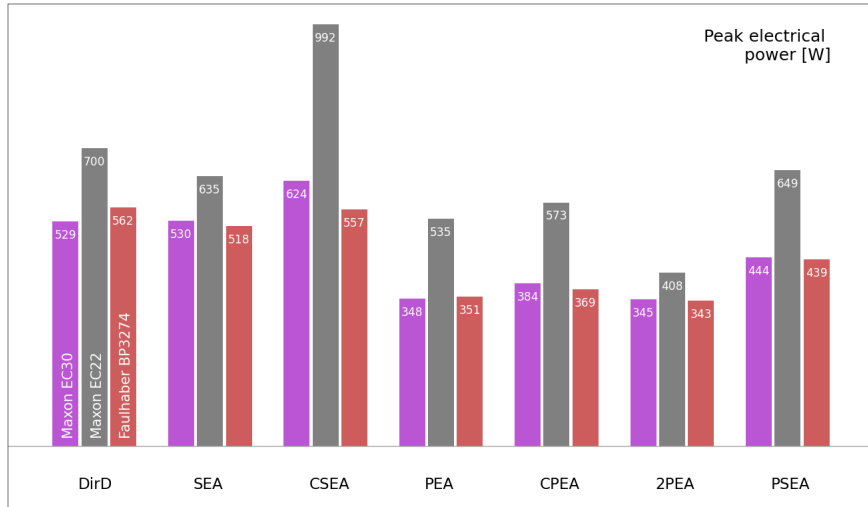
(b)

Figure 5.21: Comparison of walking performance between direct drive and all six elastic actuators configurations optimized simultaneously for walking and stairs ascent (a) Electrical energy consumed by the motor over one walking cycle (b) Peak motor electrical power during walking

This choice is further strengthened when looking at electrical energy consumption over one stairs ascent cycle (Figure 5.22a), for which the CSEA with a Faulhaber BP3274 displays again the minimal value among all studied combinations. Indeed it is estimated by the model around 109 [J], constituting a 26.8% reduction with respect to the direct drive estimation of 149 [J]. The stairs ascent peak electrical power estimate is 557 [W], which represents a negligible reduction of 0.9% compared to the estimation of 562 [W] in direct drive.



(a)



(b)

Figure 5.22: Comparison of stairs ascent performance between direct drive and all six elastic actuators configurations optimized simultaneously for walking and stairs ascent (a) Electrical energy consumed by the motor over one stairs ascent cycle (b) Peak motor electrical power during stairs ascent

Figure 5.23 and Figure 5.24 gather interesting graphs to better understand the benefits of the CSEA during walking and stairs ascent, respectively, while Table 5.6 gives the optimal parameters for the CSEA.

The series spring has a high stiffness that fits the linear torque-angle relationship

during the walking ESF. As already developed in subsection 5.5.3, this optimal stiffness allows the series spring to capture the power fluctuations by acting as a mechanical energy buffer between the load and the motor (see Figure 5.23d), therefore flattening the motor mechanical power profile.

The series spring provides the torque at the joint however, an equivalent reactionary torque is still necessary on the motor side. The whole advantage of the CSEA lies in its ability to clutch the motor during a certain period of the cycle so that it doesn't consume energy during that time. To achieve a significant gain in energy consumption, the clutch should engage passively or using much less energy than the motor would need to provide the reactionary torque itself.

As illustrated in Figure 5.23a, the motor is bypassed during ESF, letting the spring render the conservative torque-angle relation and the clutching element provide the reaction torque on the motor shaft. During the bypassing period, it is assumed that the motor doesn't provide torque, and therefore doesn't consume any mechanical or electrical power, as visible in Figure 5.23c and Figure 5.23d. Lastly, the motor operates within the continuous zone, ensuring that it never overheats (Figure 5.23b).

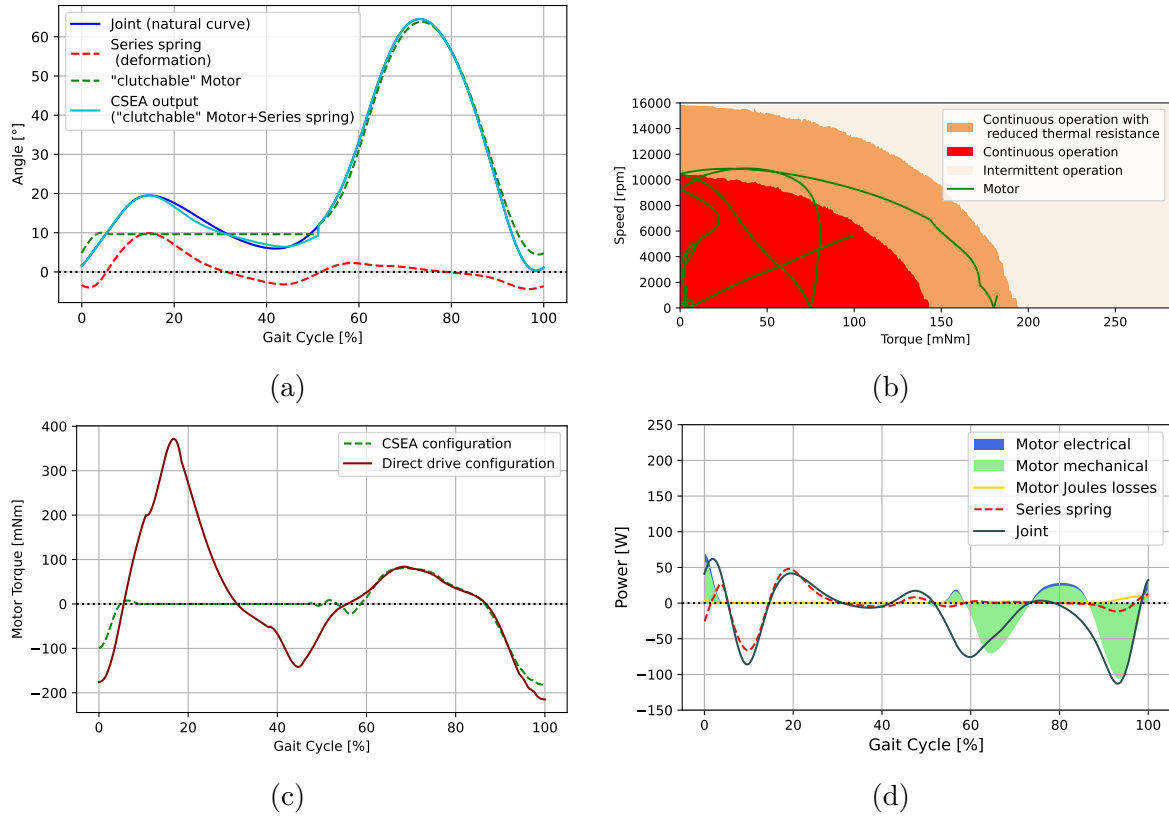


Figure 5.23: Walking profiles for the CSEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) speed-torque operating range (c) motor torque (d) powers

For stairs ascent, the gains are less evident. This is partly because the series spring stiffness can't be optimal for different tasks. Additionally, springs are passive elements that need to store mechanical energy before releasing it. In the case of the walking ESF, a spring is particularly adapted since a negative power flow during early flexion allows to

store energy while the early extension that follows directly is associated with a positive power flow that the spring can render by releasing its accumulated energy.

This is not the case for stairs ascent. Figure 5.24d shows that the high positive power peak needed during stairs ascent early stance to climb the step is not preceded by a negative power peak. At first, the spring stores energy and works against the motor providing a positive power flow, therefore increasing the motor peak required torque and Joule's losses (that consequently increase the motor electrical power).

This peak torque is visible in Figure 5.24b and Figure 5.24c, constituting a torque of around 700 [mNm] that is 4.5 times larger than the maximum continuous torque, hence located in the intermittent zone in the speed-torque operating range diagram. When the spring is releasing its energy, the demand on the motor in terms of mechanical power is decreased, to the point where the clutch can be engaged to bypass the motor.

As seen in Figure 5.24a, the motor is bypassed during the forward continuance and initial swing phase, preventing the CSEA output position (cyan curve) from extending as much as a natural knee joint would during those phases. The numerical value of the scalar agreement F_3 defined in Equation 5.36 is $0.012 [rad^2/s]$.

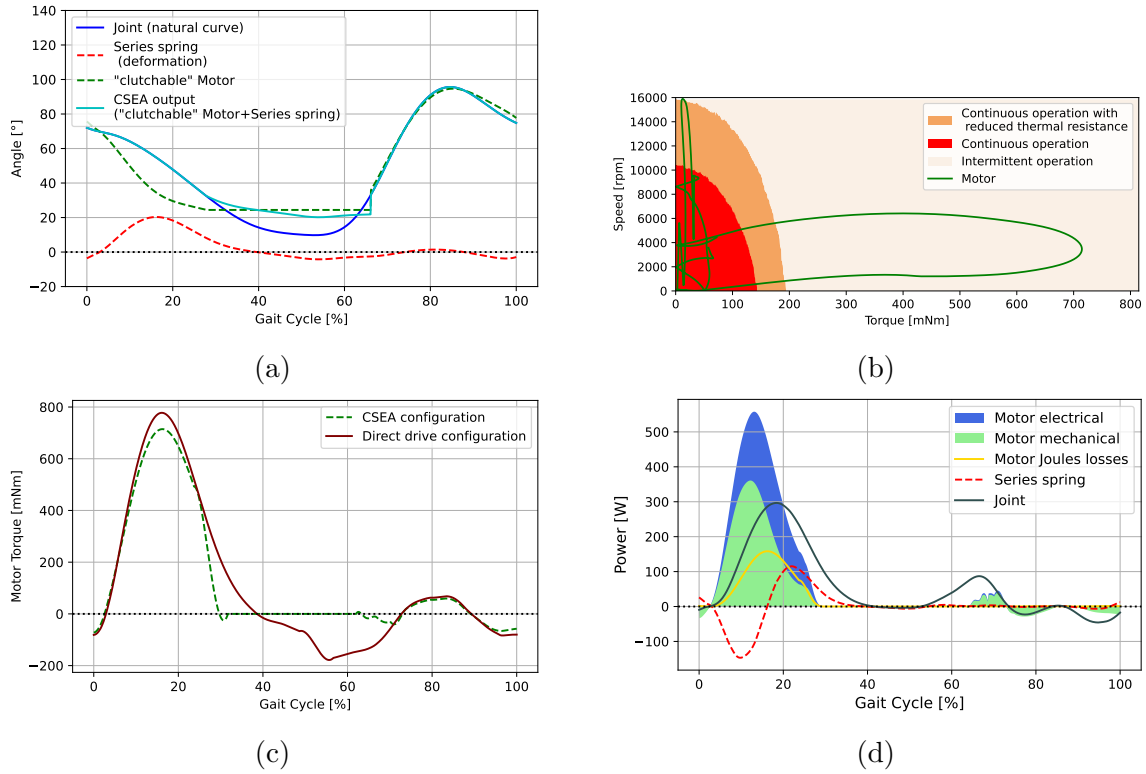


Figure 5.24: Stairs ascent profiles for the CSEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) speed-torque operating range (c) motor torque (d) powers

k_s [Nm/rad]	W β_{en} [%]	W β_{disen} [%]	SA β_{en} [%]	SA β_{disen} [%]	N_{red} [-]
258	6.4	51.1	27.7	66	179

Table 5.6: Parameters for the 2PEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (W stands for "Walking" and SA for "Stairs Ascent")

The autonomy of the CSEA configuration with a Faulhaber BP3274 can be approximately evaluated with the same model as previously, i.e. with equations from Equation 5.42 to Equation 5.45, and considering a six-cell 24[V] lithium polymer battery with a 2400 [mAh] capacity (mass around 400 [g]). A battery with a higher capacity is needed since the electrical energy consumption for walking has increased. Still considering 2 [W] of electrical power for the electronic system, the total electrical energy consumption are:

- $5 \text{ [J]} + 2 \text{ [W]} \cdot 1 \text{ [s]} = 7 \text{ [J]}$ per walking cycle
- $109 \text{ [J]} + 2 \text{ [W]} \cdot 1.4 \text{ [s]} = 111.8 \text{ [J]}$ per stairs ascent cycle

Assuming that a transfemoral amputee may climb up 100 stairs (so 50 stairs ascent cycle) with one battery recharge, the number of walking cycles that can be performed is evaluated as:

$$\frac{207\,360 \text{ [J]} - 50 \cdot 111.8 \text{ [J]}}{7 \left[\frac{\text{J}}{\text{cycle}} \right]} = 28\,820 \text{ [cycle]} \quad (5.46)$$

which corresponds again to more than 57 000 steps that significantly exceed the need of an amputee (around 1500 steps in average). This leads to a battery autonomy of 8 hours that respects the specifications (reminder: 5-8 [h]). This is similar to the results obtained with the optimization focused only on walking, except that here, a heavier battery with twice the capacity is required.

Let's denote that not all graphical results for each configuration were displayed. However, any diagrams not included in this chapter can be found in Appendix D.

5.7 Conclusion

In this chapter, the modeling and optimization of kinematic chains incorporating elastic actuators in active transfemoral prostheses were studied. Six different configurations were investigated to assess how the integration of elastic elements can improve energetic efficiency and reduce power demands during various lower limb tasks. A program was developed to compare the outcomes of optimizations for walking alone and for both walking and stairs ascent, focusing on electrical energy consumption and peak electrical power. The key findings are summarized below.

1. Walking Optimization:

- When optimizing for walking only, elastic actuator configurations including a (dis)engageable parallel spring (CPEA and 2PEA) perform better.
- The best performance for walking is achieved with a double parallel elastic actuator (2PEA) combined with a Faulhaber BP3274 motor.
- This configuration results in an electrical energy consumption of approximately 1.6 [J] per walking cycle, a reduction of 90.8% from the 17.3 [J] required for a direct drive.
- The peak electrical power is reduced by 42.3%, from 104 [W] in direct drive to 60 [W].

2. Simultaneous Optimization for Walking and Stairs Ascent:

- When optimizing for both walking and stairs ascent, elastic actuator configurations with a series spring (SEA, CSEA, and PSEA) perform better.
- The optimal configuration is a clutchable series elastic actuator (CSEA) with a Faulhaber BP3274 motor.
- This setup reduces the energy consumption for walking to 5.0 [J] per cycle, a 71.1% decrease from the 17.3 [J] for direct drive, and reduces peak electrical power by 14.4% (from 104 [W] to 89 [W]).
- For stairs ascent, the CSEA configuration consumes 109 [J] per cycle, a reduction of 26.8% from the 149 [J] for direct drive, with a peak electrical power of 557 [W], a marginal reduction of 0.9% from 562 [W] for direct drive.

Each configuration presented its advantages, enhancing energetic efficiency by minimizing the motor's workload during specific phases of various lower limb tasks. It was shown that inserting elastic actuators into the kinematic chain, particularly in clutchable configurations, significantly reduces the electrical energy consumption and peak power requirements of active transfemoral knee prostheses.

The results of the combined optimization are acknowledged as debatable. When optimized only for walking, the 2PEA configuration demonstrates the best performance in terms of the most crucial objective function, walking energy consumption. However, when its parameters are re-optimized for both walking and stairs ascent, this consumption rises from 1.6 [J] to 11.4 [J]. In contrast, the CSEA configuration shows more stability between the two optimizations, with its consumption increasing from 3.8 [J] to 5 [J].

Therefore, it would be worthwhile to compare a walking-optimized 2PEA that performs stairs ascent in direct drive (to preserve optimal walking parameters) and a CSEA whose clutch would only engage during a fraction of the walking cycle to avoid deviating too much from the curves during stairs ascent and reaching under-extensions (see Figure 5.24a). Chapter 7 addresses these comparisons and also takes into account thermal performance, weight, and complexity.

Chapter 6

Variable transmissions - Kinematic chains modeling and optimization

In chapter 3, the biomechanical analysis of the knee joint revealed a crucial insight that will motivate this whole chapter: The biological knee joint exhibits a wide range of torque and speed requirements that vary significantly within and between the lower limb common tasks.

For example, for both ground walking and stairs ascent, the knee joint requires a high torque at low velocity during the stance phase, whereas the swing phase demands a high velocity at low torque. Consequently, the biological knee joint generates a broad range of torques and speeds throughout ambulation, with the peaks of knee torque and speed occurring at different times.

Additionally, the peak torque differs substantially between the different tasks. The required peak torque is nearly double in stairs ascent (around 1.2 [Nm/kg]) than during walking (around 0.7 [Nm/kg]).

This biomechanical complexity presents a significant challenge for the actuators in active knee prostheses, which must replicate these variable requirements to provide natural and efficient movement without exceeding their operational limits. Indeed, as discussed in subsection 5.4.3, the operating range of a DC motor is constrained by its maximum allowable speed and nominal torque, determined by its design. These parameters do not represent absolute limits but define a range within which the motor can operate continuously without risking mechanical damage or thermal failure.

Moreover, minimizing the encumbrance and weight of the motor—and consequently, of the prosthesis—is highly desirable for several reasons:

- Energy efficiency: a smaller motor typically has lower rotor inertia, reducing the inertial torque and thus the mechanical torque that the motor has to provide (see Equation 5.17). This leads to lowered Joule's losses and increased energy efficiency.
- Amputee comfort: a lighter prosthesis reduces the physical burden on the user, allowing for higher activity levels. A discreet design can also be important for some patients who might enjoy a less noticeable appearance.
- Gait dynamics: reducing the weight of the prosthesis helps achieve a more natural gait, as the device mimics more closely the weight and dynamics of a biological limb.

To design a kinematic chain capable of providing a large range of torque and speed while minimizing the size and weight of the motor, integrating a transmission with a variable reduction ratio throughout the gait cycle is a popular approach. In this chapter, four different concepts of variable transmission are investigated:

- a Connecting Rod-Crank design (CRC)
- a Continuously Variable Transmission (CVT) based on [9]
- a 2-stage Planetary Gear (2PG)
- Utah bionic leg [61]

As the objective of this work is to propose an energy-efficient prosthesis, the variation of the reduction ratio for each configuration is performed passively, relying on 1 or 2 input variables among the following: the knee joint angular position, the vertical component of the ground reaction force (GRF) and the torque applied at the knee joint.

For each concept, a simulation model is developed to optimize performance, first exclusively for walking and then simultaneously for both walking and stairs ascent, exactly as in chapter 5. The results are compared and thoroughly discussed to determine whether or not variable reduction ratio concepts hold potential in this work.

6.1 Angle-based variable transmission

A first possibility is to passively adapt the reduction ratio based on the knee joint angular position. Indeed, this variable fluctuates during one gait cycle, and a well-tuned transmission could take advantage of that to adjust its reduction ratio without any energy input. A simple system for implementing this concept is a connecting rod-crank mechanism.

6.1.1 Connecting rod-crank model

Figure 6.1 shows the schematic section in the sagittal plane of a possible kinematic chain incorporating a connecting rod-crank (CRC) mechanism. A linear actuator (*DC motor + Gearhead + Ball screw*) drives the CRC assembly directly linked to the fixed revolute knee joint. The coordinate origin is situated at the knee joint, and the position of the ball screw nut is described by a (fixed) along the x-axis and λ (variable) along the y-axis. Lengths of the connecting rod and the crank are fixed, and have values of L and r , respectively, while θ_1 represents the angle formed between the crank and the positive x-axis. As this angle evolves during the gait cycle, the length of the moment arm associated with the CRC assembly changes, adjusting the reduction ratio of the transmission.

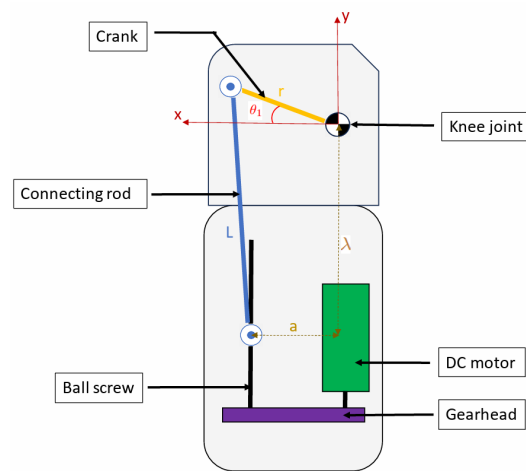


Figure 6.1: Scheme of a prosthesis incorporating a connecting rod-crank (CRC) mechanism.

As done in [65], it is assumed in the kinematics study that the lower section of the knee remains stationary, while the upper part rotates around the knee joint. Consequently, the knee joint angle θ_j and the vertical displacement of the nut, s , can be deduced as:

$$\theta_j = \theta_{1,init} - \theta_1 \quad (6.1)$$

$$s = \lambda_{init} - \lambda \quad (6.2)$$

with $\theta_{1,init}$ and λ_{init} the initial values for θ_1 and λ in a position of full knee extension. From there, the variable reduction ratio between the ball screw and the knee joint can be derived as done in Appendix B. The model leads to:

$$N_{crc} = \frac{r\lambda \cos \theta_1 - ra \sin \theta_1}{\lambda - r \sin \theta_1} \quad (6.3)$$

The contribution of the ball screw and gearhead are taken into account to obtain the global reduction ratio from the motor to the knee joint (graphic example in Figure 6.2):

$$N_{red} = N_g \cdot N_{bs} \cdot N_{crc} \quad (6.4)$$

with $N_{bs} = \frac{2\pi}{P_h}$, P_h being the lead i.e. the linear displacement the nut makes for each rotation of the screw ($\left[\frac{mm}{rot.}\right]$ units in datasheets). Data from the Ewellix precision rolled ball screws catalog [11] are chosen. Among the different types of ball screws presented, the *SD/BD* category was selected (Rolled thread miniature ball screw, nut with threaded nose) for its several advantages, listed below:

- Smooth running
- Extremely compact nut design with threaded nose for easy assembly
- Excellent repeatability with high positioning accuracy

The catalog also provides the following formula to estimate the practical efficiency of the ball screw:

$$\eta_{bs} = \frac{0.9}{1 + \frac{\pi d_0}{P_h} \mu} \quad (6.5)$$

with d_0 the nominal diameter and $\mu = 0.006$ for the *SD/BD* category.

Choosing a lead $P_h = 2$ [mm] for the ball screw leads to $N_{bs} = 3141$, which is a very high reduction ratio. Given that, a gearhead composed of a single gear pair is sufficient to achieve the needed global reduction ratio. Moreover, helical gears are preferred over spur gears since they're less noisy. In the case of the CRC model, the reduction ratio of the gearhead N_g is set as an optimization variable, knowing that it will be feasible to find a pair of helical gears whose ratio is close to the optimized one. In terms of efficiency, a typical value for a gear is considered: $\eta_g = 97\%$. The other optimization variables are all geometric parameters, as shown in Table 6.1

L [cm]	r [cm]	λ_{init} [cm]	a [cm]	N_g [-]
[11, 15.5]	[3.5, 5]	[8, 13.5]	[2.5, 3.5]	[0.2, 10]

Table 6.1: Optimization variables ranges for the CRC concept

Figure 6.2 displays the evolution of the CRC global reduction ratio as a function of the knee joint, for a CRC mechanism powered by a Maxon EC30 motor. This graph results from a concurrent optimization of walking and stairs ascent, where the simulation converged to a maximal reduction ratio within the initial 60° of knee flexion. This range covers the torque peaks for most high-torque activities, as already outlined in [65].

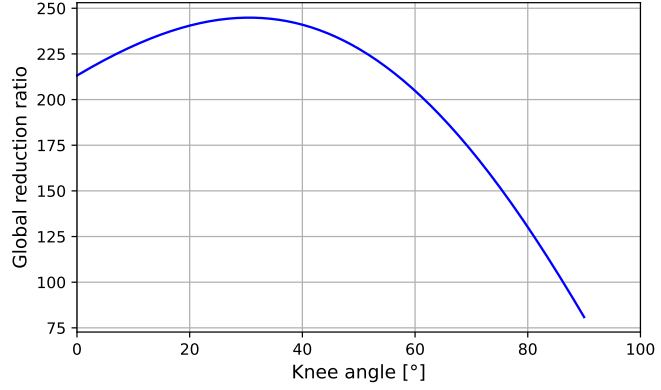


Figure 6.2: Evolution of the global reduction ratio relative to the knee angle for a CRC equipped with a Maxon EC30 motor.

6.2 GRF-based variable transmission

Another innovative approach could be to use the ground reaction force (GRF) as a means to passively adapt the reduction ratio of the transmission. In chapter 3, data for the vertical component of the GRF (vGRF) during both walking and stairs ascent cycles were extracted from previous scientific research. Using the GRF as an input variable to change the reduction ratio could be interesting because it distinguishes the stance phase from the swing phase, which respectively requires a high and a low reduction ratio. Indeed the GRF acts on the prosthesis during the stance phase while disappearing during the swing phase. To implement this concept, 2 models were developed: a continuously variable transmission and a 2-stage planetary gear.

6.2.1 Continuously variable transmission model

The continuously variable transmission (CVT) studied in this section and illustrated in Figure 6.3 is attributed to [9]. This CVT is a modified planetary gear, whose planets are conical and mounted on inclined shafts, and ring is made of diabolo-shaped rollers. This configuration allows translating the ring, thereby adjusting the contact point radius on the cones and consequently, the reduction ratio. The first modified planetary gear is followed by a second classical planetary gear (PG), to extend the range of reduction ratio achievable by the transmission.

