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Doctoral Dissertation

Doctoral Program in Mechanical Engineering (36th cycle)

Advancements in Robotic Knee Prostheses: Electro-Hydrostatic Actuation as a Means to Enhance Passivity.

By

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Declaration

I hereby declare that, the contents and organization of this dissertation constitute my own original work and does not compromise in any way the rights of third parties, including those relating to the security of personal data.

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2024

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Abstract

This research focused on designing, developing, and validating a novel partially-powered knee prosthesis, aiming to enhance the beneficial aspects of energetically passive microprocessor-controlled knees (MPKs). The enhancement involved integrating a compact electro-hydrostatic actuation unit with a controllable hydraulic valve. The design philosophy prioritized optimizing the prosthesis for swing-related activities, recognizing their vulnerability to external perturbations compared to load support phases. The proposed solution advanced prosthesis backdrivability and elevated versatility through a highly integrated actuation unit, facilitating integration by sharing the same motion transmission system for both active and passive locomotion activities. This resulted in augmented active power capabilities at the knee joint, contributing to a more responsive and adaptable prosthetic experience.

The doctoral work detailed the refinement of passivity enhancement, providing insights into the design and prototyping of the prosthesis. Supported by experimental evidence, the prototype's actuation unit underwent validation on a test bench, evaluating both active and passive functionalities. Subsequently, the prosthesis was evaluated considering a healthy subject equipped with an able-body adapter, during level ground walking and stair ascending. Comparative analysis against a commercial MPK showcased the distinct advantages of the developed prototype. With a weight of 2.15 kg, the prosthesis exhibited a markedly improved walking gait. The integration of the actuation unit enhanced the robustness of swing-related activities, contributing to increased toe clearance, reducing the likelihood of falling and scuffing. Experiments involving stair ascent highlighted the prototype's capability to perform a step-over step gait, showcasing remarkable versatility compared to traditional MPKs (i.e., minimal knee flexion is attained during stair ascent).

A functional analysis revealed that the prosthesis provided a minimum damping impedance of 2.6 Nms/rad at the knee joint and a maximum static friction of

approximately $1.4 \text{ N} \cdot \text{m}$. Under power generating conditions, particularly during a sinusoidal trajectory tracking task, the prototype could generate up to 52 W of power, achieving a peak flexion of up to 90° and a maximum rotational speed of 294 deg/s, highlighting the actuation unit's effectiveness in replicating the swing phase without necessitating hip effort. In conclusion, an energetic analysis, considering a 21.6 V battery pack with a capacity of 3350 mA · h, revealed promising device autonomy. The prosthesis could provide significant autonomy at various speeds, ranging from 4260 to 9350 strides per battery charge, translating to a usage range from a maximum of 3 to a minimum of 1.5 days, based on the average number of steps per day performed by transfemoral amputees (TFAs) according to established literature.

In essence, this research suggested that supplementing an MPK prosthesis with a compact actuation unit could not only enhance the robustness of the swing phase during walking but also decrease the likelihood of scuffing, stumbling or even falling. Additionally, the innovative design enabled negotiating stair ascent in a step-over fashion, marking a significant advancement with respect to state-of-the-art MPK devices.

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Chapter 1

Introduction

Globally, the frequency of limb amputations has consistently risen over the years. According to the 1996 National Health Interview Survey (NHIS), approximately 1.2 million individuals in the United States were reported to be living with limb loss [1]. This estimate experienced a 33 % increase by 2005, with 54 % attributed to dysvascular diseases, predominantly diabetes mellitus (two-thirds of cases), and 45 % associated with trauma [2]. Notably, 42 % of the affected population is aged over 65, 65 % are male, and 42 % belong to non-white racial groups. Projections for the year 2050 anticipate a substantial increase to 3.6 million, primarily driven by the prevalence of dysvascular conditions. The aging demographic and heightened incidence of vascular diseases among older adults are identified as primary causal factors.

Moreover, the World Health Organization (WHO) predicts a doubling of the proportion of individuals aged 60 and above, from 12 % to 22 %, between 2015 and 2050 [3]. A discernible 33 % rise in the incidence of dysvascular amputations has been recorded over the years, whereas both trauma and cancer-related amputations have exhibited approximately a 33 % reduction [4]. Specifically, the incidence of lower extremity amputation ranges from 46.1 to 9600 considering 100 thousand individuals in populations with vascular diseases, in contrast to 5.8 – 31 considering 100 thousand individuals in the general population [5]. In support of these projections, Lin et al. conducted an investigation into the burden and trends of diabetes across countries from 1990 to 2025. Their findings anticipate a 17 % and 20 % increase in global diabetes incidence and prevalence, respectively [6].