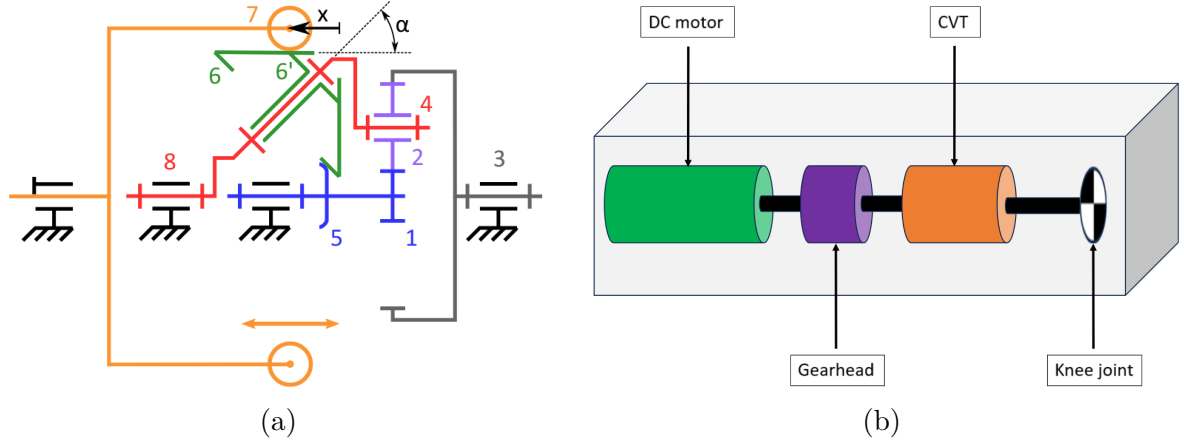


Figure 6.3: Continuously variable transmission (CVT) (a) kinematic scheme of the CVT. Taken from [9] (b) Schematic drawing of a kinematic chain incorporating the CVT

Among various existing CVT technologies, this one was chosen because changing the ratio requires virtually no energy and can be performed even when the input shaft is at rest. This is due to its reliance on rolling rather than sliding, unlike common CVT technologies that must overcome viscous friction (needed to transmit power), requiring the transmission to rotate quickly enough to make elements slip at their contact points. However, the ability to change ratio at low speeds is particularly desirable for lower limb prosthesis applications.

As demonstrated in [9], the variable reduction ratio is function of the different pitch radii, the CVT ring varying axial position x and the inclination angle of the CVT conical planets α :

$$N_{cvt} = \frac{\omega_{in}}{\omega_{out}} = \frac{\omega_{pg,3}}{\omega_{cvt,8}} = 1 - \frac{R_{cvt,7} R_{cvt,6} R_{pg,1}}{R_{cvt,5} R_{pg,3}} \cdot \frac{1}{x \sin \alpha} \quad (6.6)$$

where

- CVT ring radius: $R_{cvt,7} = 30$ [mm]
- CVT cone radius: $R_{cvt,6} = 12$ [mm]¹
- CVT sun radius: $R_{cvt,5} = 10.9$ [mm]
- PG sun radius: $R_{pg,1} = 12$ [mm]
- PG ring radius: $R_{pg,3} = 30$ [mm]
- CVT cone inclination: $\alpha = 45$ [°]

These dimensions were kept since this transmission was originally developed with particular attention to encumbrance, aiming for potential use in lower limb prostheses. The global reduction ratio from the motor to the knee joint is obtained by taking into account the gearhead reduction ratio:

$$N_{red} = N_g \cdot N_{cvt} \quad (6.7)$$

Equation 6.6 shows that the reduction ratio varies with the axial position of the CVT ring x . The translation of this ring could be passively rendered by the deformation of a linear spring, loaded by the user's body weight during the stance phase, and unloaded

¹Radius at the cone base

during the swing phase.

A schematic drawing of this mechanism is displayed in Figure 6.4. The axial displacement of the CVT ring (in orange) is limited by hard stops (in dark blue). The minimal position is $x_{min} = 8$ [mm] as intended in the original design and the maximal one, x_{max} , is set as an optimization variable since it defines the maximal reduction ratio achievable by the CVT.

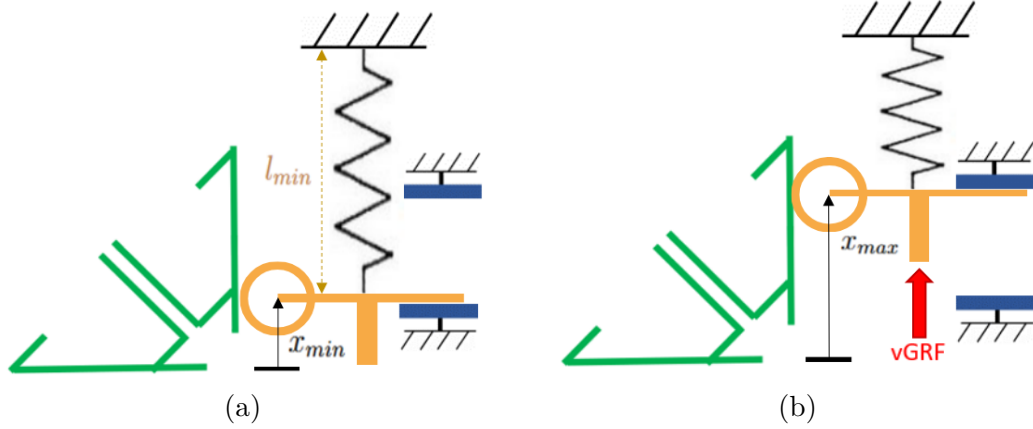


Figure 6.4: Schematic drawing of the linear spring mechanism to control the axial displacement of the CVT ring (a) minimal ring position reached during the swing phase (unloaded prosthesis) (b) maximal ring position reached during the stance phase (loaded prosthesis)

A linear spring with stiffness k and free length l_0 is integrated between the ring and the housing, pre-constrained to a length l_{min} inferior to l_0 . This ensures that even when the prosthesis is unloaded (no GRF) during the swing phase, the spring applies a compressive force

$$F_c = -k \cdot (l_0 - l_{min}) \quad (6.8)$$

maintaining the ring position at x_{min} to keep the reduction ratio at its minimal value and prevent unwanted displacement of the ring. The three characteristics of the spring are set as optimization variables. Their optimal values define F_c that the force recovered from the axial GRF (simulated in the model as the vertical component $vGRF$) must overcome before implying displacement of the ring.

Assuming the range of displacement of the ring is so small that the compressive force stays approximately constant, whatever the ring position, the net force acting on the ring is:

$$F_{net} = vGRF - F_c \quad (6.9)$$

Knowing the net force acting in the vertical orientation, the total deformation of the spring Δl can be computed as soon as this force becomes positive (directed upwards).

$$\Delta l = \frac{F_{net}}{k} = \underbrace{(l_0 - l_{min})}_{\text{init. deformation}} + \underbrace{(x - x_{min})}_{\text{new deformation}} \quad (6.10)$$

$$\leftrightarrow x = \frac{F_{net}}{k} - (l_0 - l_{min}) + x_{min} \quad (6.11)$$

Equation 6.11 gives the evolution of the CVT ring position x as a function of the net force acting on it and the characteristics of the spring. Let's denote that the simulation considers one single spring however, for a more balanced application of the force on the ring, the number of springs could easily be doubled or quadrupled by dividing the optimized stiffness and/or initial deformation by the appropriate ratios. Eventually, Table 6.2 gives the variables set as optimization parameters in the optimization as well as their range.

k [N/m]	l_0 [cm]	l_{min} [cm]	x_{max} [mm]	N_g [-]
[200, 5000]	[5, 12]	[3, 6]	[10, 16]	[50, 300]

Table 6.2: Optimization variables ranges for the CVT concept

The evolution of the vertical ground reaction force and the CVT global reduction ratio over one cycle can be observed in Figure 6.5, for both walking and stairs ascent. These are the results of a combined optimization of the 2 tasks, for a CVT driven by a Maxon EC30 motor. The optimal parameters are such that the high reduction ratio is kept during the whole stance phase, in response to the high torque requirement, before returning to the low reduction ratio when the GRF vanishes.

This force is applied very abruptly to the prosthesis, and due to the narrow displacement range of the CVT ring, the change of ratio is almost instantaneous, failing to utilize the continuous nature of the mechanism. Another significant drawback is the mechanical efficiency of this variable transmission. Experiments reported in [9] show an average efficiency of approximately 85% for negative power flow and only about 75% for positive power flow. Furthermore, this variable transmission provides relatively low reduction ratios, requiring the gearhead to achieve a large reduction ratio, which results in poor mechanical efficiency as discussed in subsection 5.4.1.

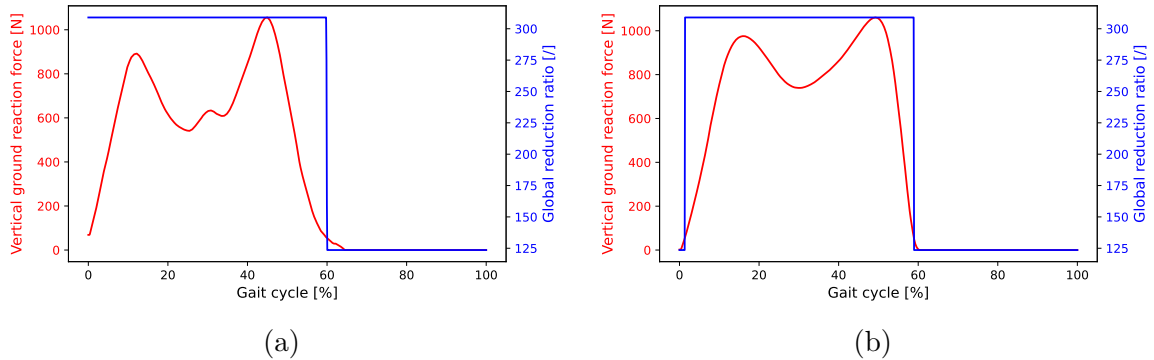


Figure 6.5: Evolution of the vertical ground reaction force (in red) and the CVT global reduction ratio (in blue) over one cycle of (a) walking (b) stairs ascent

6.2.2 2-stage planetary gear model

Another approach to implementing a variable reduction ratio using GRF as its input variable is a 2-stage planetary gear (2PG) transmission, illustrated in Figure 6.6. The motivation for investigating this second GRF-based variable transmission is to evaluate the performance of a discrete ratio change compared to a continuous one. Additionally, the mechanical efficiency of classical planetary gears is higher than that of the CVT analyzed in the previous section, which positively impacts the speed-torque requirements imposed on the motor.

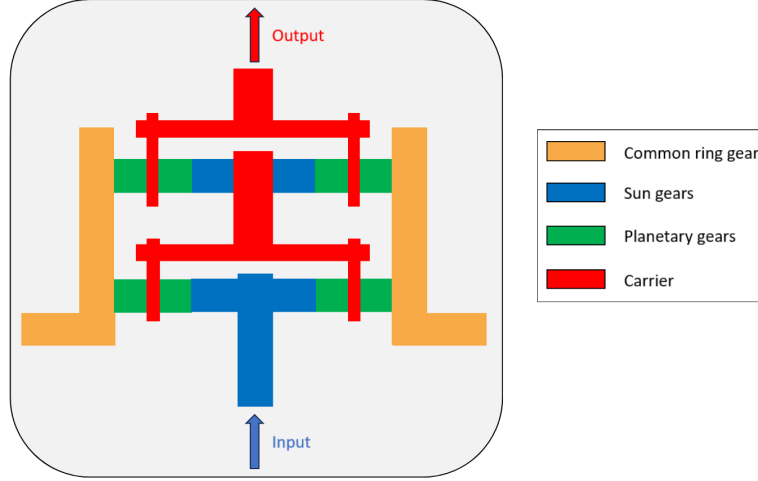


Figure 6.6: Schematic drawing of a 2-stage planetary gear (2PG) mechanism.

The input shaft of the 2PG is connected to the first sun gear (s_1). The first carrier links the first set of planetary gears to the second sun gear (s_2). The ring gear (r) is common to both stages of the mechanism. Lastly, the second carrier connects the second set of planetary gears with the output shaft. This type of transmission is well known for its ability to provide different reduction ratios, depending on which components are allowed to move and which are fixed. The Willis equation for planetary gears reads as follows:

$$\omega_r \cdot Z_r = \omega_c \cdot (Z_r + Z_s) - Z_s \cdot \omega_s \quad (6.12)$$

with ω_i the rotational speed and Z_i the number of teeth of part i . The indices r , c , and s denote the ring gear, carrier, and sun gear, respectively. From there, several scenarios are possible:

1. Fixed ring gear (input: sun gear & output: carrier)

$$\omega_r = 0 \rightarrow 0 = \omega_c \cdot (Z_r + Z_s) - Z_s \cdot \omega_s \quad (6.13)$$

$$\Leftrightarrow N_{pg} = \frac{\omega_s}{\omega_c} = \frac{Z_r + Z_s}{Z_s} = 1 + \frac{Z_r}{Z_s} \quad (6.14)$$

2. Fixed carrier (input: sun gear & output: ring)

$$\omega_c = 0 \rightarrow \omega_r \cdot Z_r = -Z_s \cdot \omega_s \quad (6.15)$$

$$\Leftrightarrow N_{pg} = \frac{\omega_s}{\omega_r} = -\frac{Z_r}{Z_s} \quad (6.16)$$

3. Fixed sun gear (input: ring gear & output: carrier)

$$\omega_s = 0 \rightarrow \omega_r \cdot Z_r = \omega_c \cdot (Z_r + Z_s) \quad (6.17)$$

$$\leftrightarrow N_{pg} = \frac{\omega_r}{\omega_c} = \frac{Z_r + Z_s}{Z_r} = 1 + \frac{Z_s}{Z_r} \quad (6.18)$$

Inspired by [7], 2 states are defined for the 2PG. A state where the common ring gear is fixed letting the power flows as shown in Figure 6.7a. This configuration results in a high reduction ratio desirable for the stance phase, given by Equation 6.14 applied at both stages:

$$N_{2pg,h} = \left(1 + \frac{Z_r}{Z_{s1}}\right) \cdot \left(1 + \frac{Z_r}{Z_{s2}}\right) \quad (6.19)$$

The second state stems from locking the first carrier, consequently fixing the second sun gear, allowing the power to flow as depicted in Figure 6.7b. This leads to a low reduction ratio for the swing phase given by the combination of Equation 6.16 and Equation 6.18:

$$N_{2pg,l} = -\left(\frac{Z_r}{Z_{s1}}\right) \cdot \left(1 + \frac{Z_{s2}}{Z_r}\right) \quad (6.20)$$

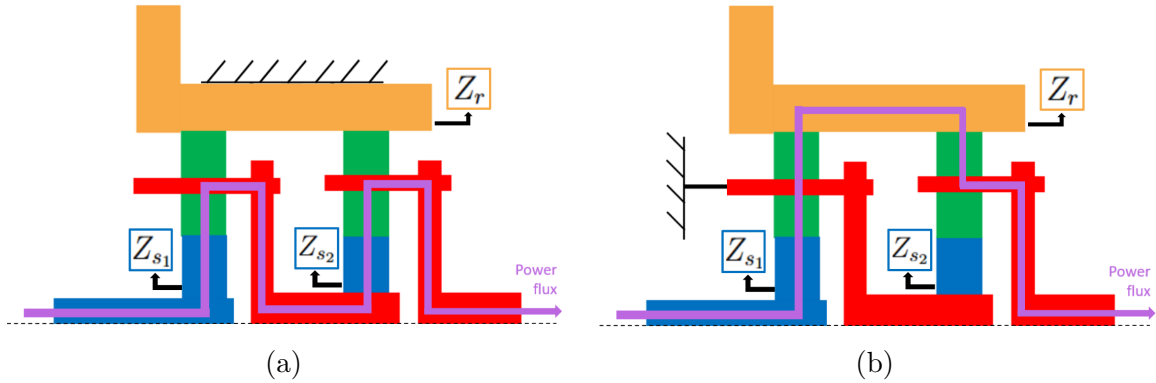


Figure 6.7: Schematic drawing highlighting the power flow for both states of the 2PG (a) High reduction ratio state for stance phase (common ring gear is fixed) (b) Low reduction ratio state for swing phase (first carrier and second sun gear fixed)

The discrete change between the high and low reduction ratios must rely on the application of the user's body weight on the prosthesis. To ensure that the variation can occur at any angular position, a braking concept is developed (Figure 6.8). Similar to the CVT concept, the principle is to recover the axial component of the GRF to translate a part of the mechanical system. However, in this case, instead of directly affecting the motion of a transmission part², the GRF during the stance phase displaces a circular braking pad. This pad, combined with another one located above the rim of the common ring gear, brakes the latter until it locks (Figure 6.8a). When the GRF vanishes during the swing phase, the brake pad returns to its initial position and, combined with another circular

²As it was the case for the CVT, where the GRF displaced the ring gear

brake pad, clamps the disk brake fixed coaxially to the first carrier until it locks. Preloaded linear springs provide the clamping force necessary to brake the disk (Figure 6.8b).

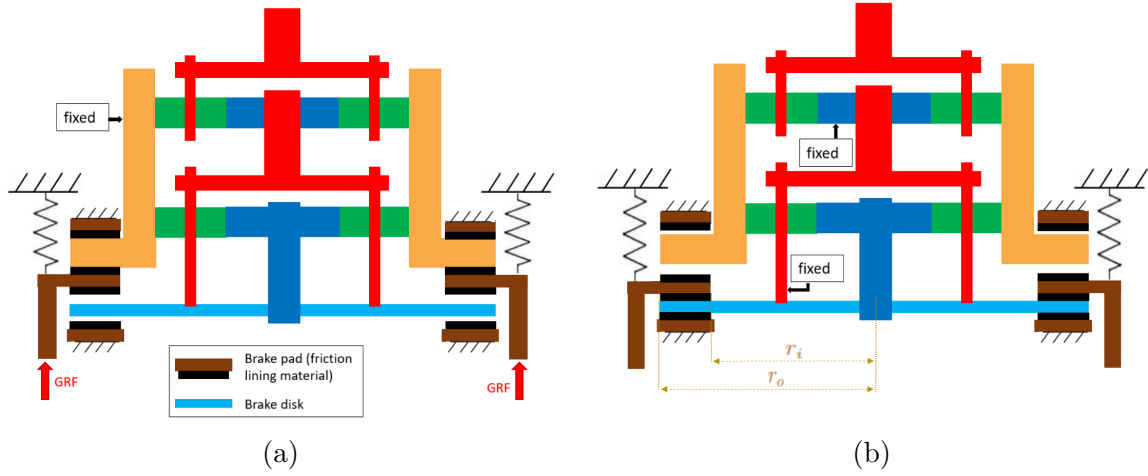


Figure 6.8: Schematic drawing of the 2PG braking concept (a) Common ring gear locked during the stance phase (b) First carrier and second sun gear locked during the swing phase

A disk brake is similar to a clutch except that one of the shafts is replaced by a fixed member. Assuming a uniform rate of wear at the friction interfaces, [24] gives a formula for the torque capacity T_{lock} as a function of the axial clamping force F_{clamp} :

$$T_{lock} = F_{clamp} \cdot f \cdot \left(\frac{r_o + r_i}{2} \right) \cdot N \quad (6.21)$$

with r_i and r_o the inner and outer radius of the circular braking pad (see Figure 6.8b), $N = 2$ the number of friction interfaces, and f the dynamic friction coefficient that can be estimated with a factor of safety around 0.3 if the pads are in sintered metal and the disks are in cast iron or steel. The clamping force along the gait cycle is given by:

$$F_{clamp} = vGRF + F_c \quad (6.22)$$

$$\text{where } F_c = -k \cdot \Delta l \cdot 4 \quad (6.23)$$

F_c is the compressive force developed by the linear springs that the $vGRF$ must overcome before being able to displace the brake pad. This force depends on the stiffness k and deformation Δl of those springs. The factor four corresponds to the number of springs, chosen to provide a balanced and well-distributed force.

[7] implemented a similar transmission and found that to cover all lower-limb tasks, the locking torque must hold against 5 [Nm] during the stance phase (when fixing the common ring gear) and only against 0.5 [Nm] during the swing phase (when fixing the first carrier and second sun gear). As a first approximation, those values were used in the simulation to determine the instant of the 2PG reduction ratio discrete change:

$$N_{2pg} = \begin{cases} N_{2pg,l} = -\left(\frac{Z_r}{Z_{s1}}\right) \cdot \left(1 + \frac{Z_{s2}}{Z_r}\right) & \text{if } T_{lock} < 5 \text{ [Nm]} \\ N_{2pg,h} = \left(1 + \frac{Z_r}{Z_{s1}}\right) \cdot \left(1 + \frac{Z_r}{Z_{s2}}\right) & \text{if } T_{lock} \geq 5 \text{ [Nm]} \end{cases} \quad (6.24)$$

$$N_{red} = N_g \cdot N_{2pg} \quad (6.25)$$

In the simulation, the time required to vary from one locking position to the other as well as the braking time are neglected, considering instantaneous variations of the reduction ratio. This seems to be a good approximation when looking at Figure 3.10e or Figure 3.13e, where one can see that the vGRF appears and vanishes very abruptly at the beginning and the end of the stance phase. Table 6.3 gives the optimization variables ranges chosen for the simulation, where $\Delta r = r_o - r_i$.

Z_r/Z_{s1} [-]	Z_r/Z_{s2} [-]	k [N/m]	r_i [cm]	Δr [cm]	N_g [-]
[1, 10]	[1, 10]	[100, 2500]	[1, 2]	[0.5, 1]	[1, 20]

Table 6.3: Optimization variables ranges for the 2PG concept

Similarly to the CVT concept, The evolution of the vertical ground reaction force and the 2PG global reduction ratio over one cycle can be seen in Figure 6.5, for both walking and stairs ascent. These graphs were obtained through a combined optimization of the 2 tasks, for a 2PG driven by a Maxon EC30 motor. Above a certain value of the GRF, the braking system locks the common ring gear, switching the transmission to its high reduction ratio state for the entire stance phase. During the swing phase, the first carrier and second sun gear are locked again, which reduces the ratio by more than half.

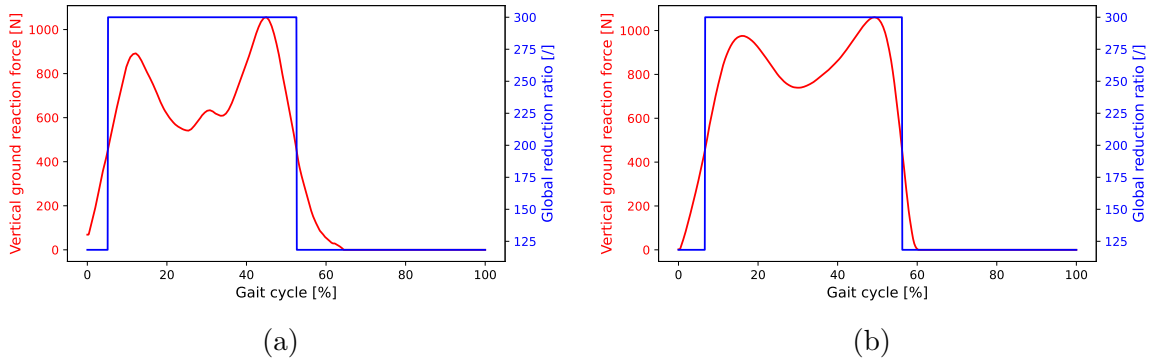


Figure 6.9: Evolution of the vertical ground reaction force (in red) and the 2PG global reduction ratio (in blue) over one cycle of (a) walking (b) stairs ascent

6.3 Torque-based variable transmission

A last input variable that could be used to passively adjust the reduction ratio is the torque applied at the knee joint. Indeed, the initial motivation for investigating variable transmission concepts was to reduce the range of speed-torque requirements at the level of the motor, while keeping a large range at the knee joint to cover all ambulation tasks. An ideal variable transmission should therefore continuously adapt its reduction ratio based on the torque required at the knee joint. An example of such transmission is found in the Utah bionic leg [61], which is modeled in the next section.

6.3.1 Utah bionic leg

The Utah Bionic Leg project aims to replicate the natural behavior of the knee, ankle, and toe while remaining within the weight, size, and autonomy ranges of semi-active prostheses. Here the focus is set on the knee module, which is designed as a combination of two kinematic chains and has thus two degrees of freedom. As a result, the torque ratio is not the inverse of the reduction ratio, unlike the single-degree of freedom concepts studied previously in this chapter.

The Utah bionic leg knee combined two input variables to adapt its reduction ratio: the knee angle, through a CRC mechanism as described in section 6.1, and the knee torque through a torque-sensitive joint, linked to a spring-actuated joint. The position δ_2 of this joint depends on the torque applied at the knee, and along with the knee angle, it defines the length of the moment arm formed by the connecting rod-crank assembly (as shown in Figure 6.10).

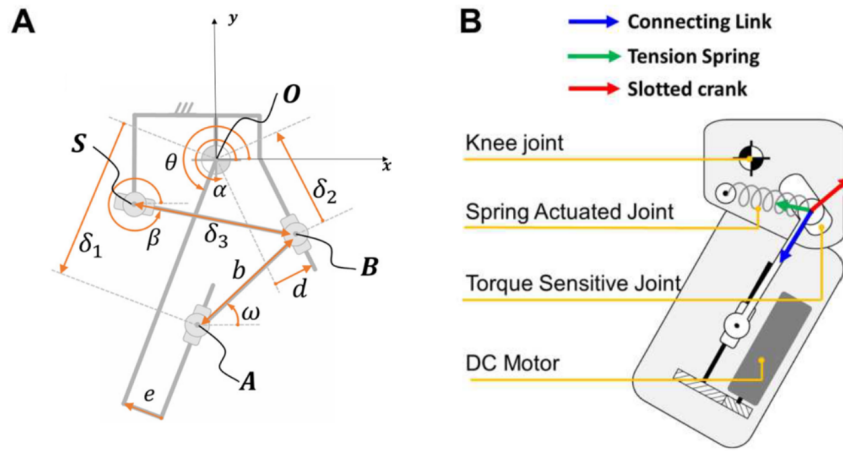


Figure 6.10: Taken from [61] (A) Kinematic diagram along with variable names (B) Schematic drawing highlighting the forces acting on the torque-sensitive joint.

It should be noted that in Figure 6.10 (A), θ does not represent the knee joint angle θ_j . The relation between both is $\theta = 270^\circ - \theta_j$. The other variables are described in Table 6.4 along with their numerical values in the original design.

Symbol	Description	Values from [61]
$\delta_{2,l}$	Lower limit of δ_2 translation	22 [mm]
$\delta_{2,h}$	Upper limit of δ_2 translation	35 [mm]
F_{s0}	Spring preload force	150 [N]
k	Spring stiffness constant	35 [N/mm]
S_x	Abscissa of point S	-5 [mm]
S_y	Ordinate of point S	-22 [mm]
b	Length of the connecting rod (the 'b' bar)	107 [mm]
e	Length of the 'e' bar	35 [mm]
d	Length of the 'd' bar	16 [mm]
α	Alpha angle of the mechanism	275 [°]
P_h	Ballscrew lead	2 [mm]
N_g	Gearhead ratio	2.5 [-]

Table 6.4: Utah bionic leg knee variables: description and values in the original design

The mathematical model is thoroughly described in Appendix C. The kinematics study leads to a second-order equation whose solving gives δ_1 , i.e. the position of point A in Figure 6.10 (A) which represents the position of the nut on the ballscrew. Its derivative allows to find the motor speed by taking into account the reduction ratios of the gearhead N_g and that of the ballscrew N_{bs} (recall: $N_{bs} = \frac{2\pi}{P_h}$):

$$\omega_m = (N_g N_{bs}) \cdot \dot{\delta}_1 \quad (6.26)$$

Moreover, the kinetics study allows to determine F_a , the force parallel to the ballscrew applied at point A , knowing the expression of the variable torque ratio τ_{torque} between the ballscrew output and the knee joint. From there, the motor torque T_m can be derived:

$$T_m = I \cdot a_m + \frac{F_a}{(N_g N_{bs})} \cdot \eta_{mec} \quad (6.27)$$

To better visualize the passive variation of the global torque ratio relative to the 2 input variables (knee joint angle and torque), Figure 6.11 was reproduced based on the original design of the Utah bionic leg. It can be seen that globally, the torque ratio increases with the absolute value of the knee torque, reaching its maximal values around 300 for knee flexion angles between 50° and 70° . This is ideal for stairs ascent, for which the peak torque falls within that zone of maximal torque ratio (Figure 6.11b). This is however less ideal for walking since the peak torque is reached during ESF for small flexion angles (between 10° and 20° , which are associated with intermediate values of the torque ratio (Figure 6.11a).

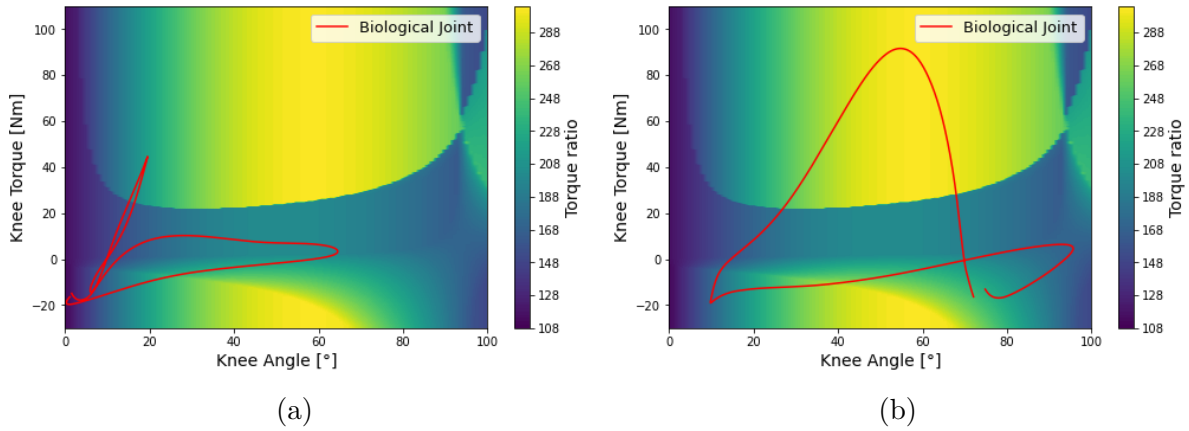


Figure 6.11: Knee torque ratio as a function of the knee angle and torque, reproduced for the original Utah bionic leg design. Red curves show the biological knee torque-angle relationship for (a) walking (b) stairs ascent

Of all the parameters presented in Table 6.4, only five are identified as having a significant influence on the torque ratio. The other parameters are fixed to the values attributed in the original design. Table 6.5 shows the ranges of the optimization variables³.

k [N/mm]	F_{s0} [N]	$\delta_{2,h}$ [mm]	b [cm]	N_g [/]
[17.5, 53.5]	[70, 315]	[25, 50]	[5.5, 16]	[1.25, 3.75]

Table 6.5: Ranges of the optimization variables kept for the Utah simulation

³Each of them encompasses the original design value

6.4 Comparison and discussion of the optimization results

In the previous sections, the global reduction ratio from the motor to the knee joint (or the torque ratio for the Utah bionic leg) is modeled for each concept and can be used to derive the motor speed and acceleration based on the biological curves of the knee joint (recall: $N_{red} = \omega_m/\omega_j$). From there, the DC motor dynamic equations and the procedure used to optimize the different variable transmission concepts are identical to those outlined in section 5.4 and section 5.5. The results for all combinations of motors and variable transmission concepts are compared by evaluating the minimized objective functions, initially for an optimization focused exclusively on walking, and then including both walking and stairs ascent.

The variable transmissions are compared to a direct drive transmission, which has a fixed optimized reduction ratio. The objective is to demonstrate the gain obtained by varying the reduction ratio throughout the cycle.

Most importantly, the thermal performance will be assessed for the optimization including stairs ascent. This evaluation aims to identify variable transmission concepts that can reduce the motor torque demands sufficiently to allow the use of a smaller motor, specifically the Maxon EC22 motor, to drive the prosthesis. Indeed, some concepts achieve this whereas all elastic actuator configurations equipped with a Maxon EC22 motor had to be excluded due to thermal constraints (see Figure 5.20).

6.4.1 Walking optimization

As visible in Figure 6.12, the best walking performance in terms of electrical energy consumption and peak electrical power is achieved by **a connecting rod-crank mechanism (CRC) powered by a Faulhaber BP3274 motor**. The two other motors provide very similar results, and could thus also be chosen to drive the prosthesis.

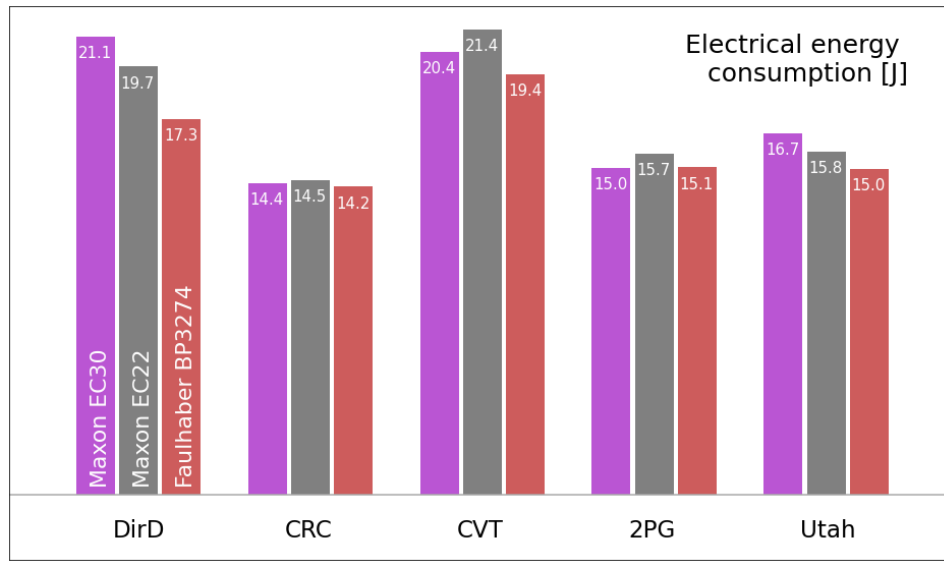
The electrical energy consumption over one walking cycle for this combination is estimated to be around 14.2 [J], representing a reduction of 17.9% compared to the estimation of 17.3 [J] in direct drive. The walking peak electrical power estimate is 93 [W], which corresponds to a reduction of 10.6% compared to the estimation of 104 [W] in direct drive.

These gains are quite low compared to the ones obtained with the best elastic actuator configurations (see section 5.6). This is explained by the fact that the primary objective of a variable transmission is not to reduce the mechanical power that the motor needs to provide, but rather to adapt its speed-torque requirements to limit power losses and allow choosing smaller motors without compromising the scope of tasks that can be performed. To this end, a CRC mechanism optimized exclusively for walking gives notable advantages, illustrated by the profiles shown in Figure 6.13.

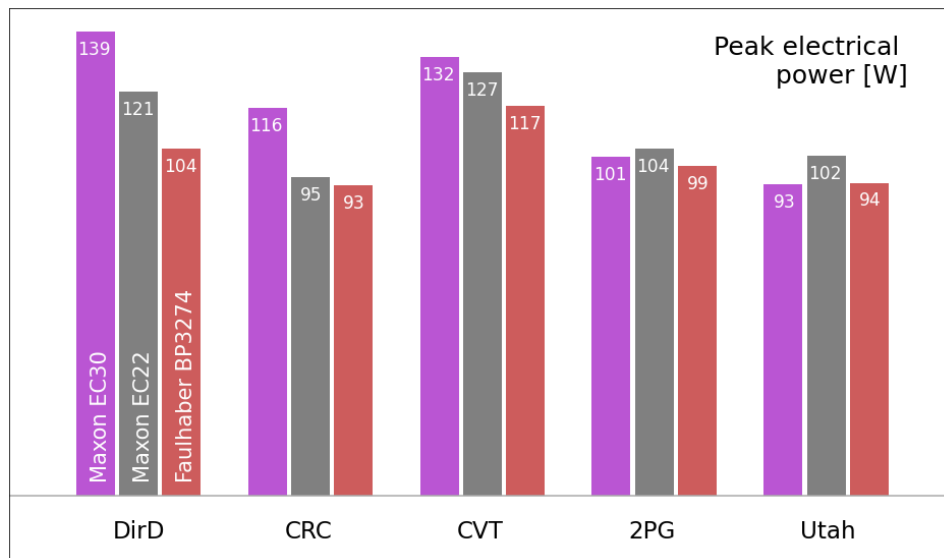
The optimal CRC parameters shown in Table 6.6 lead to a desirable global reduction ratio profile, given as a function of the knee flexion angle in Figure 6.13a. Indeed, the ratio is at its maximum value during the first 20° of flexion and progressively decreases for higher angles. This is optimal for walking, for which the peak torque (required during ESF) has to be provided at angles between 10 and 20° (see Figure 3.10d). On the other

hand, the swing phase requires lower reduction ratios, decreasing the speed demand on the motor compared to a direct drive configuration with a high fixed reduction ratio, as visible in Figure 6.13b.

However, the torque demand on the motor does not decrease compared to direct drive (Figure 6.13c), leading to high power losses, and consequently high motor electrical power during the stance phase (Figure 6.13d). The motor speed-torque operating range is displayed in Figure 6.13e, and even though if there is a short incursion in the intermittent zone, the motor RMS current is kept under the nominal current (Figure 6.13f), preventing the motor winding from overheating.



(a)



(b)

Figure 6.12: Comparison of walking performance between direct drive and all four variable transmission concepts optimized for walking only (a) Electrical energy consumed by the motor over one walking cycle (b) Peak motor electrical power during walking

L [cm]	r [cm]	λ_{init} [cm]	a [cm]	N_g [-]
11.6	3.9	11.6	3.1	1.2

Table 6.6: Parameters for the CRC mechanism embedding a Faulhaber BP3274 motor optimized for walking only

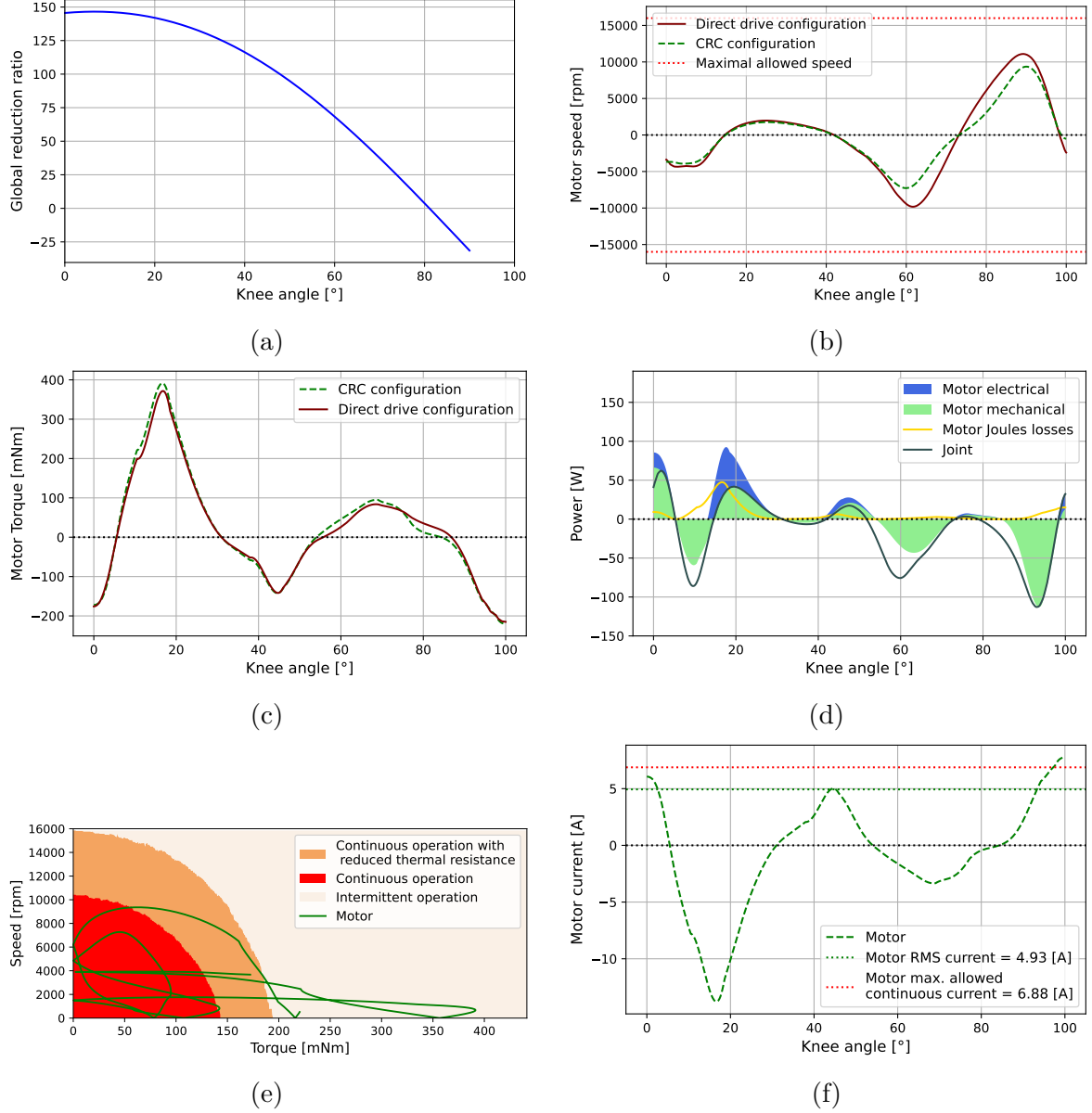


Figure 6.13: Walking profiles for the CRC configuration embedding a Faulhaber BP3274 motor optimized for walking only (a) global reduction ratio (b) motor speed (c) motor torque (d) motor powers (e) motor speed-torque operating range (f) motor current

Exactly as done in section 5.6, considering a six-cell 24[V] lithium polymer battery with a 2400 [mAh] capacity (mass around 400 [g]), the autonomy of an active prosthesis incorporating a CRC mechanism driven by a Faulhaber BP3274 motor can be roughly evaluated as follows:

1. Total energy stored in the battery:

$$24 \text{ [V]} \cdot 2.4 \text{ [Ah]} = 57.6 \text{ [Wh]} \quad (6.28)$$

$$57.6 \text{ [Wh]} \cdot 3600 \left[\frac{\text{J}}{\text{Wh}} \right] = 207\,360 \text{ [J]} \quad (6.29)$$

2. Number of walking cycles:

$$\frac{207\,360 \text{ [J]}}{(14.2 + 2) \left[\frac{\text{J}}{\text{cycle}} \right]} = 12\,800 \text{ [cycle]} \quad (6.30)$$

3. Autonomy:

$$12\,800 \text{ [cycle]} \cdot 1 \left[\frac{\text{s}}{\text{cycle}} \right] = 12\,800 \text{ [s]} \approx 3.5 \text{ [h]} \quad (6.31)$$

The optimized combination could provide up to 12 800 strides and 25 600 steps which exceed the 1500 steps that a transfemoral amputee takes on average per day. However, this corresponds to a battery autonomy of 3.5 hours falling outside the targetted range defined in the specifications (reminder: 5-8 [h]). Larger batteries should be considered to reach the specified autonomy.

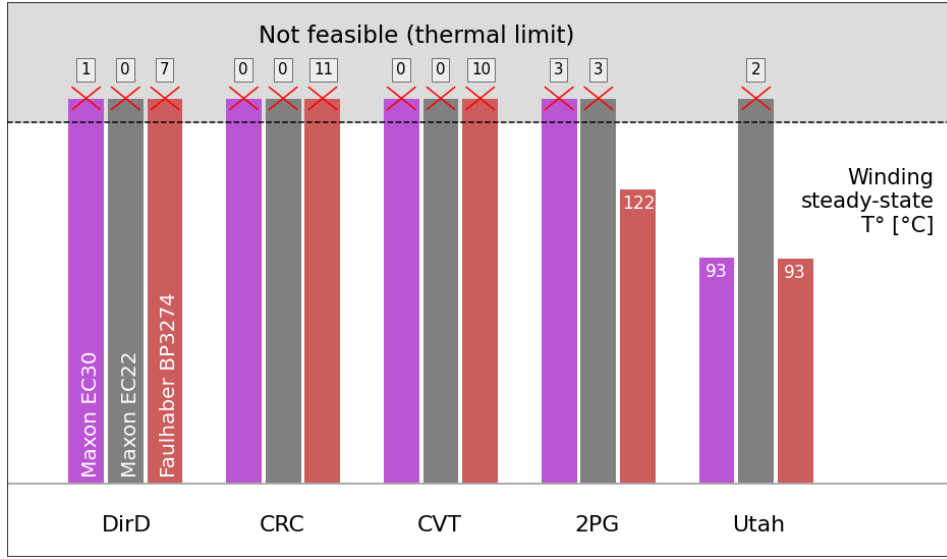
6.4.2 Combined walking and stairs ascent optimization

First, when considering an active task like stairs ascent, the motor's thermal behavior must be assessed to prevent winding overheating during prolonged activity. Figure 6.14 illustrates the results from the thermal model. Motor and variable transmission combinations that exceed the thermal limit are marked with a red cross topped by the number of cycles achievable before reaching this limit, which corresponds to the maximum allowable winding temperature of around 150-155 [°C] for all three motors.

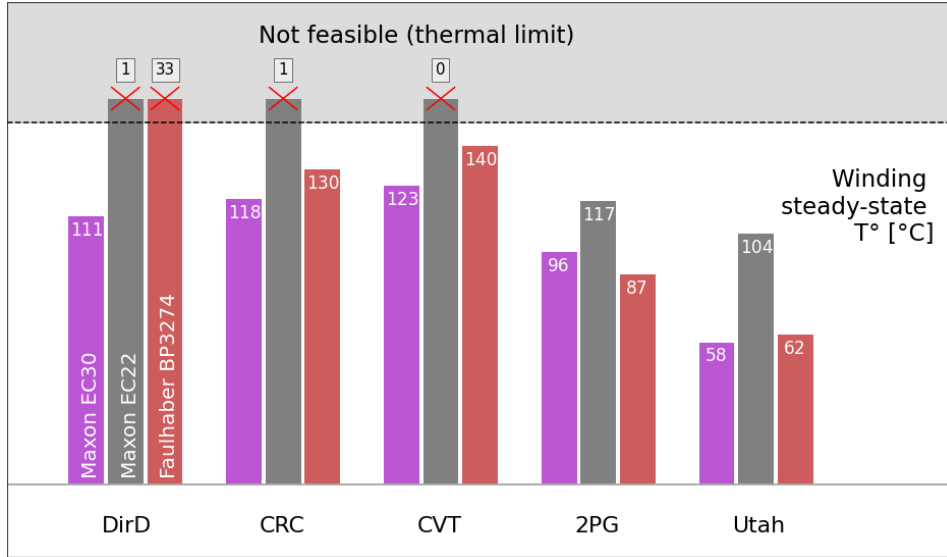
The other combinations can theoretically perform stairs ascent indefinitely without overheating and are represented in the graph together with their winding steady-state temperature, or the temperature reached at thermal equilibrium after several time constants τ_w .

Some combinations do not overheat even when the motor operates without additional cooling (non-reduced thermal resistance R_{th2}), as shown in Figure 6.14a. This was not observed with the elastic actuator configurations. Moreover, these configurations could not power stairs ascent using a Maxon EC22 motor, even with a reduced thermal resistance R_{th2} .

In contrast, Figure 6.14b shows that **the 2PG concept and the Utah bionic leg can power stairs ascent with a Maxon EC22 motor without overheating**. In other words, these are the only configurations considered in this work that allow the use of the smallest of the three candidate motors without compromising the scope of tasks that can be safely performed.



(a)



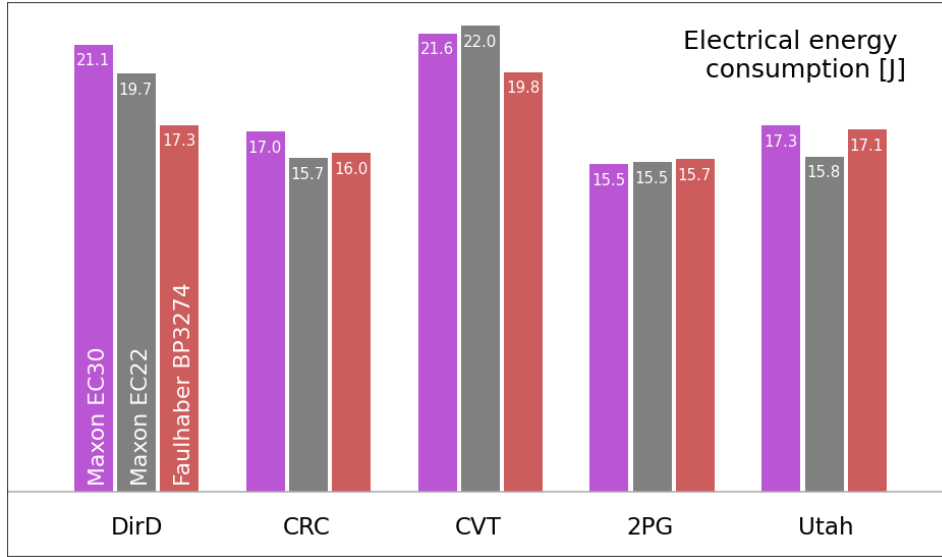
(b)

Figure 6.14: Comparison of the steady-state winding temperature reached at thermal equilibrium during sustained stairs ascent between direct drive and all four variable transmission concepts optimized simultaneously for walking and stairs ascent. The red crosses indicate that the temperature exceeds the maximum allowed temperature in the winding, and the framed numbers above them correspond to the number of stairs ascent cycles that can be performed before reaching overheating. (a) Normal housing-ambient thermal resistance (b) Reduced housing-ambient thermal resistance ($R_{th2} - 50\%$)

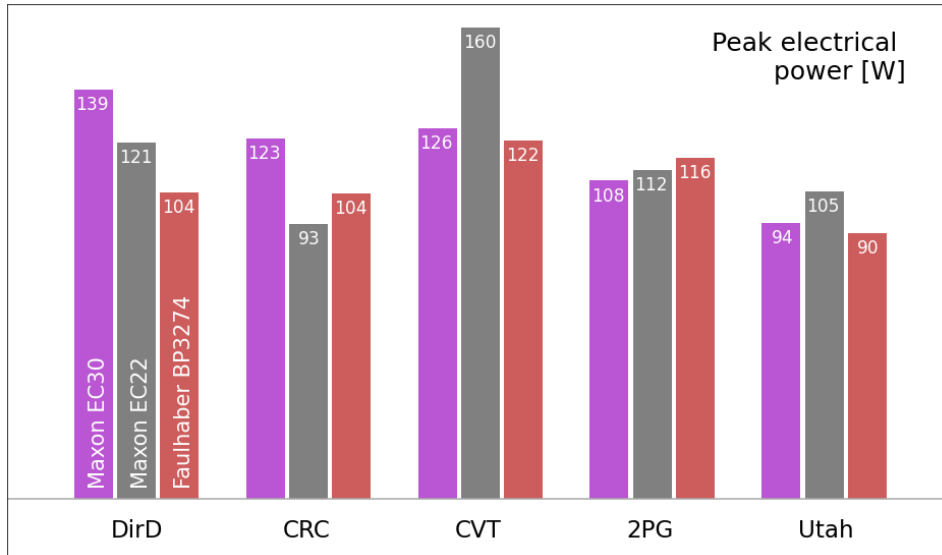
Since the reduction of the motor size and weight was the primary motivation underlying the investigation of variable transmissions, the 2PG and Utah bionic leg equipped with a Maxon EC22 motor are identified as the best candidates to perform various lower limb tasks. To choose between them, one could consider looking at the electrical energy consumption, especially during walking which is performed far more frequently than stairs ascent. Nevertheless, as depicted in Figure 6.15a and Figure 6.16a, the 2 concepts yield very similar results regarding energy efficiency:

- Walking: 15.5 [J] for 2PG and 15.8 [J] for Utah, leading to a gain of around 20% compared to direct drive (19.7 [J])
- Stairs ascent: 132 [J] for 2PG and 119 [J] for Utah, representing a gain of around 30% relative to direct drive (180 [J])

The small differences in terms of the objective functions do not allow to definitively reject one solution over the other, so both options are retained at this point.