In light of these considerations, the increasing forecast of amputations demands effective steps towards advancing assistive and prosthetic care. These will serve as pivotal instruments in reinstating essential functionalities for individuals suffering from the loss of a limb, thereby significantly enhancing their overall quality of life. This doctoral work is specifically oriented toward lower limb amputations (LLAs). LLAs represent approximately 65 % of the overall amputation cases. In contrast, major amputations in upper limbs account for 8 % of the total.

From a physical perspective, the repercussions of lower limb amputations on mobility and locomotion activities are profound, demanding considerable adaptation or exertion in comparison to their able-bodied counterparts. Irrespective of the prosthetic device employed and the specific nature of the amputation, individuals with lower limb amputations consistently manifest alterations in their gait patterns [7]. Notable gait deficits in this population include augmented step length and width, extended double support duration, pelvic obliquity, and pronounced trunk lateral bending and rotation. These compensatory mechanisms are systematically adopted to ensure frontal plane stability, facilitate the effective elevation of the affected limb, and concurrently augment both stance time and force/torque production in the unaffected side during loading phases. Understanding and addressing these issues is imperative for the development of more effective and tailored interventions to enhance the rehabilitation and overall well-being of individuals with lower limb amputations. On the opposing spectrum, an augmentation in step length appears to correlate with heightened movements in the trunk and pelvis. An additional manifestation of asymmetric gait patterns involves a reduction in stance time on the affected side, counterbalanced by an increase on the non-affected side, potentially evolving into an extreme hopping gait cycle. The primary repercussions of these compensatory mechanisms manifest in amputees walking at a slower pace than their healthy counterparts across diverse terrain conditions [8–11]. Furthermore, they expend more energy per step [12–14] and impose excessive load on their healthy limb [15], leading to prevalent issues such as back pain, residual limb pain, and hip deterioration among lower limb amputees [16, 17].

In tandem with these physical constraints, amputees have to face profound psychological challenges stemming from the loss of a limb. Specifically, between 21 % and 35 % of individuals with amputations experience depressive disorders and body image anxiety, figures notably higher than the 10–15 % observed in the general population [18]. However, it is noteworthy that these psychological effects tend to

diminish after the initial two years post-amputation. Furthermore, aspects related to social functioning and discomfort significantly impact individuals with amputations. A study by Burger et al. [19] revealed that approximately two-thirds of participants were less inclined to go out in public. Common prosthetic-related psychological issues include anxiety, social withdrawal, and post-traumatic stress disorder [20]. Finally, Horgan et al. [21] sought to explore the impact of prostheses on the psychological adjustment to amputation. Beyond fulfilling functional requirements, a prosthesis should function as a tool to conceal the amputation and restore an intact body image for the wearer. Their analysis identified aesthetic and weight features as the most pertinent prosthetic user needs from a psychological perspective. Consequently, the overarching objective of lower limb prostheses is not only to faithfully replicate the locomotion gait of healthy subjects but also to ensure profound embodiment and acceptance for the user. Emphasis should be particularly placed on the latter, given the substantial rate of prosthesis abandonment in lower limb amputations, estimated to be around 37 %, with this figure escalating as the level of amputation increases [22].

Within the scope of this research, the focus is on individuals grappling with transfemoral amputations (TFAs). The primary objective is to develop a robotic knee prosthesis characterized by versatility and adaptability, aiming to restore a diverse range of knee functionalities. However, preceding the in-depth exploration of the design and experimental validation phases of the prosthetic device, it is imperative to establish a comprehensive foundation. This involves a meticulous examination of the biomechanical requirements inherent in human locomotion and a nuanced identification of the paramount needs of prosthesis users.

The foundational understanding of human biomechanics and user needs serves as a pivotal precursor, ensuring that the robotic knee prostheses seamlessly integrate with the intricacies of human movement. By combining technological aspects with the natural patterns of human biomechanics, the developed prosthesis aspires to not only address the functional aspects but also to address the evolving practical requirements of individuals relying on such prosthetic solutions. This holistic approach underscores the significance of a thorough groundwork to enhance the effectiveness and user experience of robotic knee prostheses for individuals with transfemoral amputations.

1.1 Biomechanics of human locomotion

The human gait constitutes a remarkably intricate maneuver, appearing simplistic due to the harmonized functioning of the musculoskeletal, neuromuscular, and neurocontrol systems within our body. The limbs collectively engage in a synchronized manner, optimizing the efficiency of the gait. The dynamic behavior of each joint undergoes variations contingent upon diverse conditions, including walking speed and the specific terrain traversed. To comprehensively analyze this array of movements, it proves beneficial to categorize pertinent reference planes and corresponding movements. Figure 1.1 delineates the three foundational anatomical planes: sagittal, coronal, and transverse.