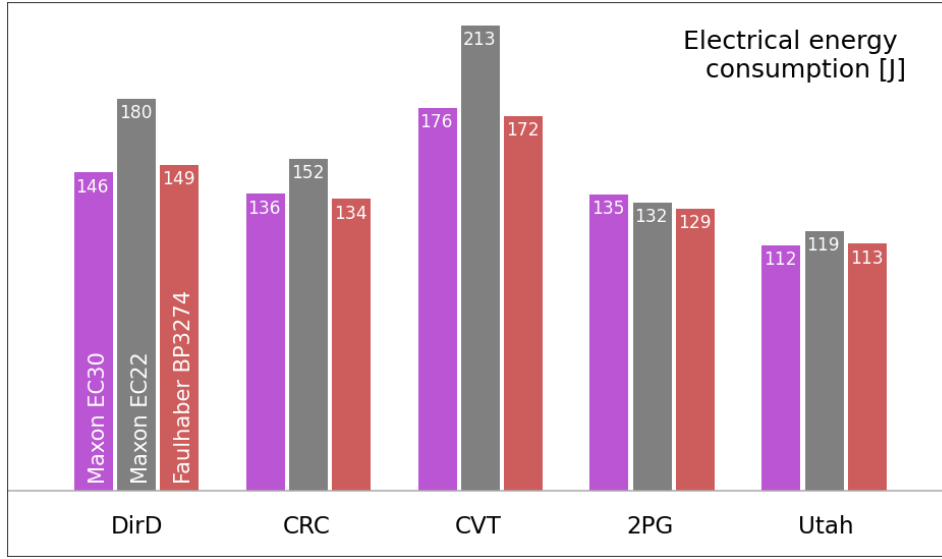


(a)

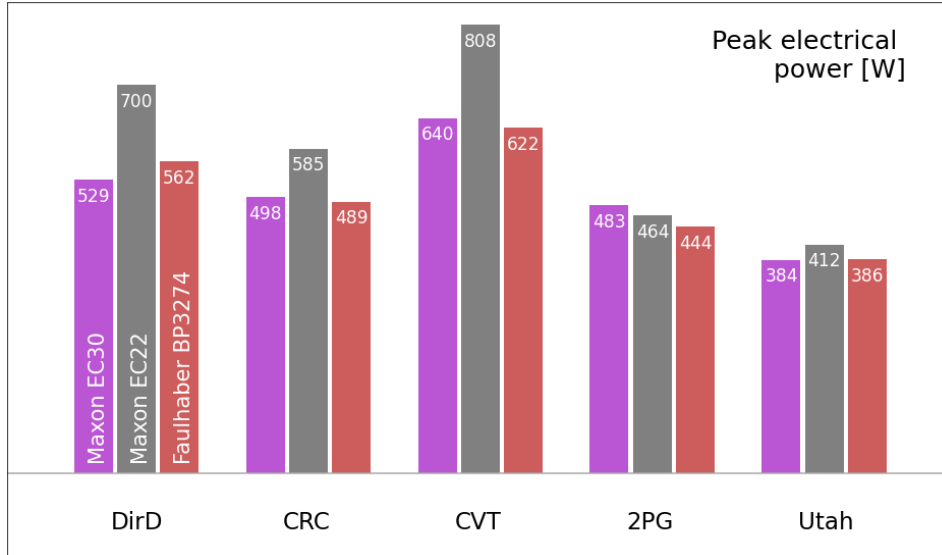


(b)

Figure 6.15: Comparison of walking performance between direct drive and all four variable transmission concepts optimized simultaneously for walking and stairs ascent (a) Electrical energy consumed by the motor over one walking cycle (b) Peak motor electrical power during walking



(a)



(b)

Figure 6.16: Comparison of stairs ascent performance between direct drive and all four variable transmission concepts optimized simultaneously for walking and stairs ascent (a) Electrical energy consumed by the motor over one stairs ascent cycle (b) Peak motor electrical power during stairs ascent

Figure 6.17 displays the advantages inherent to the 2PG concept optimized for both walking and stairs ascent (optimal parameters values can be found in Table 6.7), given the different profile associated to the walking cycle.

As already explained in subsection 6.2.2, above a certain value of the GRF, the transmission switches for the entire stance phase to its high reduction ratio, which is almost triple the low reduction ratio associated with the swing phase (Figure 6.17a). This adaptation flattens the motor speed and torque compared to a direct drive configuration with a high fixed reduction ratio, as shown in Figure 6.17b and Figure 6.17c, consequently narrowing the speed-torque operating range (Figure 6.17e).

The reduced motor torque during the stance phase allows the motor to draw less current, thereby lowering Joule's losses, as indicated by the yellow curve in Figure 6.17d. Furthermore, Figure 6.17f demonstrates that the motor's RMS current is inferior by 9% to its nominal value.

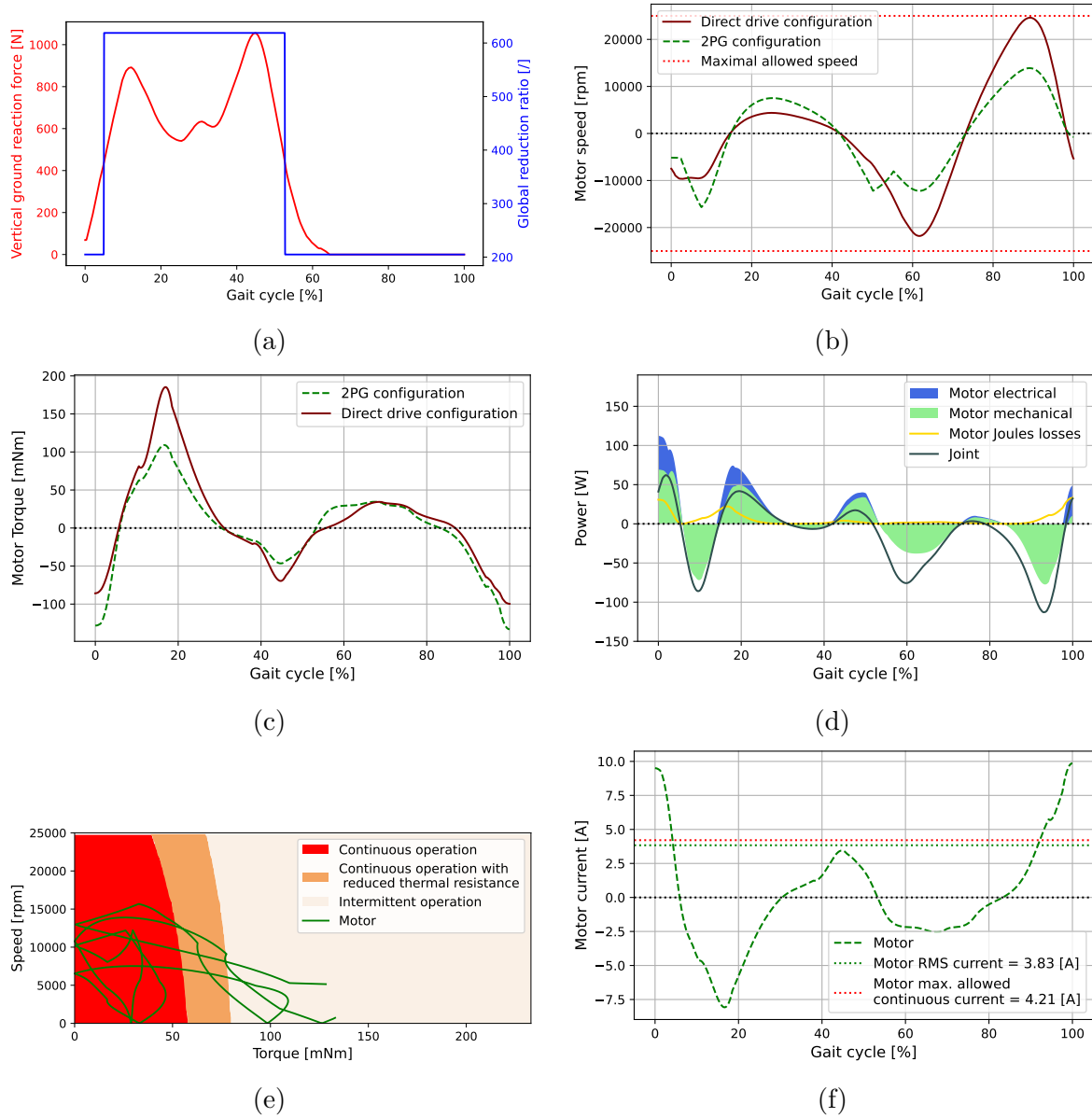


Figure 6.17: Walking profiles for the 2PG configuration embedding a Maxon EC22 motor optimized simultaneously for walking and stairs ascent (a) vertical GRF (in red) & 2PG global reduction ratio (in blue) (b) motor speed (c) motor torque (d) motor powers (e) motor speed-torque operating range (f) motor current

Z_r/Z_{s1} [-]	Z_r/Z_{s2} [-]	k [N/m]	r_i [cm]	Δr [cm]
6.7	2.6	1243	1.8	0.8

Table 6.7: Parameters for the 2PG concept embedding a Maxon EC22 motor optimized for both walking and stairs ascent

Figure 6.18 along with Table 6.8 show the same information as for the 2PG concept but for the Utah bionic leg and given the example of stairs ascent. In Figure 6.18c, the significant reduction in torque during the stance phase, attributable to the high reduction ratio, is particularly visible. This reduction leads to decreased power losses and consequently lowers the motor's electrical power requirements, as shown in Figure 6.18d.

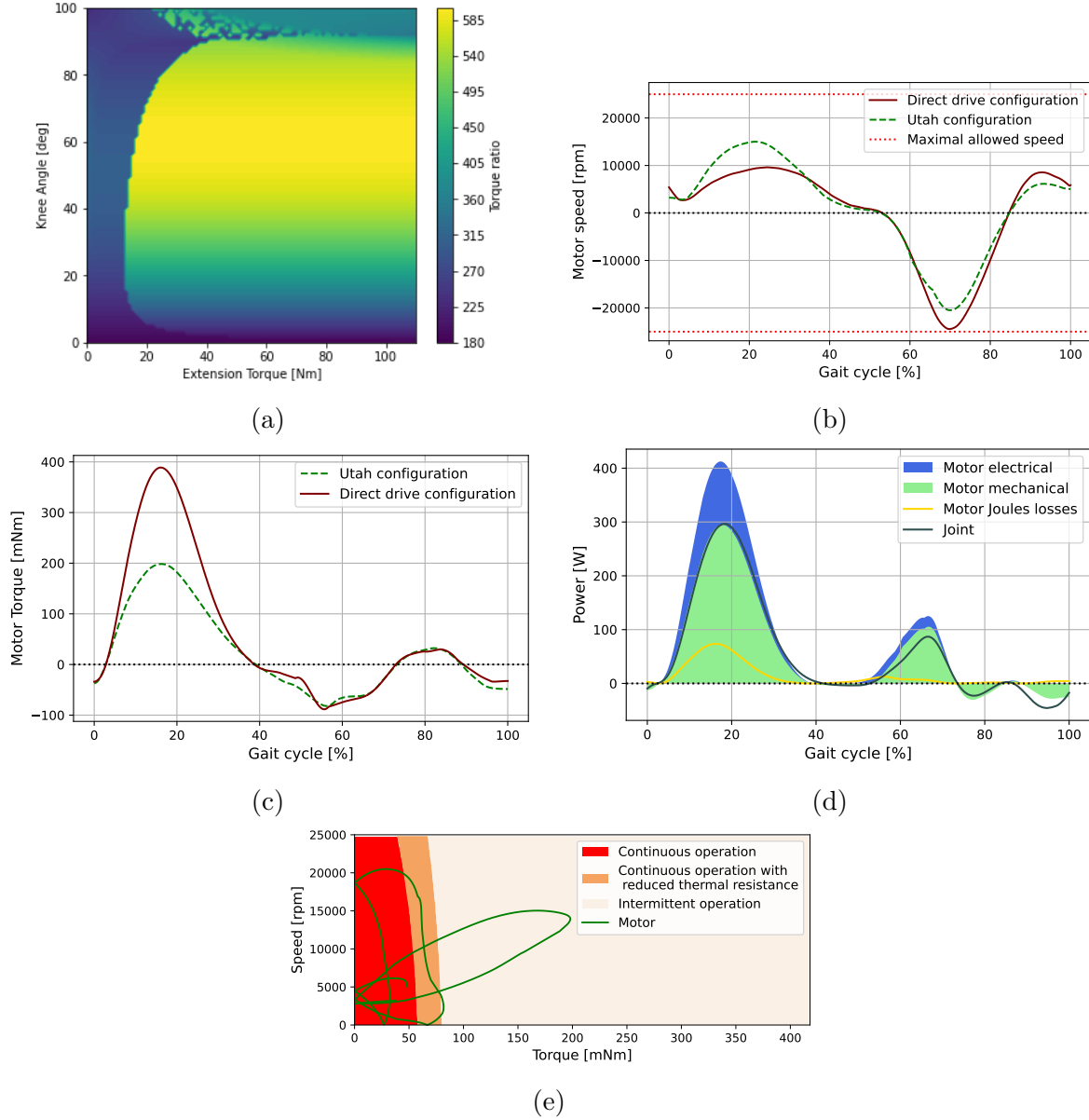


Figure 6.18: Stairs ascent profiles for the Utah bionic leg embedding a Maxon EC22 motor optimized simultaneously for walking and stairs ascent (a) Knee torque ratio as a function of the knee angle and torque (b) motor speed (c) motor torque (d) motor powers (e) motor speed-torque operating range

k [N/mm]	F_{s0} [N]	$\delta_{2,h}$ [mm]	b [cm]	N_g [/]
44	142.2	47	14.1	3.7

Table 6.8: Parameters for the Utah bionic leg embedding a Maxon EC22 motor optimized for both walking and stairs ascent

The autonomy of the 2PG concept and the Utah bionic leg equipped with a Maxon EC22 can be approximately evaluated with the same model as previously, based on the 2400 [mAh] capacity of a six-cell 24[V] lithium polymer battery (mass around 400 [g]). Still considering 2 [W] of electrical power for the electronic system, the total electrical energy consumptions are:

- $\begin{cases} \text{2PG: } 15.5 \text{ [J]} + 2 \text{ [W]} \cdot 1 \text{ [s]} = 17.5 \text{ [J]} & \text{per walking cycle} \\ \text{Utah: } 15.8 \text{ [J]} + 2 \text{ [W]} \cdot 1 \text{ [s]} = 17.8 \text{ [J]} & \text{per walking cycle} \end{cases}$
- $\begin{cases} \text{2PG: } 132 \text{ [J]} + 2 \text{ [W]} \cdot 1.4 \text{ [s]} = 134.8 \text{ [J]} & \text{per stairs ascent cycle} \\ \text{Utah: } 119 \text{ [J]} + 2 \text{ [W]} \cdot 1.4 \text{ [s]} = 121.8 \text{ [J]} & \text{per stairs ascent cycle} \end{cases}$

Assuming that a transfemoral amputee may climb up 100 stairs (so 50 stairs ascent cycle) with one battery recharge, the number of walking cycles that can be performed is evaluated as:

$$\begin{cases} \text{2PG: } \frac{207\,360 \text{ [J]} - 50 \cdot 134.8 \text{ [J]}}{17.5 \left[\frac{\text{J}}{\text{cycle}} \right]} = 11\,460 \text{ [cycle]} \\ \text{Utah: } \frac{207\,360 \text{ [J]} - 50 \cdot 121.8 \text{ [J]}}{17.8 \left[\frac{\text{J}}{\text{cycle}} \right]} = 11\,300 \text{ [cycle]} \end{cases} \quad (6.32)$$

which corresponds to more than 11 000 strides (22 000) steps for both concepts, which significantly exceeds the need of an amputee (around 1500 steps on average). This leads to a battery autonomy of only 3 hours that does not respect the specifications (reminder: 5-8 [h]). A larger battery is required to fall within the specified range.

6.5 Conclusion

This chapter examined the integration of variable transmissions into active knee prosthetic devices, aiming at minimizing motor size and weight by reducing speed and torque demands on it, while still replicating the large biological range of speed and torque required at the joint. The key findings are summarized below, reflecting on the optimal kinematic chains identified for each type of optimization.

1. Walking Optimization:

- When optimizing exclusively for walking, variable transmissions whose reduction ratio adapts based on the knee joint angular position perform better
- The optimal configuration is given by the connecting rod-crank (CRC) mechanism powered by a Faulhaber BP3274 motor
- This configuration results in an electrical energy consumption of approximately 14.2 [J] per walking cycle, a reduction of 17.9% from the 17.3 [J] required for a direct drive.
- The peak electrical power is reduced by 10.6%, from 104 [W] in direct drive to 93 [W].

2. Simultaneous Optimization for Walking and Stairs Ascent:

- When optimizing for both walking and stairs ascent, variable transmissions taking the ground reaction force or the torque applied at the knee joint as input variables to adjust their reduction ratio perform better.
- The optimal configurations are the 2-stage planetary gear (2PG) and the Utah bionic leg, which are the only ones capable of operating with the smallest candidate motor, the Maxon EC22, without overheating during stairs ascent (under the condition of a reduced housing-ambient thermal resistance).
- The electrical energy consumption is approximately 15.5 [J] for 2PG and 15.8 [J] for Utah per walking cycle, a reduction of around 20% compared to direct drive (19.7 [J])
- For a stairs ascent cycle, the electrical energy consumption is around 132 [J] for 2PG and 119 [J] for Utah, representing a gain of about 30% relative to direct drive (180 [J]).

The comprehensive analysis presented here confirms the potential of variable transmissions to power active knee prostheses with smaller and lighter motors, which is highly desirable for all the reasons listed in the introduction of this chapter. It was demonstrated that inserting transmissions whose reduction ratio can be adjusted along the gait cycle successfully reduces the speed and torque requirements on the motor. It enables the selection of motors with smaller continuous operating ranges without sacrificing the scope of achievable lower limb tasks or risking thermal failure during sustained, torque-demanding activities.

The results reported above are once again acknowledged as debatable. However, this chapter emphasized the prosthesis weight constraint along with the minimization of the objective functions, unlike chapter 5. The complexity of implementation has not been considered here but will be addressed in the next chapter.

Chapter 7

Global comparison

This chapter aims to compare the best configurations identified for each category of kinematic chains examined in chapter 5 and chapter 6. The comparison focuses on the objective functions and thermal behavior, but also on the estimated weight and complexity of implementation. A weighted cost function based on these criteria is computed for the variant designs to select the optimized kinematic chain which provides the best balance between performance, weight, and feasibility.

7.1 General discussion

The optimization for each category of kinematic chains yielded different conclusions. Elastic actuator configurations showed their potential to significantly reduce the motor electrical energy consumption during walking. Among all elastic actuators investigated, clutchable configurations stand out, especially the 2PEA which exhibits the best performance when the parameters are optimized exclusively for walking (see Figure 5.18), and the CSEA which provides the best trade-off to perform various lower limb tasks (see Figure 5.21 and Figure 5.22).

On the other hand, variable transmissions are less efficient at reducing electrical energy consumption compared to elastic actuators but allow to adapt speed and torque requirements on the motor, enabling the use of smaller motors to power the prosthesis. Notably, the 2PG concept and the Utah bionic leg embedding the smallest candidate motor—the Maxon EC22—can operate indefinitely without overheating during stair ascent, which is not observed for any other configuration.

From this analysis, it appears that combining both categories would be advantageous, leveraging the benefits of each. The proposed kinematic chain would include an elastic actuator configuration whose reduction ratio is rendered by a variable transmission. This approach aligns with two primary objectives outlined in the project specifications: extending the autonomy of the active prosthesis (required range: 6 - 9 [h]) by minimizing the motor electrical energy consumption during the most frequent activity (i.e., walking), and striving towards meeting the weight constraint (required range: 2.5 - 2.7 [kg]) by enabling the use of a small motor, specifically the Maxon EC22.

It could be argued that adding a variable transmission to the elastic actuator configuration would increase the weight of the kinematic chain. However, whether variable or not, a transmission with a high reduction ratio is needed to relate the speed at the knee joint

(up to 400 [°/s] $\approx$ 70 [rpm] as shown in Figure 3.10c) with the motor speed (up to 25 000 [rpm] for a Maxon EC22). Additionally, the variable transmission concepts retained from chapter 6 are lightweight. Indeed, planetary gears are known for their compact and light construction, and the Utah bionic leg was developed with the weight as one of their most important design objectives.

7.2 Final configuration

With either the 2PG transmission or the Utah bionic leg, the Maxon EC22 motor has proven capable of powering the prosthesis for various tasks, making it the preferred choice in the final configuration due to its lower weight compared to other candidate motors. For the rest of the kinematic chain, the elastic actuators 2PEA and CSEA, as well as the variable transmissions 2PG and the Utah bionic leg, have been identified as the best configurations in their respective categories. The four possible combinations derived from these configurations will be compared based on their ability to minimize the following weighted cost function C_{tot} :

$$C_{tot} = w_{E_e} \cdot C_{E_e} + w_{th} \cdot C_{th} + w_w \cdot C_w + w_{imp} \cdot C_{imp} \quad (7.1)$$

Criteria	Weights
C_{E_e} : Walking electrical energy consumption	$w_{E_e} = 3$
C_{th} : Thermal behavior during stairs ascent	$w_{th} = 2$
C_w : Estimated weight	$w_w = 1$
C_{imp} : Implementation complexity	$w_{imp} = 1$

Table 7.1: Evaluated criteria and their weight in the cost function

To compare the energetic and thermal performance of the combinations, the scores C_{E_e} and C_{th} are scaled between 0 and 1 by performing a *min-max* normalization, i.e.:

$$C_{E_e} = \frac{E_e - \min(E_e)}{\max(E_e) - \min(E_e)} \quad \text{and} \quad C_{th} = \frac{T_{w,\infty} - \min(T_{w,\infty})}{\max(T_{w,\infty}) - \min(T_{w,\infty})} \quad (7.2)$$

where E_e is the motor electrical energy consumption over one walking cycle, associated with the highest weight ($w_{E_e} = 3$) since the energetic efficiency of the prosthesis is the main design objective in this work. $T_{w,\infty}$ is the steady-state winding temperature reached at thermal equilibrium during sustained stairs ascent. The closer it gets to the maximum allowed temperature of 155 [°C] in the winding, the higher C_{th} gets. The associated weight of the thermal criterion is $w_{th} = 2$ since the temperature reached at thermal equilibrium should maintain a certain margin below 155 [°C], allowing for some relaxation of the 50% reduction in thermal resistance considered up to now, as will be discussed later in this chapter. The scores for the remaining two criteria, the weight C_w and the implementation complexity C_{imp} are estimated between 0 and 1, based on global considerations. This may be less accurate, therefore these criteria are assigned a unit weight.

Table 7.2 displays the score of each criterion for the four combinations of elastic actuators and variable transmissions. Taking into account their weight, the scores are then added to obtain the value of the cost function through Equation 7.1.

	2PG concept	Utah Bionic leg
2PEA	$E_e = 4.6 \text{ [J]} \rightarrow C_{E_e} = 1.0$ $T_{w,\infty} = 119 \text{ [}^\circ\text{C]} \rightarrow C_{th} = 0.35$ $C_w = 0.6$; $C_{imp} = 0.8$ $\Rightarrow C_{tot} = 5.1$	$E_e = 2.9 \text{ [J]} \rightarrow C_{E_e} = 0.0$ $T_{w,\infty} = 120 \text{ [}^\circ\text{C]} \rightarrow C_{th} = 0.37$ $C_w = 0.8$; $C_{imp} = 0.4$ $\Rightarrow C_{tot} = 1.94$
CSEA	$E_e = 3.2 \text{ [J]} \rightarrow C_{E_e} = 0.18$ $T_{w,\infty} = 101 \text{ [}^\circ\text{C]} \rightarrow C_{th} = 0.0$ $C_w = 0.2$; $C_{imp} = 0.6$ $\Rightarrow C_{tot} = 1.34$	$E_e = 3.0 \text{ [J]} \rightarrow C_{E_e} = 0.06$ $T_{w,\infty} = 152 \text{ [}^\circ\text{C]} \rightarrow C_{th} = 1.0$ $C_w = 0.4$; $C_{imp} = 0.2$ $\Rightarrow C_{tot} = 2.78$

Table 7.2: Comparison of the 4 combinations (elastic actuator - variable transmission) based on the criteria scores and the weighted cost function. Minimal and maximal values are represented in green and red, respectively.

Walking electrical energy consumption

To obtain the values displayed in Table 7.2, the following assumptions are made:

- 2PEA: The two parallel springs are fully disengaged from the kinematic chain during stairs ascent. Indeed, compared to the other elastic actuators, this configuration exhibits the best energetic efficiency when optimized exclusively for walking but the results degrade significantly when other tasks such as stairs ascent are taken into account, as detailed in section 5.6).
- CSEA: The clutch is never engaged when stairs ascent is performed, since bypassing the motor during this task leads to under-extension during the forward continuance phase, failing to faithfully replicate the knee kinematics as observable in Figure 5.24a.

The energy consumption over a walking cycle for a direct drive configuration equipped with a Maxon EC22 motor is 19.7 [J]. For the 2PEA alone, it is 4.6 [J] and for the CSEA alone, it is 4.8 [J] (see Figure 5.18a). Adding a variable transmission has helped to further reduce or at least preserve the consumption of electrical energy over one cycle. The least effective combination regarding this criterion is the 2PEA-2PG, with 4.6 [J] of energy consumption, a reduction of 76.6% relative to direct drive. The most energy-efficient combination is the 2PEA-Utah, with 2.9 [J], representing a reduction of 85.3% compared to direct drive.

Thermal behavior during stairs ascent

The steady-state winding temperatures reached during sustained stairs ascent indicated in Table 7.2 are once again calculated considering a housing-ambient thermal resistance reduced by half. The incorporation of the 2PG or the Utah bionic leg allows to operate with the Maxon EC22 motor without ever trespassing its thermal limit. However, for the CSEA-Utah combination, the winding temperature stabilizes at thermal equilibrium around 152 [°C], which is very close to 155 [°C], making it the worst candidate regarding that criterion. The largest margin is obtained with the CSEA-2PG (winding steady-state temperature around 101 [°C]).

Estimated weight

The score C_w for the weight criterion is attributed based on 2 main global considerations. Firstly, it is estimated that the 2PEA is heavier than the CSEA configuration since it requires two springs instead of one, and that both springs in the 2PEA should be equipped with a (dis)engagement mechanism¹ compared to the CSEA that only needs one clutching mechanism. Moreover, the same reasoning is applied for the variable transmission. The 2PG concept, which only needs a 2-stage planetary gear and a gear pair, is estimated lighter than the Utah bionic leg that comprises a gear pair, a ballscrew, a linear guide, a connecting rod, and a linear spring [61]. However, the weight difference is considered smaller between the variable transmissions than between the elastic actuator configurations.

Implementation complexity

Similarly, the score C_{imp} for the implementation complexity criterion is assigned based on overall statements that distinguish the two candidates in each category. The 2PEA is considered to be more complex to implement since it requires two accurate mechanisms to (dis)engage each spring at strategic, but different, instants in the cycle while the CSEA needs one clutching mechanism which (dis)engages at the right time during walking and never engages during stairs ascent. Nevertheless, complexity difference is considered less important than for the variable transmissions. Indeed the Utah bionic leg implements its change in reduction ratio based on the knee torque and angular position, which are two input variables that *directly act* in the kinematic chain. This lowers the complexity compared to the 2PG whose change in reduction ratio relies on the ground reaction force that must be recovered in some way to actuate a translating mechanism (see Figure 6.8).

Weighted cost function

Eventually, the calculation of the weighted cost function C_{tot} allows to identify the best combination, constituting the kinematic chain final configuration: **a clutchable series elastic actuator (CSEA) configuration, whose variable reduction ratio is rendered by a 2-stage planetary gear (2PG) transmission, and equipped with a Maxon EC22 motor.**

The schematic representation of a prosthesis including a CSEA configuration and a 2PG transmission is shown in Figure 7.1 along with the optimization results in Table 7.3, i.e. the optimal parameters and the minimized objective functions for walking and stairs ascent. A DC motor, specifically a Maxon EC22, drives a gearhead with a reduction ratio of 5 that provides a part of the fixed gearhead ratio N_g whose optimal value is 19. From the output of the gearhead, the rest of the kinematic chain is aligned and begins with the 2PG transmission.

The optimal teeth ratios for the 2PG are 7.6 for the first stage, and 2.4 for the second stage. For example, if the ring gear has $Z_r = 60$ teeth, the first and second sun gears need to have respectively $Z_{s_1} = 8$ teeth and $Z_{s_2} = 25$ teeth. Given these values, the global

¹Regarding the assumption made above: The two parallel springs are fully disengaged during stairs ascent

reduction ratio varies between 210 in the low reduction ratio state of the 2PG and 580 in its high reduction ratio state, according to Equation 6.24 and Equation 6.25.

The change from one reduction ratio to the other is governed by the other optimized parameters and is shown in Figure 7.2a for walking and Figure 7.3a for stairs ascent, together with the evolution of the vGRF during those tasks. The principle of the braking mechanism used to vary the reduction ratio is detailed in subsection 6.2.2. In order to have enough friction surface, the outer radius must be $r_o = r_i + \Delta r = 2.5$ [cm], i.e. a 5 [cm] diameter for the 2PG.

The output of the 2PG drives the torsional series spring, whose optimal stiffness approximates the linear torque-angle relationship characteristic of the walking ESF phase, as visible in Figure 7.2c. This allows the spring to act as a mechanical energy buffer capable of rendering all power fluctuations during that phase (see Figure 7.2d). Thanks to this mechanical advantage, the motor can be bypassed during ESF.

Lastly, the remaining part of the fixed ratio N_g is provided by a bevel gear with a reduction ratio of 3.8 ($(N_g = 19)/5 = 3.8$). This gear transforms the rotation in the transverse plane into a rotation at the knee joint, in the sagittal plane.

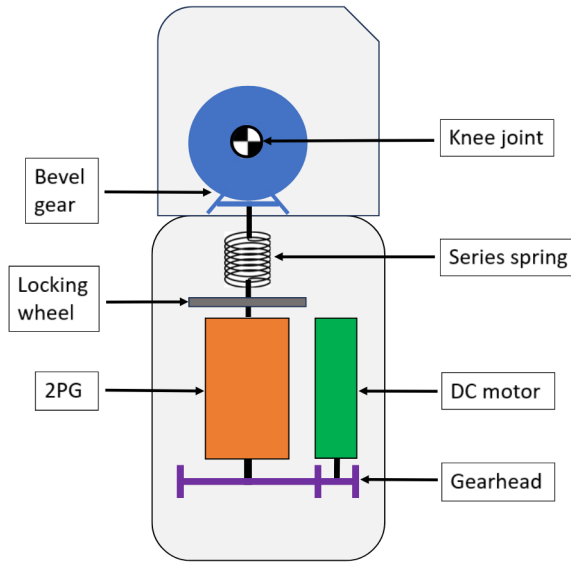


Figure 7.1: Scheme of a prosthesis incorporating a CSEA-2PG (Sagittal plane)

k_s [Nm/rad]	302
$\mathbf{W} \beta_{en}$ [%]	6.4
$\mathbf{W} \beta_{disen}$ [%]	51.1
Z_r/Z_{s1} [-]	7.6
Z_r/Z_{s2} [-]	2.4
k [N/m]	1270
r_i [cm]	1.8
Δr [cm]	0.7
N_g [-]	19
$\mathbf{W} \left\{ \begin{array}{l} E_e \text{ [J]} \\ \max(P_e) \text{ [W]} \end{array} \right.$	 3.2 82
$\mathbf{SA} \left\{ \begin{array}{l} E_e \text{ [J]} \\ \max(P_e) \text{ [W]} \end{array} \right.$	 101 444

Table 7.3: Optimization results. In order: optimal values of the CSEA parameters, 2PG parameters, and objective functions

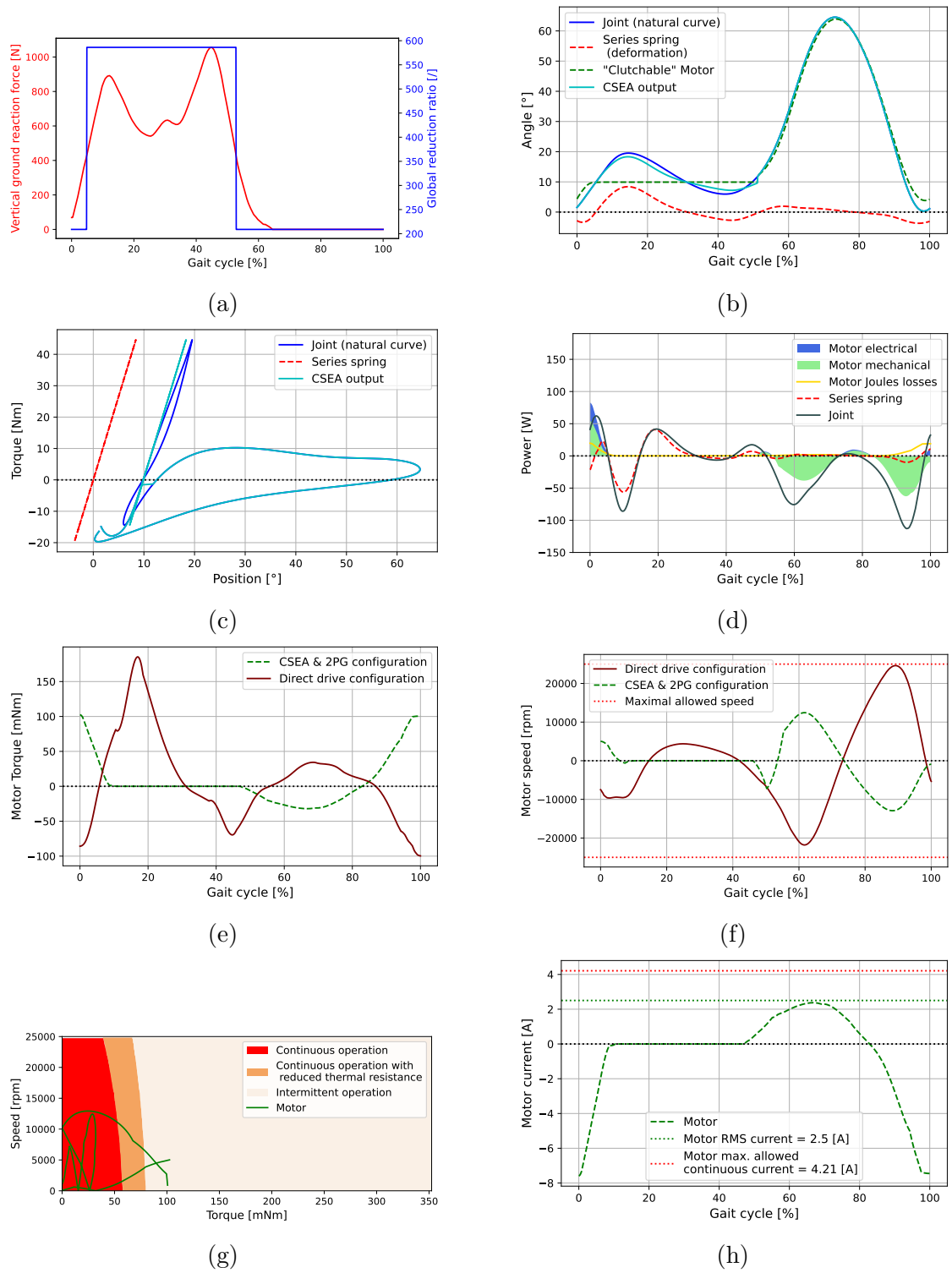


Figure 7.2: Walking profiles for the CSEA-2PG configuration embedding a Maxon EC22 motor optimized for both walking and stairs ascent (a) vGRF (in red) and global reduction ratio (in blue) (b) position (c) torque-angle (d) powers (e) motor torque (f) motor speed (g) motor speed-torque operating range (h) motor current

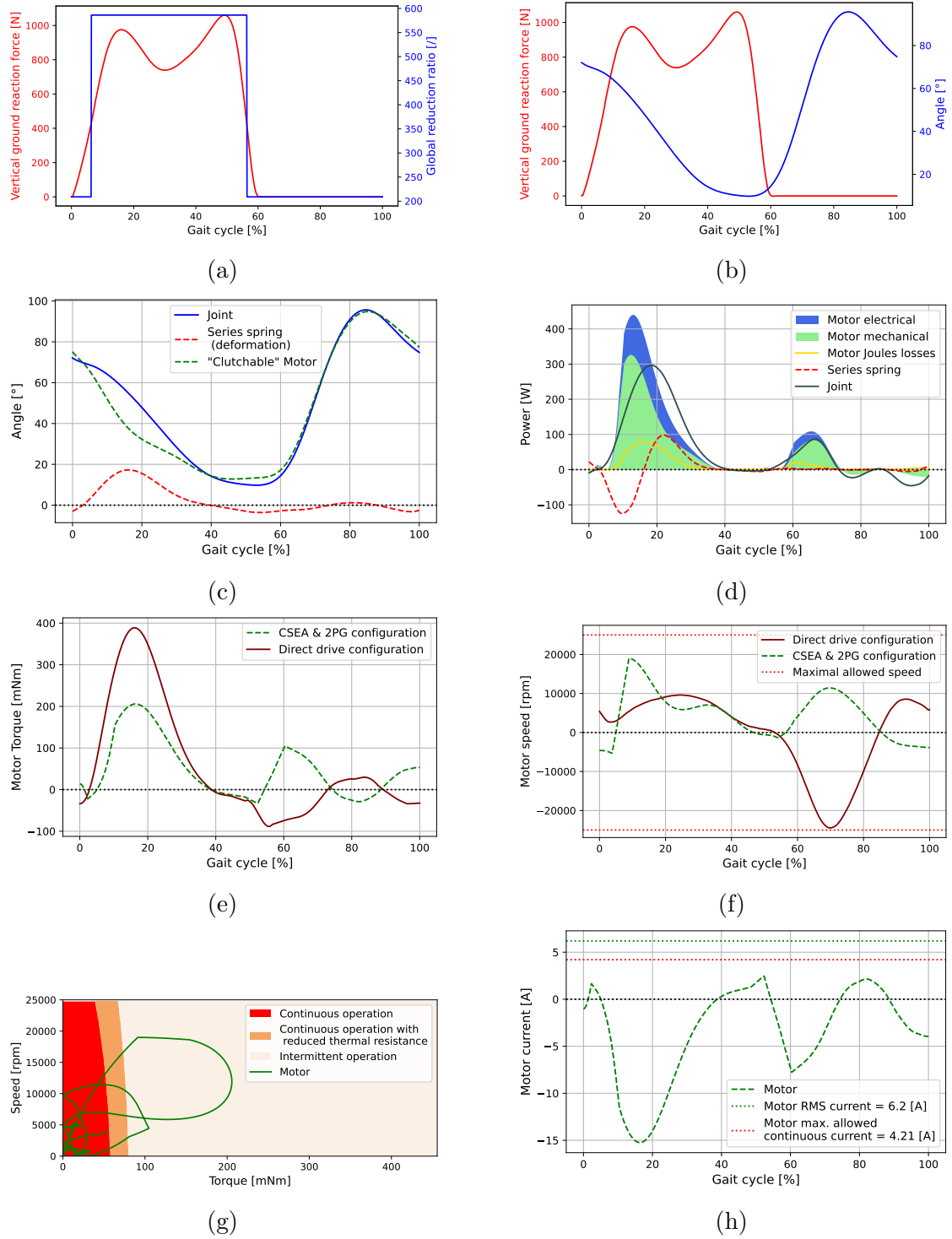


Figure 7.3: Stairs ascent profiles for the CSEA-2PG configuration embedding a Maxon EC22 motor optimized for both walking and stairs ascent (a) vGRF (in red) and global reduction ratio (in blue) (b) vGRF (in red) and knee joint angle (in blue) (c) position (d) powers (e) motor torque (f) motor speed (g) motor speed-torque operating range (h) motor current

Table 7.3 gathers the minimized values of the objective functions. The gains in terms of motor electrical energy consumption and peak electrical power for the final configuration (CSEA-2PG with Maxon EC22) compared to a direct drive configuration equipped with a

Maxon EC22 motor are the following:

- For walking, the CSEA-2PG reaches an energy consumption per cycle of 3.2 [J], a 33.3% decrease from the 4.8 [J] for the CSEA alone, and a 83.7% decrease from the 19.7 [J] in direct drive configuration. The peak power is reduced by 32.2%, from 121 [W] in direct drive to 82 [W] for the CSEA-2PG.
- For stairs ascent, the CSEA-2PG energy consumption is approximately 101 [J] per cycle, a reduction of 43.9% from the 180 [J] required for a direct drive. The peak power is reduced by 36.6%, from 700 [W] in direct drive to 444 [W] for the CSEA-2PG.

These gains can be visualized in Figure 7.2 and Figure 7.3. For walking, Figure 7.2d highlights the mechanical advantage of the series spring, which absorbs and then releases energy during ESF, allowing the bypass of the motor and the reduction of its power demand.

For stairs ascent, the substantial reductions mentioned above are attributable to the 2PG rather than the CSEA since the clutch is never engaged during this task, and the spring stiffness is closer to the optimum for walking. As seen in Figure 7.3f, the change of reduction ratio adapts the motor speed requirements between the stance and swing phase but most importantly, the high reduction ratio reached during the stance phase (when the bodyweight is applied on the prosthesis) almost halves the peak-torque requirement (from approximately 400 [mNm] to 200 [mNm]) compared to a direct drive configuration with a fixed reduction ratio (see Figure 7.3e). As visible in Figure 7.3d, this leads to low Joule's losses since the current flowing through the motor is reduced, contributing to the reduction of electrical energy consumption and peak electrical power during stairs ascent.

The autonomy of the CSEA-2PG equipped with a Maxon EC22 can be approximately evaluated with the same model as in the previous chapters, still considering 2 [W] of electrical power for the embedded electronic system. This yields electrical energy consumption of $3.2 \text{ [J]} + 2 \text{ [W]} \cdot 1 \text{ [s]} = 5.2 \text{ [J]}$ per walking cycle, and $101 \text{ [J]} + 2 \text{ [W]} \cdot 1.4 \text{ [s]} = 103.8 \text{ [J]}$ per stairs ascent cycle.

Striving towards meeting the weight constraint, a small and lightweight battery is chosen to supply the motor, namely a six-cell 24[V] lithium polymer battery with a 1200 [mAh] capacity (mass around 160 [g]). Assuming again that a transfemoral amputee may climb up 100 stairs (50 stairs ascent cycle) with one battery recharge, the number of walking cycles that can be performed is evaluated as:

$$\frac{103\,680 \text{ [J]} - 50 \cdot 103.8 \text{ [J]}}{5.2 \left[\frac{\text{J}}{\text{cycle}} \right]} = 18\,940 \text{ [cycle]} \quad (7.3)$$

This corresponds to nearly 19 000 strides, or 38 000 steps (largely exceeding the average 1500 steps of an amputee), leading to a 5 [h] 15 [min] autonomy falling within the specified range (5-8 [h]). The incorporation of the 2PG with the CSEA configuration allows to reduce the capacity of the battery compared to the CSEA alone (see subsection 5.6.2), while still respecting the project specifications. Furthermore, the results obtained here are consistent with the 30 000 walking strides achievable by the state-of-art CSEA prosthesis [55], that does not include a variable transmission but features regeneration of negative electrical power.

7.3 Overall limitations

To end up with the final configuration described in this chapter, a certain number of hypotheses had to be made throughout the analysis and comparison of all the concepts.

First, using biomechanical data from a sound leg for simulations provides a strong foundation but could be criticized, as already mentioned in section 5.1. Indeed, transfemoral amputees may intentionally deviate from biological curves, such as by relying more on the intact leg or residual hip, leading to pathological gait.

Besides, the active knee prosthesis designed in this work aims to replicate the characteristics of a biological gait by incorporating a motor, notably to perform early-stance flexion. However, this device may not be suitable for all transfemoral amputees, particularly those with functional classification levels below K3 (see subsection 2.1.2).

Moreover, to derive the simulation models, several assumptions were made. Due to uncertainties about the specific parts to be used within the kinematic chain, general design approximations were adopted to estimate variables such as the mechanical efficiency or the weight of some mechanical parts. These inaccuracies may lead the optimization results to deviate from the actual performance of a working prototype.

It is also acknowledged that the passive activation mechanisms in the final configuration (i.e. the braking system for the 2PG and the clutch mechanism for the CSEA) are still in a conceptual state and require further design steps to determine their dimensions and ensure functionality. Additionally, both mechanisms rely on the loading of the prosthesis by the body weight, but the method to capture the GRF from the ground contact was not investigated in this study, as it would likely necessitate the design of an ankle module in parallel.

Lastly, a simplified model for the thermal behavior of the motor has been considered. Up to now, a reduction by 50% of the housing-ambient thermal resistance R_{th2} and a 22 [°C] ambient temperature have been considered to evaluate the thermal performance, leading for example to a steady-state winding temperature around 101 [°C] when the CSEA-2PG performs sustained stairs ascent (dark green curve in Figure 7.4). However, these conditions may not be achievable. First, the ambient temperature can vary, and the specifications stipulate that the prosthesis must safely operate up to 40 [°C]. Moreover, to ensure that the kinematic chain is protected from the environment (specified IP value: 54), it should be enclosed in a shank frame providing structural support, safety, and protection. This could lower the ability to reduce the thermal resistance as it prevents air circulation and thus convection cooling of the motor. Nevertheless, mounting the motor on a shank frame made of metal can provide additional heat dissipation capacity.

The simulation of the motor thermal behavior indicates that the thermal resistance must

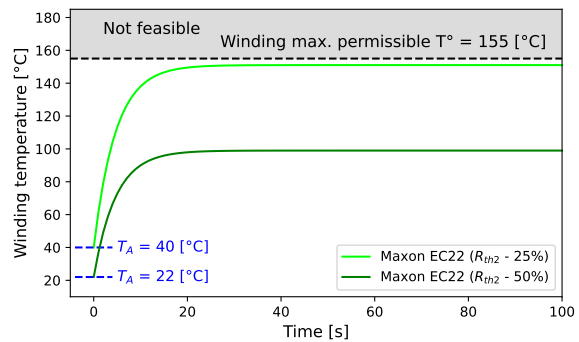


Figure 7.4: Winding temperature evolution during stairs ascent for a CSEA-2PG configuration powered by a Maxon EC22 motor

be reduced by at least 25% to keep the winding temperature below the maximum allowable limit of 155 [°C] under worst-case conditions, i.e. for a 40 [°C] ambient temperature. In this scenario, with a 25% reduction in thermal resistance, the steady-state winding temperature during sustained stair ascent reaches approximately 150 [°C] (light green curve in Figure 7.4). However, a more thorough heat transfer analysis, taking into account the metallic shank frame, is necessary to determine whether such a reduction in thermal resistance is feasible.

7.4 Additional requirements for the clutch mechanism

The 2PG transmission is designed to adjust its ratio based on the weight of the amputee on the prosthesis, generating a ground reaction force (GRF) that can be recovered to actuate the braking system detailed in subsection 6.2.2. An interesting observation can be made from Figure 7.2a and Figure 7.2b: the 2PG shifts from a low to high ratio and the CSEA clutch engages both around 5% of the walking cycle. Similarly, the 2PG switches back to a low ratio and the CSEA clutch disengages around 50% of the cycle. This indicates that the same input variable as the 2PG, namely the GRF, could be used to control the CSEA clutch mechanism. This aligns with the design objective of energetic efficiency, as the clutch could be passively actuated by the GRF.

For walking, the clutch engages and disengages at a joint angle around 10°, corresponding to the application and removal of body weight on the prosthesis. However, it should never engage during stair ascent, as previously stated. From Figure 7.3b, we observe that during stair ascent, when the GRF is applied, the knee angle is around 70°, and when it disappears, the angle is around 10°. This indicates that the clutch mechanism must differentiate between walking and other tasks, such as stair ascent, by the knee joint angle observed when body weight is first applied to the prosthesis (around 5% of the cycle). The next chapter introduces the concept of a mechanism based on the GRF and knee joint angle to implement the clutching function specific to the CSEA configuration.

Chapter 8

Design recommendations and specifications review

This chapter proposes a design for the locking wheel to achieve the mechanical combination of the two solutions assembled in the previous chapter, i.e. the CSEA and the 2PG.

8.1 Reminder and objective

As a reminder, both configurations implemented a ‘change of state’ during the cycle. The CSEA configuration blocks the motor and the 2PG (see Figure 7.1) during the stance phase of walking. During this phase, only the series spring intervenes to replicate the biological behaviour of the gait. The 2PG changes the reduction ratio between the stance and swing phases of stairs ascent. This change is achieved passively by exploiting the ground reaction force (GRF).

The aim is therefore to find a mechanism that will block the motor transmission during the stance phase of walk only. It must have no effect on the transmission chain during the swing phase of walking and during the whole stairs ascent cycle. So you need to distinguish between the stance phase and the swing phase, and between walking and climbing stairs. And this must be done passively, without a motor, using the characteristic curves of the physical variables linked to the knee joint (angle, speed, torque, vGRF) presented in chapter 3.

Analysis of the curves resulting from the simulations in chapter 5 and chapter 6 shows that the changes of state of the two solutions occur at the same time. This is an interesting feature that can be exploited.

8.2 Physical quantities

We sought to exploit the vGRF already used to change the state of the 2PG to clutch the motor during the stance phase of the walking gait cycle. The vGRF distinguishes between the stance and swing phases, since it cancels out during the swing phase. However, this is not sufficient to determine whether the patient is walking or ascending stairs, as the vGRF curves for these two tasks are very similar (see Figure 3.10 (d) and Figure 3.13 (d)). An additional variable is needed to distinguish the two tasks.

The angle makes this possible. The two angle curves are shown at Figure 8.1. The beginning of the stance phase for stairs ascent is marked by an angle of 70° which

decreases to 10° before decreasing again as the swing phase approaches. For walking, the beginning of the stance phase is marked by an angle of 0° which increases to a maximum of 20° during the Early Stance Flexion before decreasing again just before the swing phase. The angle curves do not evolve in the same way during the stance phase. The walking curve increases slightly and then decreases, whereas the stairs ascent curve begins by decreasing sharply before increasing again.

8.3 Mechanism concept: Stage 1

Observation of the curves in the previous section led to the design of the following mechanism: The locking wheel and the pawl. These are the two main parts of the mechanism. They would be made of metal, for example aluminium (used in [55] and [61]). From a geometric point of view, the locking wheel consists of two cylinders. The one with the larger diameter has a notch and a truncated circular sector and is referred to in the following as the notched-disc. The pawl takes the form of an inverted 'L' at the end of which is a rolling cylinder to enable movement on the notched-disc while limiting friction.

Initially, in order to explain the principle of the mechanism simply, the reduction ratio of the bevel gear on the Figure 7.1 is assumed to be equal to 1, so that the angle of the locking wheel is identical to that of the knee joint.

The pawl has one degree of freedom in translation in the direction of the length of the prosthesis. During the stance phase of walking or stairs ascent, vGRF is transmitted to the pawl. Under the action of the force, the pawl translates upwards. This produces the change in the reduction ratio on the 2PG as it was simulated in chapter 6 and chapter 7.

From this point on, we need to distinguish between the two tasks :

Walking:

- The rise of the pawl under the action of the reaction force takes place at small angles, from 5 to 20° at the moment of the Early Stance Flexion. At that angle range, the notched-disc is not truncated, the pawl comes into contact with the underside of the notched-disc. Thanks to the caster, it rolls with minimal friction over the surface of the notched-disc until it slides into the notch. Still under the action of the ground reaction force, it engages, blocking the rotation of the kinematic chain like a parallel key.
- At 50% of the cycle, the ground reaction force decreases as the person enters the swing phase. The pawl is no longer pushed upwards. It disengages from the notch and releases the locking wheel. It comes back down and this re-establishes the transmission of the motor to the joint.

Stairs Ascent :

- When the pawl rises under the action of the ground reaction force, the locking wheel is at an angle of 70° . Over the angular sector from 20° to 100° , the notched-disc of the locking wheel is truncated. The pawl can therefore rise higher under the action of

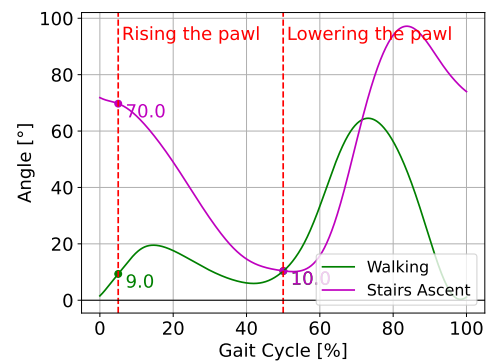


Figure 8.1: Angle curves during Walking and Stairs Ascent

the ground reaction force because it is not blocked, unlike during the stance phase of the walking gait cycle. Until the end of the stance phase, the pawl is at the level of the upper face of the notched-disc. This moves according to the angular position of the knee. There is nothing to prevent rotation during this phase, the locking wheel is free and the motor can rotate to provide the necessary power. The pawl does not interfere thanks to the caster.

- At 50% of the cycle, the ground reaction force decreases, the pawl is no longer pushed upwards but cannot descend because the knee angle is approximately 10° . At this angle, the notched-disc is not truncated. However, this does not interfere because the upper surface of the notched-disc is smooth, without a notch. The pawl rolls thanks to his caster. The cycle of stairs ascent continues. The knee goes into flexion during the swing phase. The knee angle increases again. When it reaches 20° , the pawl can disengage and go back down because the notched-disc is truncated from this angle.

This mechanism exploits the angular range of the two tasks in the stance phase to allow the pawl to pass above the notched-disc of the locking wheel under the action of the vGRF in the case of stairs, or to remain below the notched-disc and lock into the locking wheel in the case of walking.

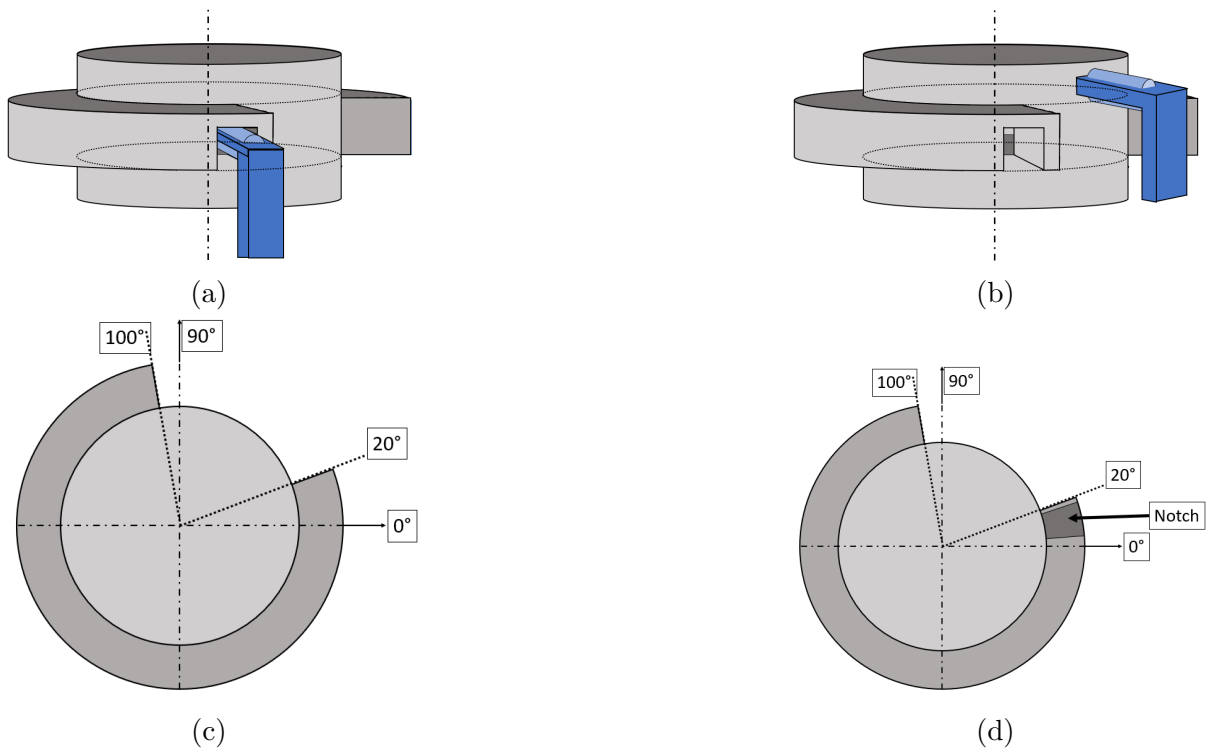


Figure 8.2: (a) Configuration of the locking wheel and the pawl during the Walking Stance Phase ; (b) Configuration of the locking wheel and the pawl during the Stairs Ascent Stance Phase ; (c) Top view of the locking wheel ; (d) Bottom view of the locking wheel. Note that figures (a) and (b) are drawn in the locking wheel frame of reference. This explains why the pawl rotates around it in the figures. In reality, it is the locking wheel that rotates and the pawl is only free to move in the direction of the length of the prosthesis. (c) and (d) : The notched-disc is in dark grey

8.4 Mechanism concept: Stage 2

The previous section provided an intuition of how the mechanism works, but did not take into account the 3.8 reduction ratio of the bevel gear between the locking wheel and the knee joint. The section here includes this reduction ratio in the reasoning. The operating principle of the mechanism remains the same, but the angular ranges of the locking wheel have been modified.

As a reminder, the reduction ratio of 3.8 comes from the final optimisation carried out in the previous chapter. The optimisation led to the need for a reduction ratio of 19 in addition to that provided by the 2PG. The maximum empirical reduction ratio to ensure robustness and reliability is 5, which is provided by the gear head in figure Figure 7.1. The remaining ratio of 3.8 is provided by the bevel gear.

This was chosen over a Maxon gearbox for greater efficiency. The Maxon gearbox with a reduction ratio of 19 has an efficiency of 81% [36]. The efficiency of a Straight Bevel Gear is between 93 and 97% and that of a Spur Gear between 94 and 98%. [**Gears_Efficiencies**]. The product of the two gives a range between 87 and 95 %, much better than the Maxon gearbox.

This reduction ratio means that the angle of the locking wheel varies 3.8 times faster. The reference angles of the notched-disc must therefore be modified. The maximum opening angle of 100° shown in Figure 8.2c and in Figure 8.2d has been reduced to 90° so as not to exceed 360° otherwise the entire notched-disc would be truncated. The truncated part is now from 76° to 342° :

Knee Angle	Notched-disc Angle
0°	0°
10°	38°
20°	76°
90°	342°

Table 8.1: Comparison of the Angular Ranges

The new geometry of the locking wheel is presented at Figure 8.3 :

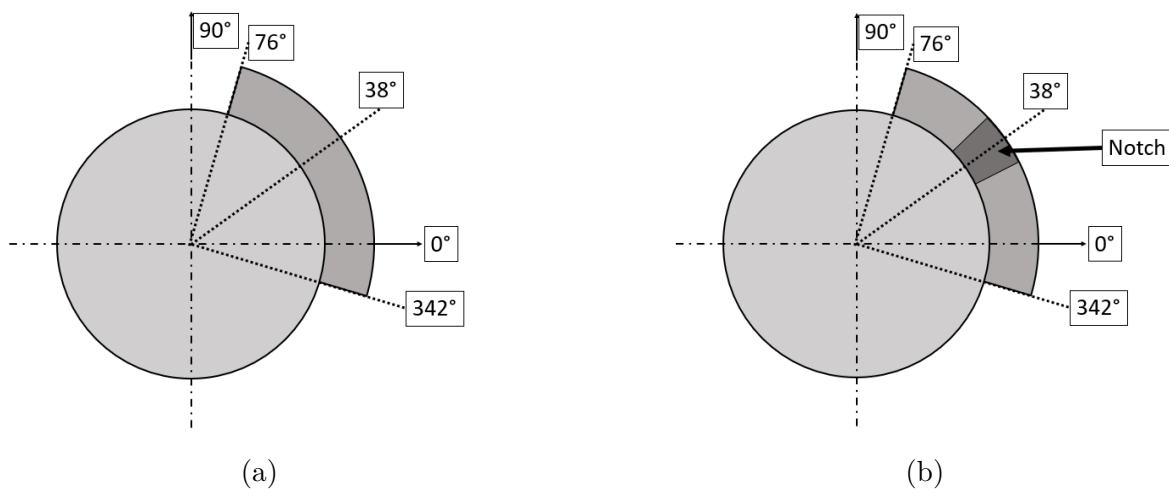


Figure 8.3: (a) Top view of the locking wheel ; (b) Bottom view of the locking wheel. The notched-disc is in dark grey

8.5 Comparison with the specifications

Although this mechanism is at a very conceptual stage, it is already possible to make an initial comparison with the specifications drawn up here [section 4.2](#).

Two main functions have been established with four criteria. The kinematic chain must be modified and restored passively. The four criteria for the two functions are identical. These criteria can be assessed for both functions at the same time because the same mechanism performs both. The evaluation is carried out in the following [Table 8.2](#).

Main Functions	
F1 & F2: Modify & Re-establish passively the kinematic chain	
It's the angle of the joint and the reaction force of the ground that modify the kinematic chain.	
Performance criteria	Performance level
Activation time	Rapid
The reaction force of the soil increases very rapidly. It starts at 0% of the cycle and reaches its peak at 15 % of the cycle, which represents between 0.15 and 0.18 seconds. It's pretty fast.	
Activation reliability	The mechanism follows the 6 σ method. It can not fail more than 3.4 cycles <i>per million</i>.
It is not possible at this stage to assess whether the mechanism complies with the 6 sigma method. An initial prototype would be necessary to identify any remaining problems.	
Activation margin	The mechanism is activated close to the triggering point.
As a precaution, the Early Stance Flexion range has been extended to 20[°] in the event that contact with the ground occurs when the prosthetic knee is not perfectly straight.	
Reversibility	The mechanism can operate in both directions. The triggering point can become the release point and vice-versa.
The mechanism can work in both directions.	

Table 8.2: Functions Check-Up for the Concept of Mechanism

The constraints are evaluated in the [Table 8.3](#) below :

Main Constraints	
Performance criteria	Performance level
C1 : The kinematic chain must be lightweight	
Weight range	• 50-60 % of prosthesis weight
It has been discussed in chapter 7. The components used are those found in other prostheses that respect the weight ranges set. The constraint should be met.	
C2 : The kinematic chain must have a low encumbrance	
Max. vertical length	• 220 [mm] corresponding to 50% of the average shank's length, i.e. 12.3% of amputee's average height of 1.8 [m] [70]
A precise assessment of the length at this stage is difficult, but we can assume that the criterion will be met because the elements of the kinematic chain are compact (motor Maxon EC22, 2PG, spring, gear).	
C3 : The kinematic chain must be quiet	
Noise limit	• 30 [dB]
At this stage, it is difficult to estimate the sound level. The engagement of the pawl in the notch will probably be the main source of noise because of the impact between the two metal parts.	
C4 : The kinematic chain must be compliant	
Shock absorption	• Allow a certain degree of flexibility in the kinematic chain to absorb energy from shocks and impact
At this stage, it is not possible to assess this constraint.	
C5 : The kinematic chain must be able to work normally at different temperatures	
Temperature range	• [-10°C ; 40°C]
It has been discussed in chapter 7. Temperatures $\leq 22[^\circ]$ are validated as they improve heat exchange. Operation at the upper limit of 40[°] is possible with reduction of thermal resistance of 25% but the temperature reached is close to the maximum permitted. A more detailed thermal model would be required.	
C6 : Lifetime of the kinematic chain	
Number of cycles	• Walking¹: $\frac{1500}{2} \left[\frac{\text{cycle}}{\text{day}} \right] \cdot 365 \left[\frac{\text{day}}{\text{year}} \right] \cdot 5 [\text{year}] = 1.4 \cdot 10^6 [\text{cycle}]$ • Stairs ascent²: $50 \left[\frac{\text{cycle}}{\text{day}} \right] \cdot 365 \left[\frac{\text{day}}{\text{year}} \right] \cdot 5 [\text{year}] = 92\,000 [\text{cycle}]$
At this stage, it is not possible to assess this constraint. It has to be assessed by a mechanical fatigue test on a final prototype.	
C7 : The kinematic chain must be robust	
Max. acceptable force	• $9.81 \cdot 120 \cdot 1.5 = 1766 [\text{N}]$ (Support the max. weight of 120 [kg] with a safety factor of 1.5³)
The same as C6 but with a mechanical compression test.	
C8 : The kinematic chain must be protected from the environment	
Weatherproof and dust-proof	• IP value of 54, i.e. protection against dust deposits (solid) and protection against splashing water (liquid) [6].
At this stage, it is not yet possible to consider the mechanism's protective shell.	
C9 : The kinematic chain must be able to meet the maximum values identified on the biological curves of the lower limb tasks. (based on Table 3.2).	
• Flexion angle range • Max. torque • Max. speed	• $[0^\circ ; 100^\circ]$ • 1.4 [N/kg] • 420 [°/sec]
The mechanism is capable of bending up to 90[°]. Simulation of the chapter 7 confirms that it is possible to ascend the stairs on which these maximum torque and speed values are based. The locking wheel is integrated into the drive shaft. It therefore achieves the required performance.	

Table 8.3: Constraints Check-Up for the Concept of Mechanism

8.6 Mechanism limitations

This mechanism concept has several limitations:

- The angular position of the locking wheel depends on the knee angle and the position of the pawl depends on the reaction force. These two physical quantities are not dependent on each other. This can be a problem on flat but uneven ground where contact with the ground could be at an angle close to $20[^\circ]$. There could be cases where the horizontal part of the pawl with the caster interferes with the rotation of the notched-disc.
- This mechanism considers two very distinct cases, but we need to check that it can work on sloping ground and ramp.
- Translating the pawl into the limited space of the prosthesis seems difficult to achieve.
- The mechanism that uses the reaction force to translate the pawl has not been designed. It may need to be linked to the ankle prosthesis.
- The locking/unlocking is abrupt. It might be useful to prime the engagement of the pawl in the notch to avoid an impact due to the difference in level between the notch and the lower surface of the notched-disc. This could be done by tilting the lower surface of the notched-disc towards the notch to allow a progressive vertical movement of the pawl towards the notch.
- When the pawl is engaged in the pocket, it undergoes compressive and shear stresses.

Chapter 9

Conclusion

This chapter concludes the work that has been presented.

The chapter 2 established the landscape of knee prostheses for transfemoral amputees.

Then, chapter 3 presented the biomechanical aspects of common locomotion tasks with the aim of better understanding the constraints they imply in the design of a kinematic chain for transfemoral prostheses.

The chapter 4 set out to draw up the specifications for the design of a kinematic chain whose configuration can be modified passively during operation.

Once the specifications were established, chapter 5 looked at elastic actuators configurations. The results showed that this solution's family is particularly efficient at reducing walking electrical energy consumption, especially the 2PEA and CSEA configurations. The 2PEA exhibits outstanding results when optimized exclusively for walking, but proves to be less effective when optimized concurrently for walking and stairs ascent. The CSEA shows less efficient but more consistent results across the different optimizations.

Variable transmissions represent the second family of solutions, studied in chapter 6. In contrast to elastic actuator configurations, a variable reduction ratio has little impact on the motor electrical energy consumption or peak power. Instead, by adjusting the reduction ratio along the gait cycle to adapt the motor speed and torque requirements, these transmissions allow the use of smaller motors to drive the prosthesis. Specifically, the 2PG and the Utah bionic leg are both capable of operating with the smallest candidate motor, the Maxon EC22, without overheating during sustained stairs ascent.

A global comparison of the two approaches was then carried out in chapter 7. It states that rather than competing with each other, the 2 families of solutions investigated complement each other. The idea then arose of combining them to achieve an even better response to the design objectives identified in the specifications. Out of the four possible combinations, each pairing one of the top configurations from both families, a weighted function helped determine the most beneficial combination, identified as the CSEA-2PG. Optimizing this combination delivers better results than the individual approaches.

Finally, a draft concept for a mechanism implementing the locking function of the CSEA was proposed in chapter 8.

It is important to bear in mind that the work presented here has its limitations. The mathematical models used are simple and no perturbations have been considered. In addition, the optimizations are based on curves taken from non-amputees, whereas amputees often exhibit different movement behavior. Finally, the proposed locking mechanism is still in the development phase.

Studying various kinematic chains is a crucial step in the research and development of active knee prostheses for transfemoral amputees. By striving towards an energy-efficient and lightweight design, the aim is to create a versatile and comfortable prosthetic device capable of restoring a sense of normality to the patient. Future perspectives may include the dimensioning and prototyping of the identified optimal kinematic chain (CSEA-2PG), as well as the development of the conceptual locking mechanism to passively operate the CSEA configuration.

Appendix A

Supplementary graphs for elastic actuator configurations

The purpose of this appendix is to provide supplementary diagrams for the elastic actuator configurations whose mechanical advantages were described but not graphically displayed in chapter 5. This aims to help the reader to better understand how elastic elements integrated into the kinematic chain can improve energetic efficiency, thereby reducing the workload on the motor.

A.1 CSEA Profiles

Two graphs that were not displayed for the CSEA configuration in subsection 5.6.2 are the torque-angle profiles for walking and stairs ascent, visible in Figure A.1.

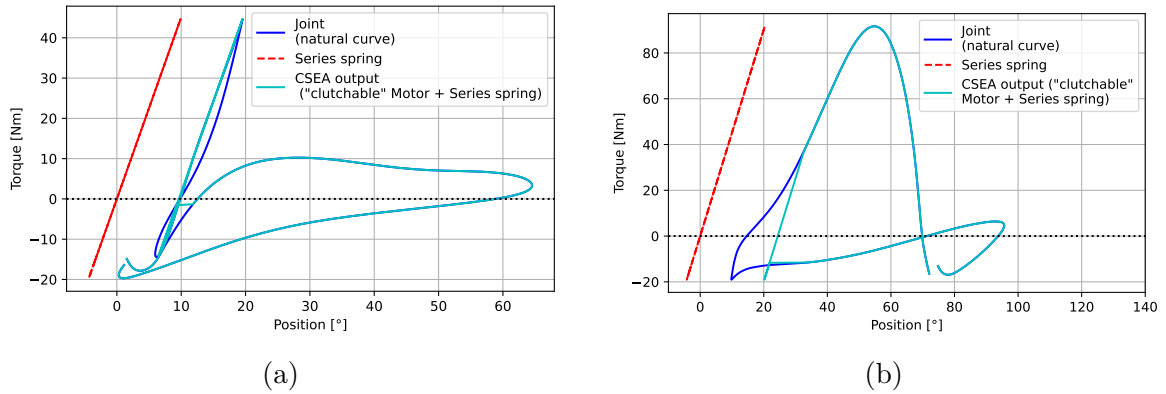


Figure A.1: Torque-angle profiles for the CSEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) task: walking (b) task: stairs ascent

A.2 PEA Profiles

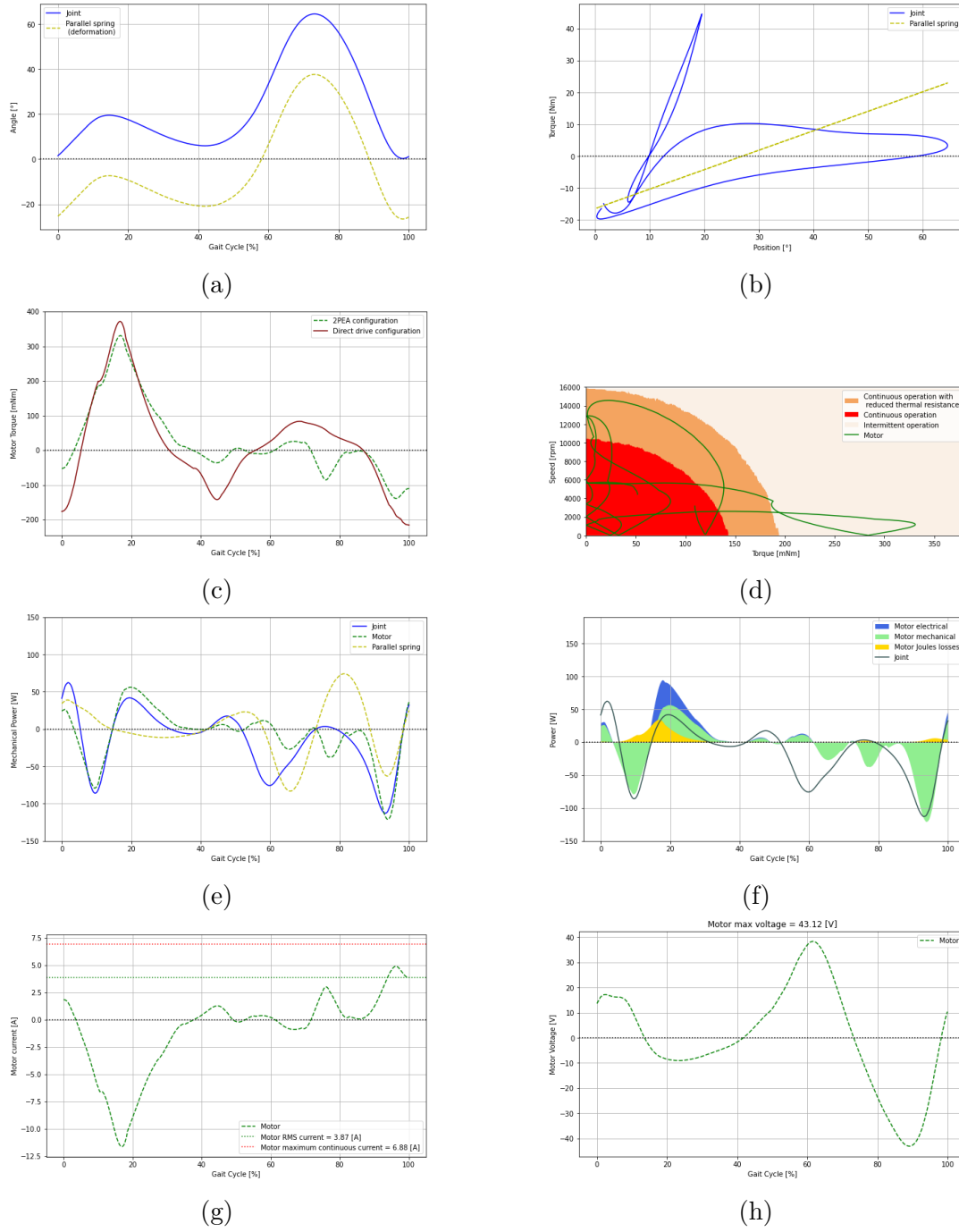


Figure A.2: Walking profiles for the PEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) mechanical powers (f) motor powers (g) motor current (h) motor voltage

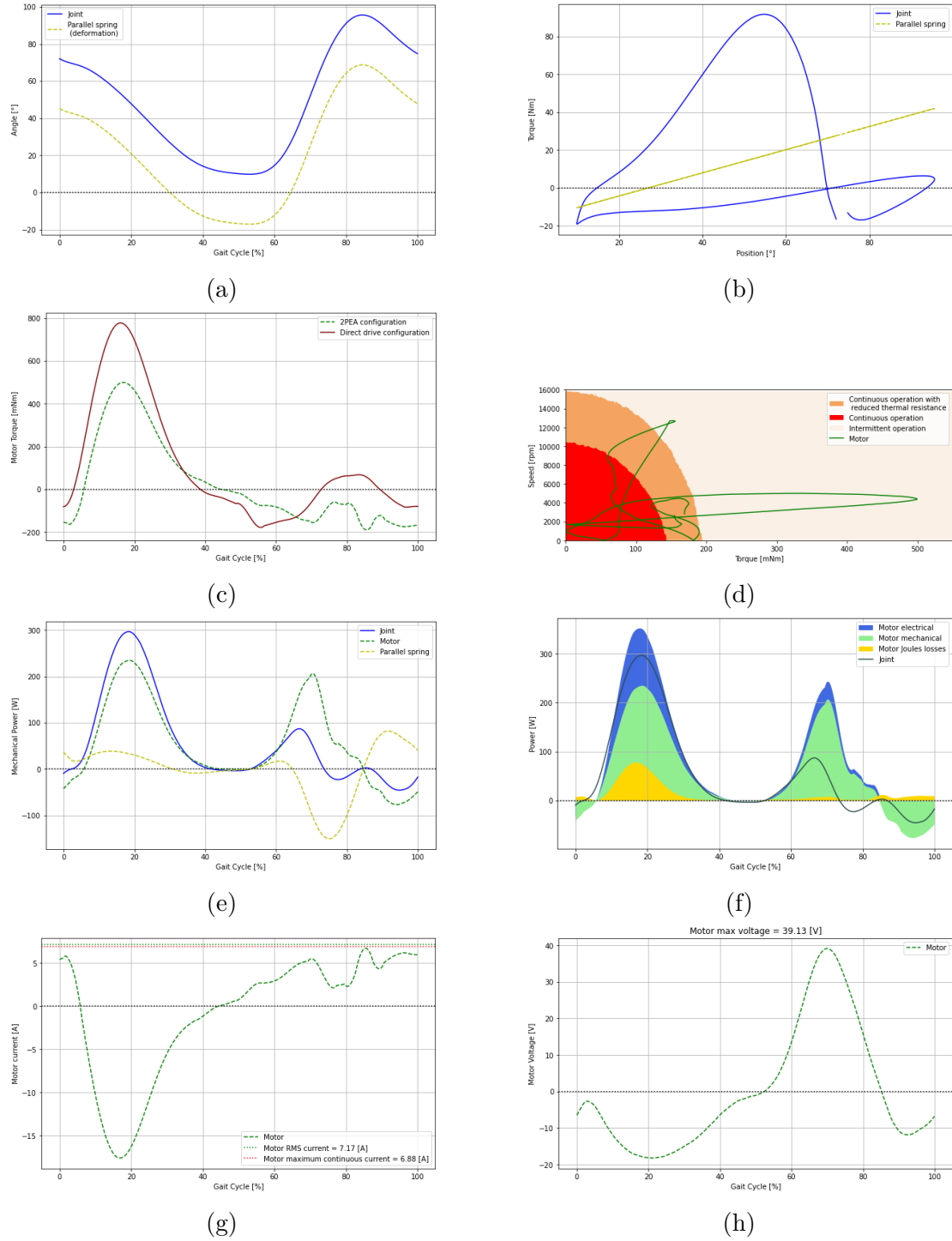


Figure A.3: Stairs ascent profiles for the PEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) mechanical powers (f) motor powers (g) motor current (h) motor voltage

A.3 CPEA Profiles

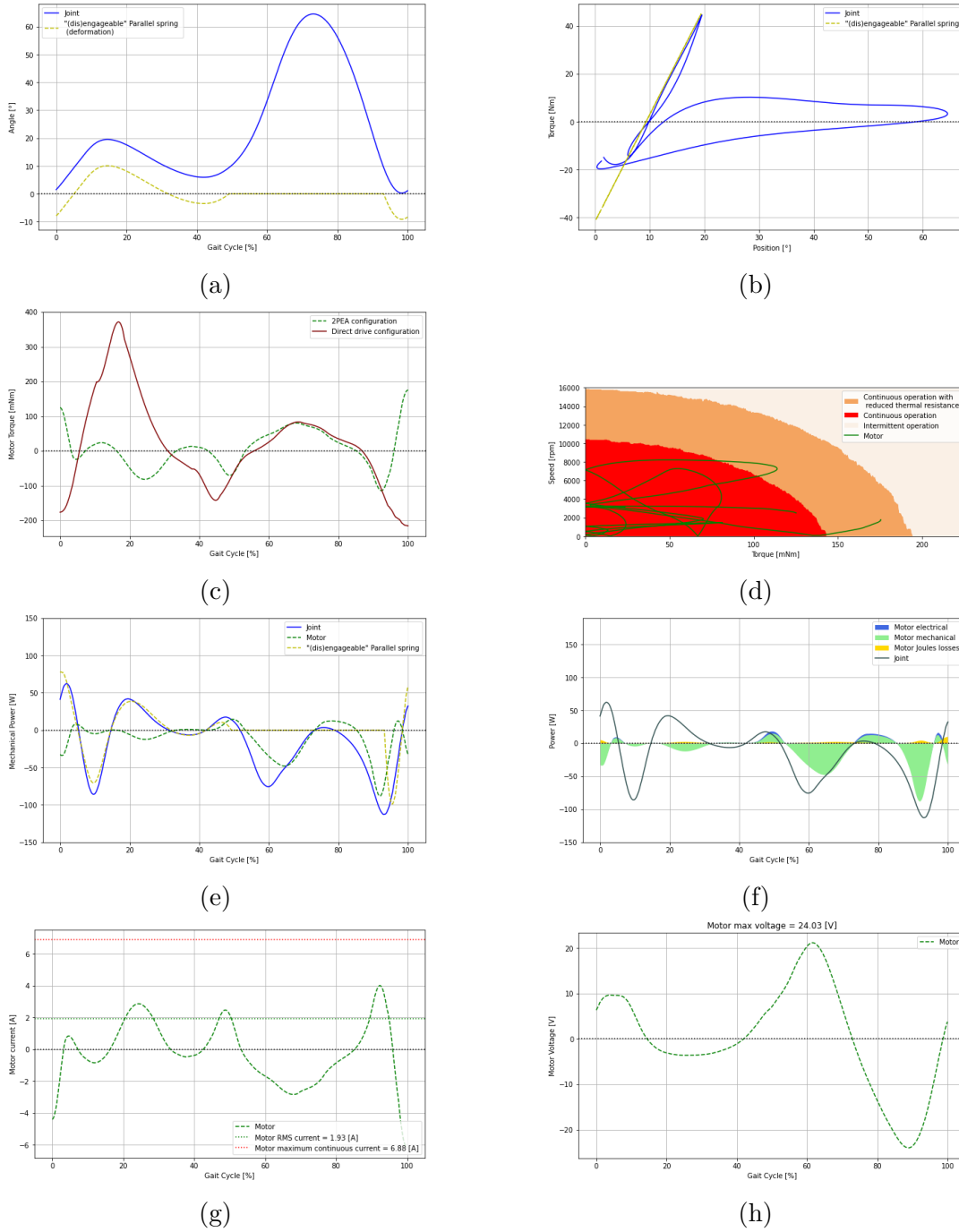


Figure A.4: Walking profiles for the CPEA configuration embedding a Faulhaber BP3274 motor optimized for walking only (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) mechanical powers (f) motor powers (g) motor current (h) motor voltage

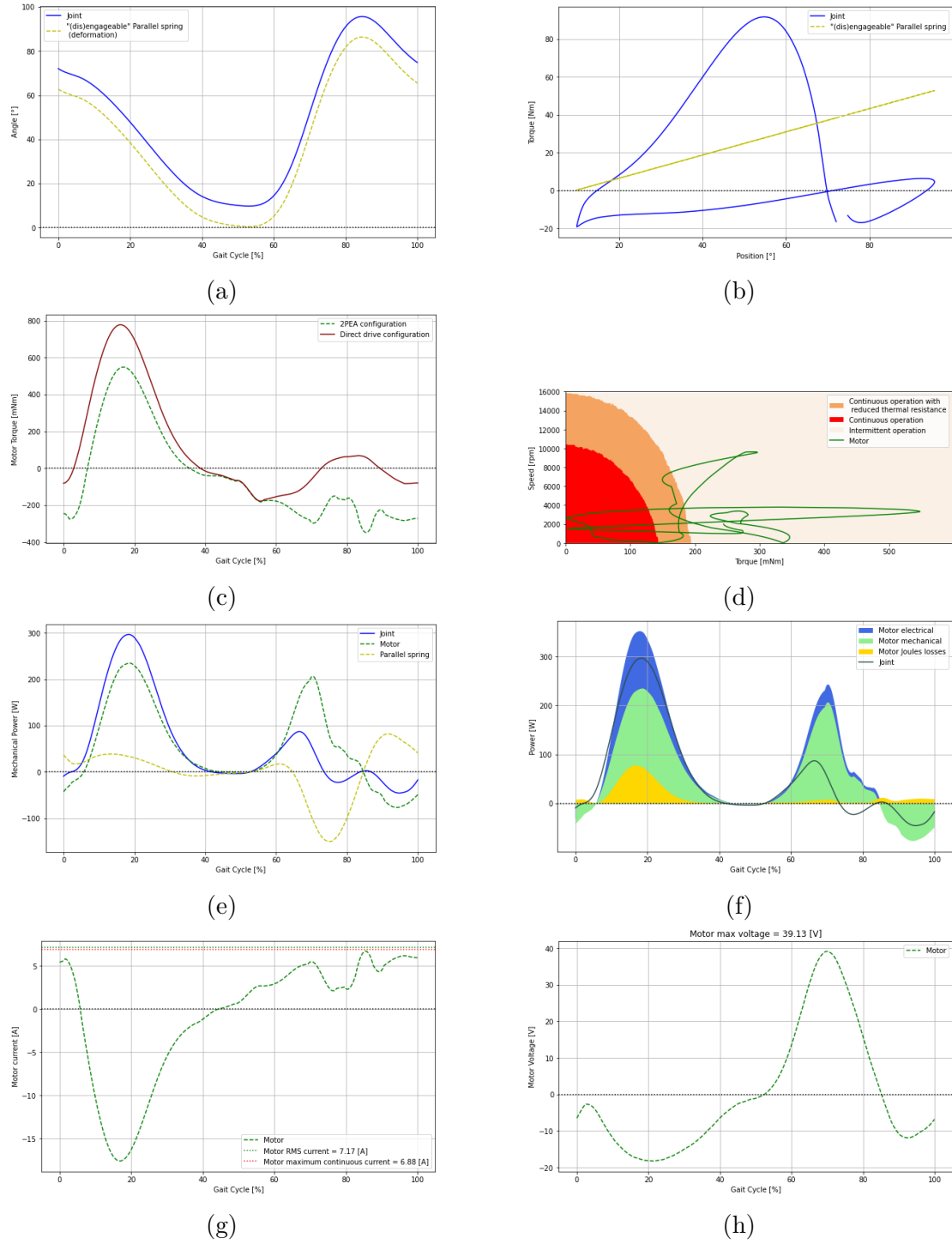


Figure A.5: Stairs ascent profiles for the CPEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) mechanical powers (f) motor powers (g) motor current (h) motor voltage

A.4 2PEA Profiles

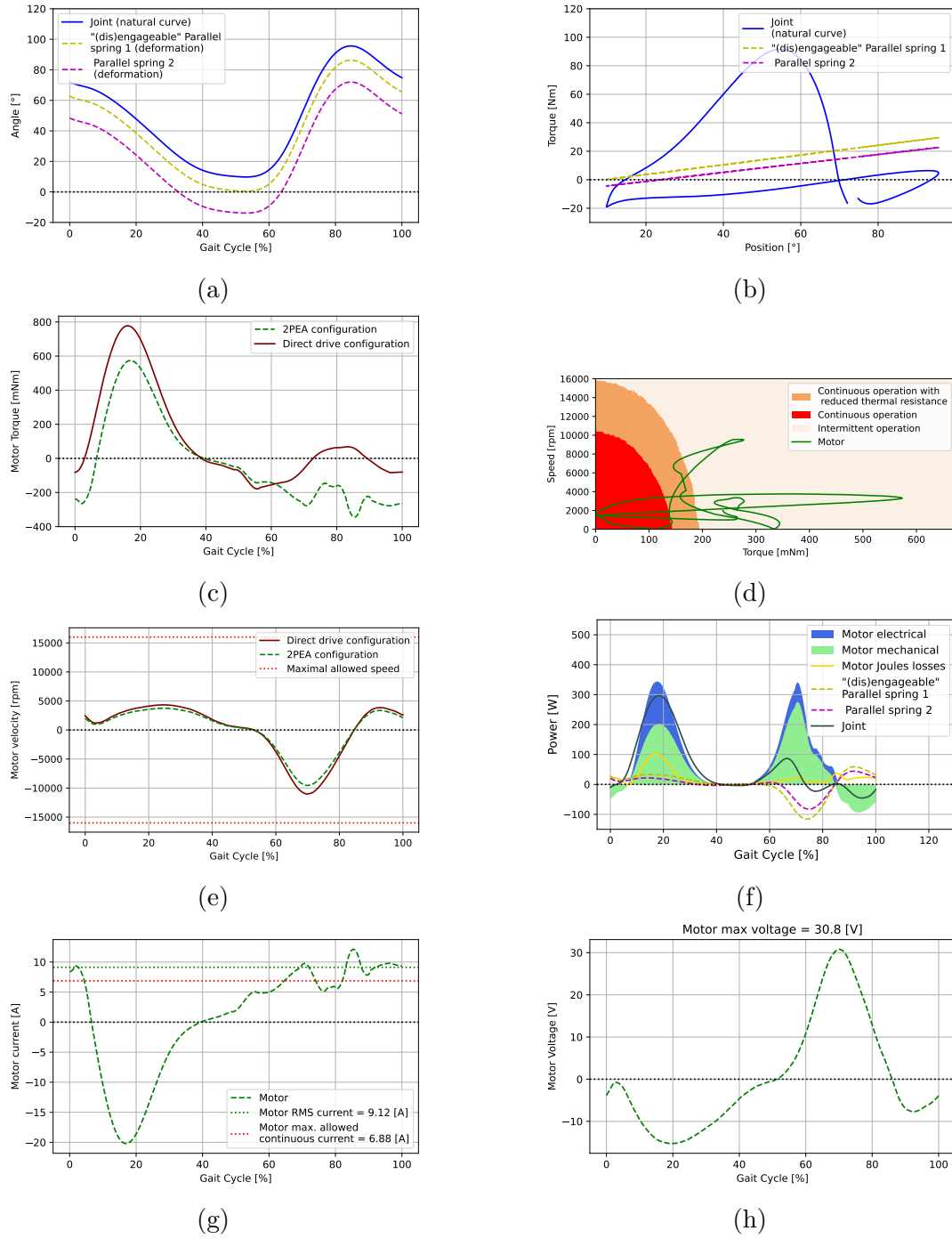


Figure A.6: Stairs ascent profiles for the 2PEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) motor speed (f) motor powers (g) motor current (h) motor voltage

A.5 PSEA Profiles

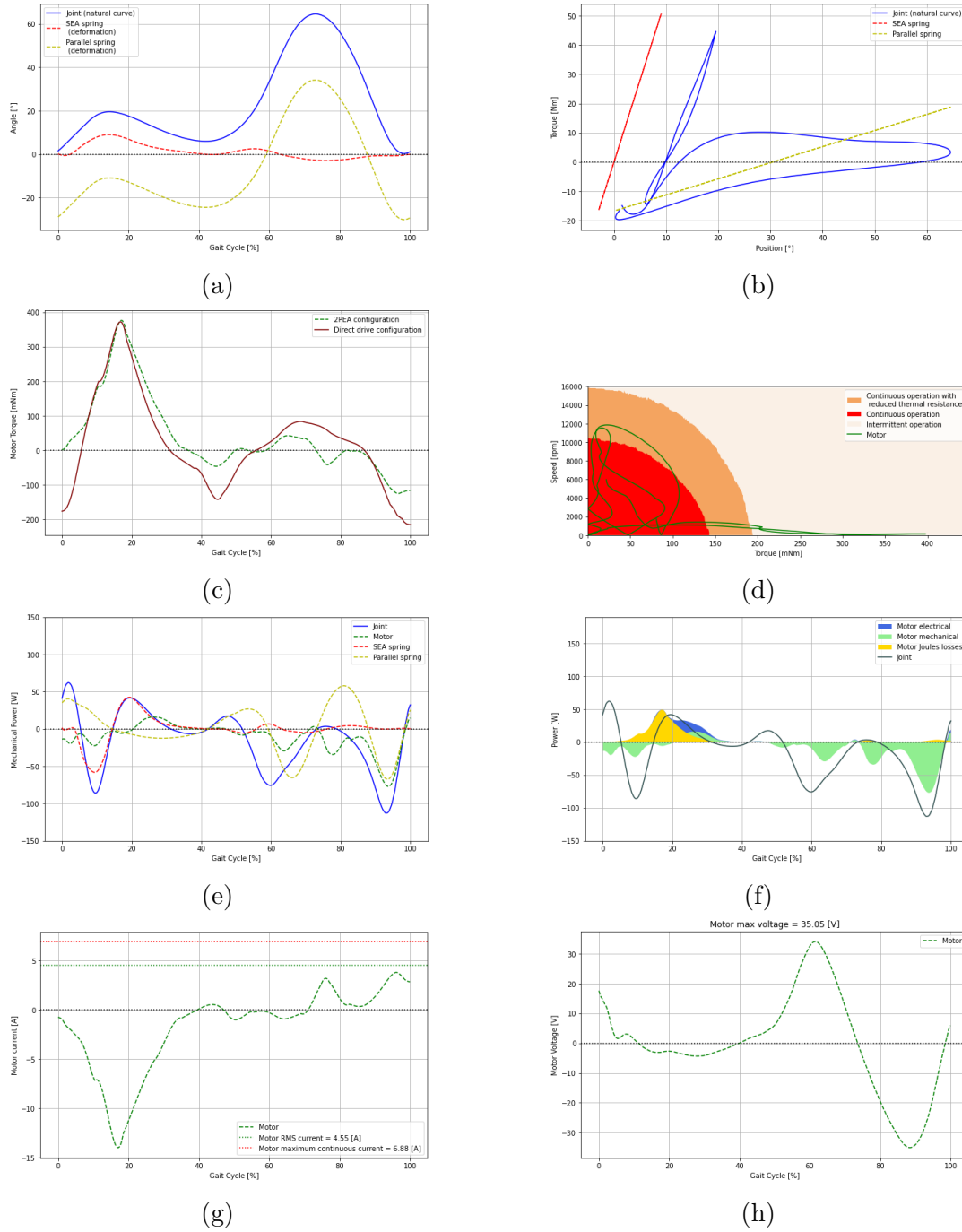


Figure A.7: Walking profiles for the PSEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) mechanical powers (f) motor powers (g) motor current (h) motor voltage

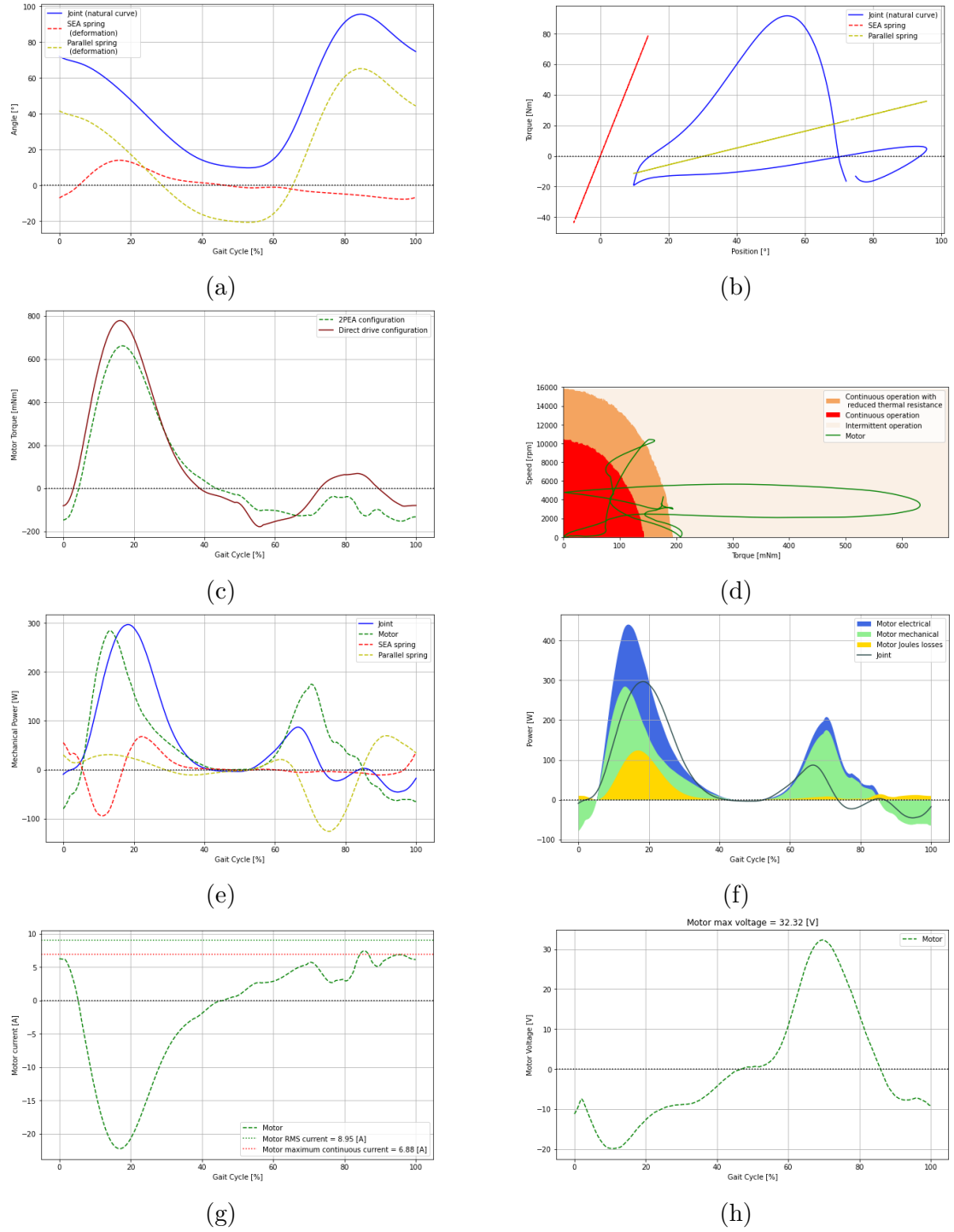


Figure A.8: Stairs ascent profiles for the PSEA configuration embedding a Faulhaber BP3274 motor optimized simultaneously for walking and stairs ascent (a) position (b) torque-angle (c) motor torque (d) speed-torque operating range (e) mechanical powers (f) motor powers (g) motor current (h) motor voltage

Appendix B

Mathematical model for the CRC mechanism

The coordinates of the two extremities of the connecting rod are $(r \cos \theta_1, r \sin \theta_1)$ and (a, λ) . Applying the well-known distance equation yields:

$$(r \cos \theta_1 - a)^2 - (r \sin \theta_1 - \lambda)^2 = L^2 \quad (\text{B.1})$$

$$\Leftrightarrow r^2 + a^2 - 2ra \cos \theta_1 + \lambda^2 - 2r\lambda \sin \theta_1 - L^2 = 0 \quad (\text{B.2})$$

Posing $A = -2r\lambda$, $B = -2ra$ and $C = r^2 - L^2 + a^2 + \lambda^2$, the above equation can be rewritten as:

$$A \sin \theta_1 + B \cos \theta_1 + C = 0 \quad (\text{B.3})$$

$$(\text{B.4})$$

Using an online solver (*WolframAlpha*), the angle θ_1 can be determined as a function of the vertical position of the ball screw nut λ as:

$$\theta_1 = 2 \tan^{-1} \left(\frac{A - \sqrt{A^2 + B^2 - C^2}}{B - C} \right) \quad (\text{B.5})$$

The variable reduction ratio between the ball screw and the knee joint is here given by the variation of the input linear speed divided by the variation of the output rotational speed:

$$N_{crc} = \frac{ds}{d\theta} = \frac{d(\lambda_{init} - \lambda)}{d(\theta_{1,init} - \theta_1)} = \frac{d\lambda}{d\theta_1} = \frac{r\lambda \cos \theta_1 - ra \sin \theta_1}{\lambda - r \sin \theta_1} \quad (\text{B.6})$$

Appendix C

Mathematical model for the Utah bionic leg

The equations of the mathematical model presented in [61] are partially erroneous. For this reason, all the calculations are shown below, together with the algorithm for solving them.

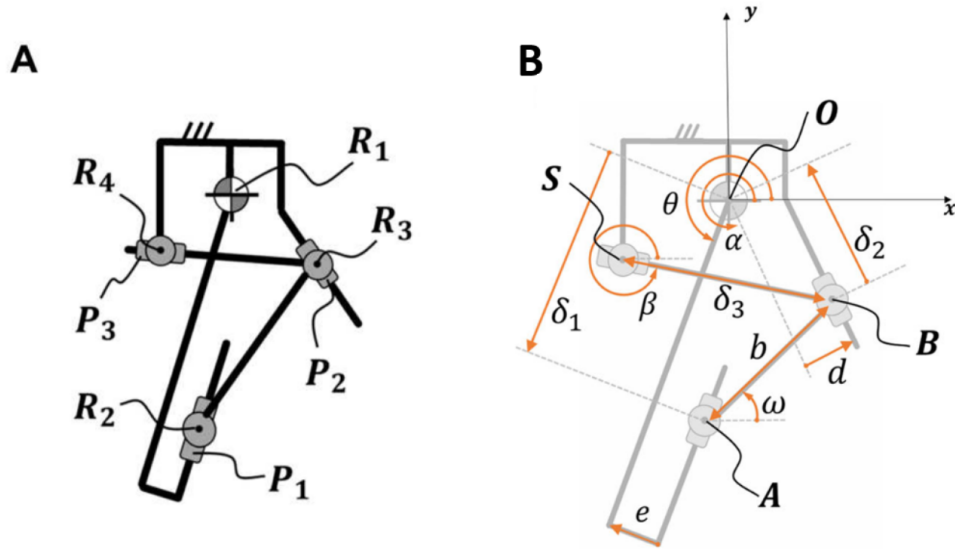


Figure C.1: Taken from [61]. Kinematic diagram of the knee module in the Utah Bionic Leg (A) highlights name of the joints (B) highlights name of the variables

Based on the kinematic diagram C.1, the points A and B can be defined as following :

$$\begin{cases} A_x = \delta_1 \cos(\theta) - e \cos\left(\theta - \frac{\pi}{2}\right) \\ A_y = \delta_1 \sin(\theta) - e \sin\left(\theta - \frac{\pi}{2}\right) \end{cases} \quad (C.1)$$

$$\begin{cases} B_x = \delta_2 \cos(\alpha) - d \cos\left(\alpha - \frac{\pi}{2}\right) \\ B_y = \delta_2 \sin(\alpha) - d \sin\left(\alpha - \frac{\pi}{2}\right) \end{cases} \quad (C.2)$$

These coordinates will make it easy to express the kinematic relationships that govern the mechanism. The length of the connecting bar, b , is defined by the following equation :

$$b = \overline{AB} = \sqrt{(A_x - B_x)^2 + (A_y - B_y)^2} \quad (C.3)$$

From the equation C.3, δ_1 can be found by solving a second order equation as $A \cdot \delta_1^2 + B \cdot \delta_1 + C = 0$:

$$\underbrace{1}_A \delta_1^2 + \delta_1 \cdot \underbrace{[-2 \cdot (\delta_2 \cos(\alpha - \theta) - d \sin(\alpha - \theta))]}_B - \underbrace{2 \cdot e \cdot [\delta_2 \sin(\alpha - \theta) + d \cos(\alpha - \theta)] + \delta_2^2 + e^2 + d^2 - b^2}_C = 0 \quad (C.4)$$

Then, δ_3 is the length of the spring which is also the distance between the fixed point S and the moving point B :

$$\delta_3 = \sqrt{(B_x - S_x)^2 + (B_y - S_y)^2} \quad (C.5)$$

With regard to kinetics, the following variables will be useful for studying forces :

$$\overline{OB} = \sqrt{B_x^2 + B_y^2} \quad (C.6) \quad \widehat{OB} = \arctan 2 \left(\frac{B_y}{B_x} \right) \quad (C.7)$$

$$\omega = \arctan 2 \left(\frac{B_y - A_y}{B_x - A_x} \right) \quad (C.8) \quad \beta = 2\pi + \arctan 2 \left(\frac{B_y - S_y}{B_x - S_x} \right) \quad (C.9)$$

Then, to introduce kinetics into the model, the force equilibrium at point B can be calculated based on the sktech C.2:

$$\begin{aligned} \vec{R}_s + \vec{R}_{ES} + \vec{R}_{P2} + \vec{F}_b &= \vec{0} \\ \vec{F}_b &= \vec{F}_s + \vec{F}_{ES} + \vec{F}_{P2} \end{aligned} \quad (C.10)$$

with F_s defined as following :

$$F_s = k \cdot (\delta_3 - \delta_0) + F_{s0} \quad (C.11)$$

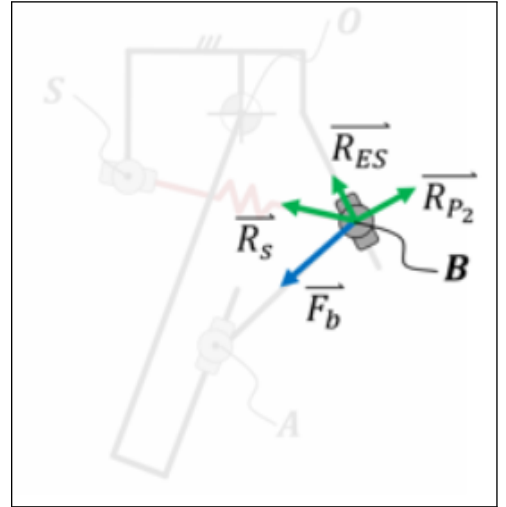


Figure C.2: Force equilibrium at point B (taken from [61])

where :

$\vec{R}_s$ is the reaction force generated by the tension spring $= -\vec{F}_s$;

$\vec{R}_{ES}$ is the reaction force longitudinal to the prismatic joint $P_2 = -\vec{F}_{ES}$;

$\vec{R}_{P2}$ is the reaction force perpendicular to the prismatic joint $P_2 = -\vec{F}_{P2}$;

$\vec{F}_b$ is the action force of the connecting bar ;

k is the spring stiffness ;

δ_0 the free length of the spring ;

F_{s0} is the preconstraint force of the extended spring.

The force equilibrium is expressed as follows in the x-y reference frame of the diagram :

$$\begin{aligned}
& +|F_{P2}| \sin(\alpha) - |F_b| \cos(\omega) + |F_{ES}| \cos(\alpha) - |F_s| \cos(\beta) = 0 \\
& -|F_{P2}| \cos(\alpha) - |F_b| \sin(\omega) + |F_{ES}| \sin(\alpha) - |F_s| \sin(\beta) = 0 \\
& \quad \quad \quad \Updownarrow \\
& -|F_b| \cos(\omega) - |F_s| \cos(\beta) = -(+|F_{P2}| \sin(\alpha) + |F_{ES}| \cos(\alpha)) \\
& -|F_b| \sin(\omega) - |F_s| \sin(\beta) = -(-|F_{P2}| \cos(\alpha) + |F_{ES}| \sin(\alpha))
\end{aligned} \tag{C.12}$$

The force equilibrium is used to calculate the position of point B at each point in the cycle of the task in question.

$$\begin{aligned}
\begin{bmatrix} -\sin(\alpha) & -\cos(\alpha) \\ +\cos(\alpha) & -\sin(\alpha) \end{bmatrix} \cdot \begin{bmatrix} |F_{P2}| \\ |F_{ES}| \end{bmatrix} &= \begin{bmatrix} -|F_b| \cos(\omega) - |F_s| \cos(\beta) \\ -|F_b| \sin(\omega) - |F_s| \sin(\beta) \end{bmatrix} \\
\begin{bmatrix} |F_{P2}| \\ |F_{ES}| \end{bmatrix} &= \begin{bmatrix} -\sin(\alpha) & +\cos(\alpha) \\ -\cos(\alpha) & -\sin(\alpha) \end{bmatrix} \cdot \begin{bmatrix} -|F_b| \cos(\omega) - |F_s| \cos(\beta) \\ -|F_b| \sin(\omega) - |F_s| \sin(\beta) \end{bmatrix}
\end{aligned} \tag{C.13}$$

The mathematical relation between F_b and F_a can also be established based on C.3:

$$\begin{aligned}
\vec{F}_b &= \begin{pmatrix} -|F_b| \cos(\omega) \\ -|F_b| \sin(\omega) \end{pmatrix} \quad (\text{in the x - y reference frame}) \\
\vec{F}_b &= \begin{bmatrix} |F_b| \cos(\omega + \pi - \theta) \\ |F_b| \sin(\omega + \pi - \theta) \end{bmatrix} = \begin{bmatrix} |F_a| \\ |R_{P1}| \end{bmatrix} \quad (\text{in the } F_a R_{P1} \text{ frame}) \\
\Rightarrow |F_b| &= \frac{|F_a|}{\cos(\omega + \pi - \theta)}
\end{aligned} \tag{C.14}$$

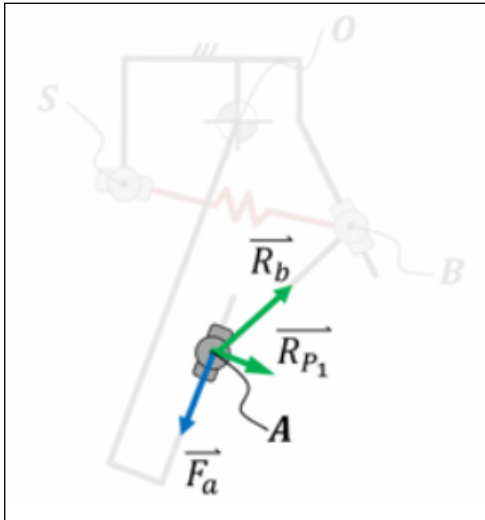


Figure C.3: Force equilibrium at point A (taken from [61])

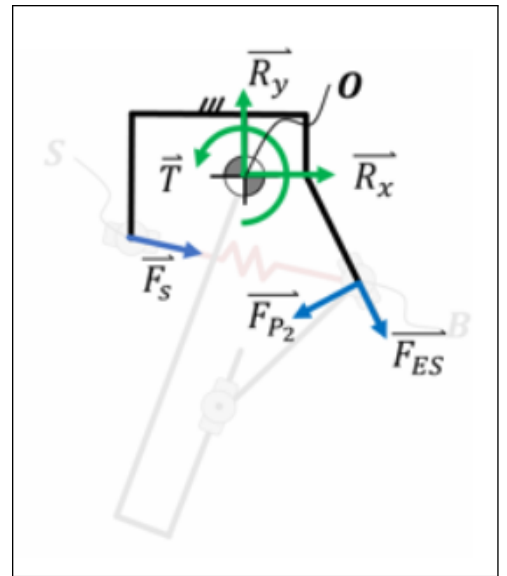


Figure C.4: Torque equilibrium at origin O (taken from [61])

A similar approach is used to calculate the relationship between torque T and force F_B by performing a torque balance at the origin C.4:

$$\begin{aligned}
\vec{T} + \vec{OB} \times \vec{F}_b &= \vec{0} \\
\begin{pmatrix} 0 \\ 0 \\ |T| \end{pmatrix} + \begin{pmatrix} \overline{OB} \cos(\widehat{OB}) \\ \overline{OB} \sin(\widehat{OB}) \\ 0 \end{pmatrix} \times \begin{pmatrix} -|F_b| \cos(\omega) \\ -|F_b| \sin(\omega) \\ 0 \end{pmatrix} &= \begin{pmatrix} 0 \\ 0 \\ 0 \end{pmatrix} \\
\Rightarrow |T| &= -(-\overline{OB} \cos(\widehat{OB}) \cdot |F_b| \sin(\omega) + \overline{OB} \sin(\widehat{OB}) \cdot |F_b| \cos(\omega)) \\
\Rightarrow \frac{|T|}{|F_b|} &= \overline{OB} (\cos(\widehat{OB}) \sin(\omega) - \sin(\widehat{OB}) \cos(\omega))
\end{aligned} \tag{C.15}$$

The torque ratio is defined as following :

$$\tau_{torque} = \frac{|T|}{|F_a|} \tag{C.16}$$

Its developed expression can be obtained from the results of equations C.14 and C.15 :

$$\begin{aligned}
\tau_{torque} &= \frac{|T|}{|F_a|} \\
&= \frac{|T|}{|F_b| \cos(\omega + \pi - \theta)} \\
&= \frac{\overline{OB} (\cos(\widehat{OB}) \cos(\omega) - \sin(\widehat{OB}) \sin(\omega))}{\cos(\omega + \pi - \theta)}
\end{aligned} \tag{C.17}$$

where $\widehat{OB}$, $\overline{OB}$, ω have been expressed in equations C.7, C.6 and C.8. The motor's speed and the motor's torque are defined as following:

$$\begin{aligned}
\omega_m &= (N_g N_{bs}) \cdot \dot{\delta}_1 \\
T_m &= I \cdot a_m + \frac{F_a}{(N_g N_{bs})} \cdot \eta_{mec}
\end{aligned} \tag{C.18}$$

To obtain them, the procedure is the following one:

At each time step, one needs to find the distance δ_2 which cancels out (or at least minimizes for the bounds of δ_2) the force F_{ES} along the axis of translation of point B. Then, an iteration over the range of δ_2 values is performed to calculate F_{ES} for each of them. The value that minimizes the force F_{ES} is selected.

Under the action of the force F_b , the spring produces a force F_s . Except in cases where the joint reaches the lower stop δ_{2l} or the upper stop δ_{2h} , the force F_{ES} is zero for any other distance δ_2 .

Once δ_2 has been determined for each of the time steps, all the other variables can be definitively calculated to obtain the motor speed and torque.

Appendix D

Supplementary graphs for variable transmission configurations

The purpose of this appendix is to provide supplementary diagrams for the variable transmission configurations whose advantages were described but not graphically displayed in chapter 6. This aims to help the reader to better understand how a variable reduction ratio integrated into the kinematic chain can adapt the speed-torque requirements of the motor.

D.1 CRC profiles

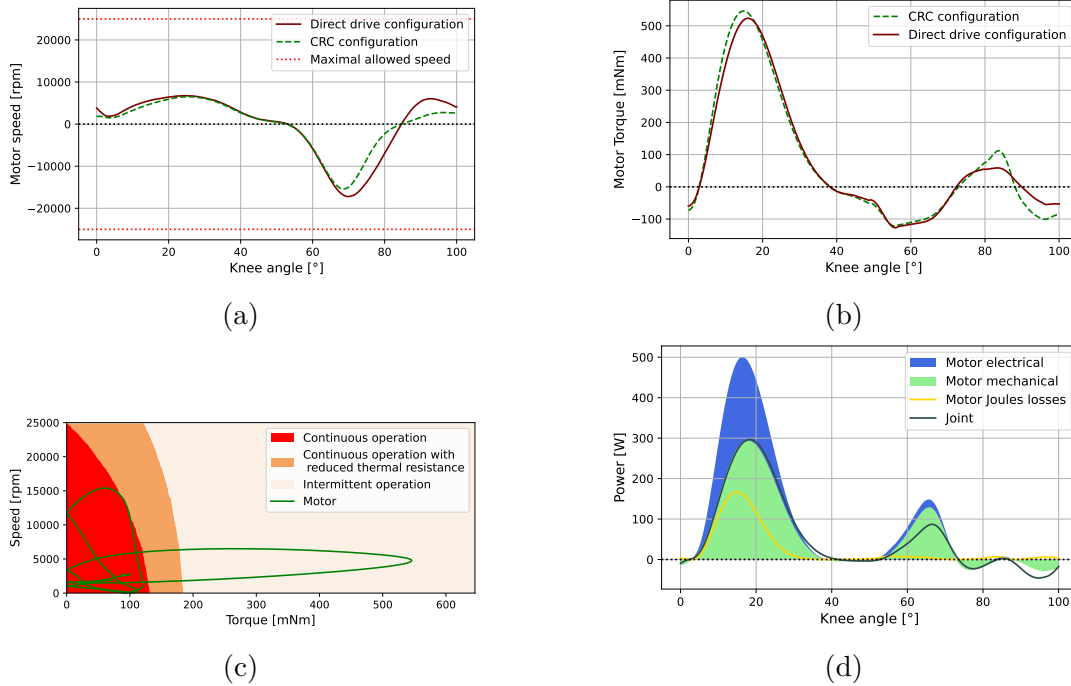


Figure D.1: Stairs ascent profiles for the CRC configuration embedding a Maxon EC30 motor optimized simultaneously for walking and stairs ascent (a) motor speed (b) motor torque (c) motor speed-torque operating range (d) motor powers

D.2 CVT profiles

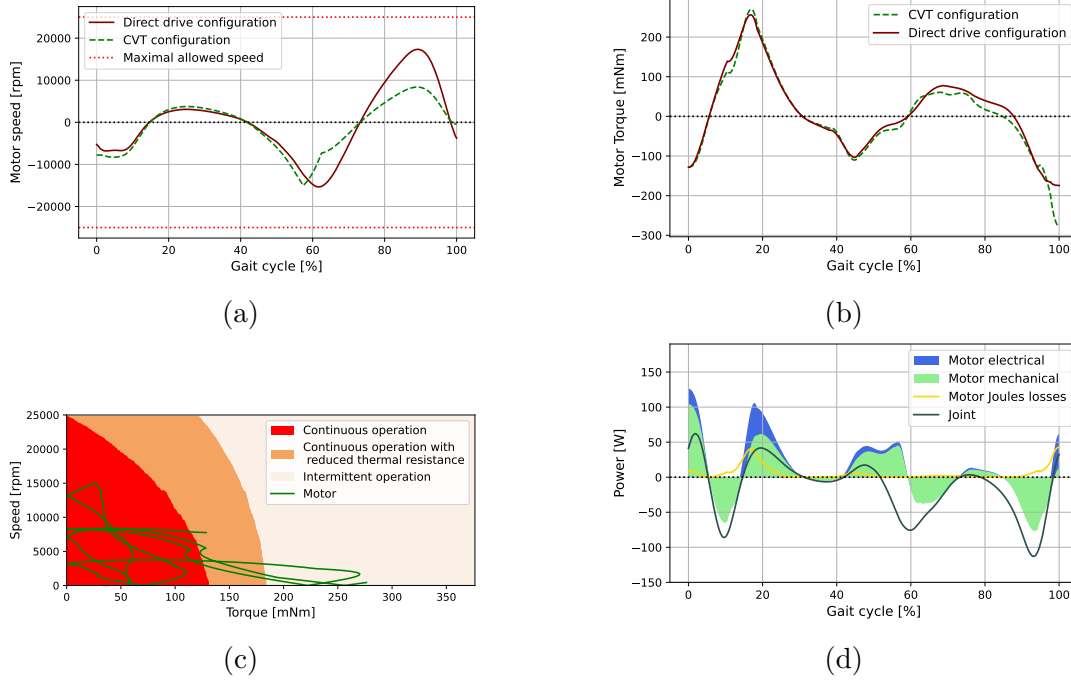


Figure D.2: Walking profiles for the CVT configuration embedding a Maxon EC30 motor optimized simultaneously for walking and stairs ascent (a) motor speed (b) motor torque (c) motor speed-torque operating range (d) motor powers

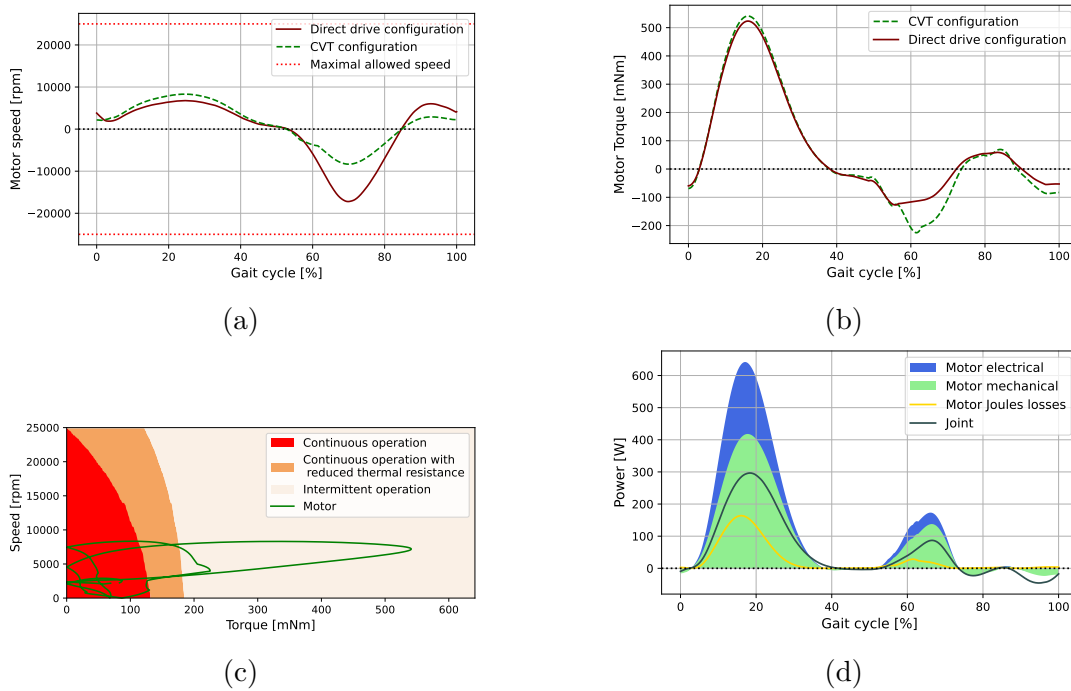


Figure D.3: Stairs ascent profiles for the CVT configuration embedding a Maxon EC30 motor optimized simultaneously for walking and stairs ascent (a) motor speed (b) motor torque (c) motor speed-torque operating range (d) motor powers

D.3 2PG profiles

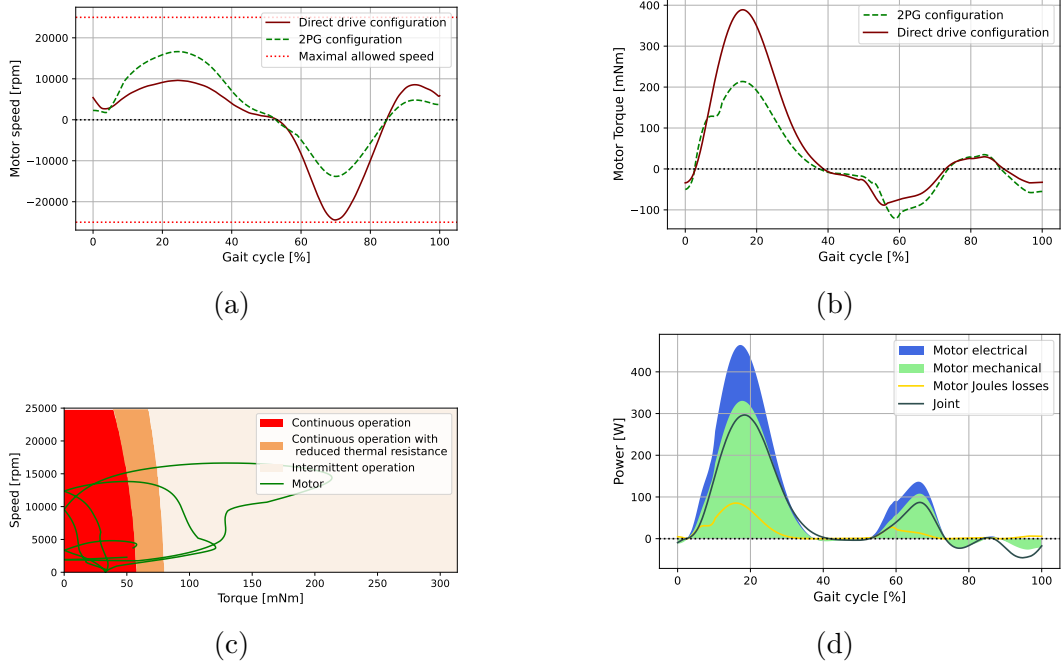


Figure D.4: Stairs ascent profiles for the 2PG configuration embedding a Maxon EC22 motor optimized simultaneously for walking and stairs ascent (a) motor speed (b) motor torque (c) motor speed-torque operating range (d) motor powers

D.4 Utah profiles

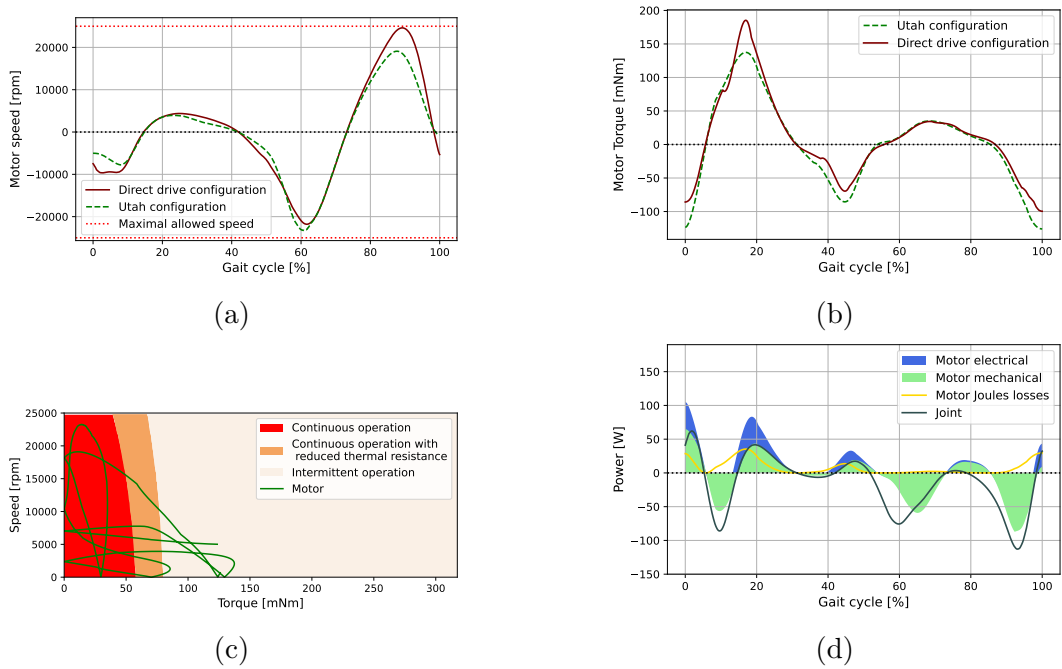


Figure D.5: Walking profiles for the Utah bionic leg embedding a Maxon EC22 motor optimized simultaneously for walking and stairs ascent (a) motor speed (b) motor torque (c) motor speed-torque operating range (d) motor powers

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