The coronal plane is commonly denoted as the frontal plane, while the transverse plane is equivalently labeled as the horizontal plane. Furthermore, if the sagittal plane serves to partition the human body into two symmetrical halves, it is appropriately identified as the medial plane. In tandem with these anatomical considerations, it becomes imperative to introduce a set of human movements that will be extensively employed throughout this study. Frequently, these movements are articulated in relation to specific anatomical landmarks or aligned with distinct anatomical planes. Figure 1.2 provides a visual representation of the most pertinent movements, essential for characterizing and elucidating various aspects of human motion. In the examination of lower extremities, it becomes imperative to describe three joints: the hip, knee, and ankle, each of which plays a distinct yet integral

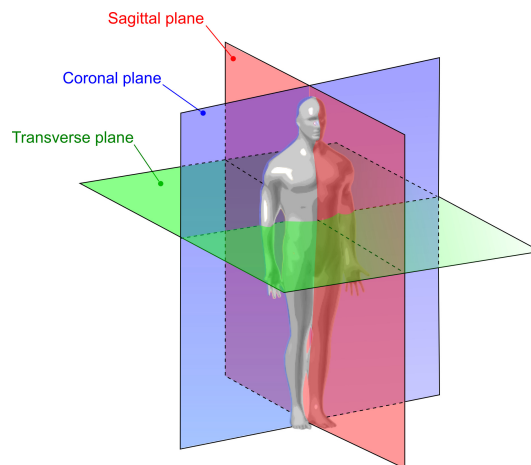


Fig. 1.1 Planes of human anatomy [23]: sagittal in red, coronal in blue and transverse in green.

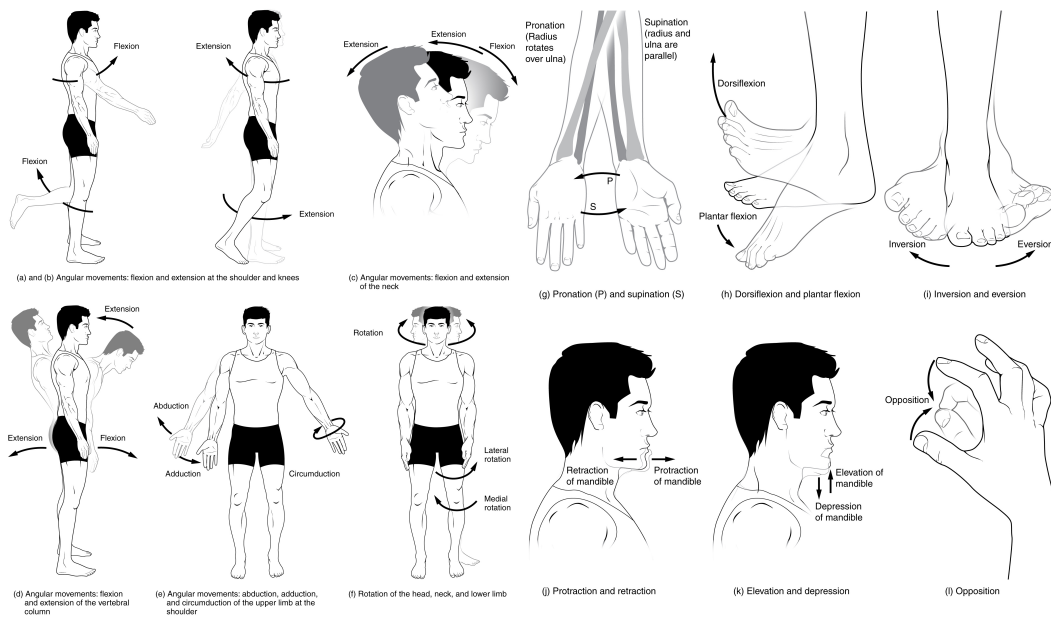


Fig. 1.2 Fundamental movements of the human body [24]

role in human biomechanics. The hip, characterized as a multiaxial, synovial socket, and ball joint, establishes a crucial connection between the acetabulum of the hip bone and the head of the femur [24]. Its primary function involves shouldering the weight of the body, demanding considerable strength and stability, particularly during weight-bearing activities such as standing and walking. Diverging from the expansive range of motion observed in the shoulder joint, the hip exhibits a more constrained movement spectrum, strategically limited to ensure optimal stability.

Moving to the knee joint, it emerges as the largest joint within the human body, featuring three distinct articulations. Functioning predominantly as a hinge joint, the knee facilitates the flexion and extension of the leg. While possessing the capability to bear weight when fully extended, it also remains susceptible to injuries associated with hyperextension and twisting, underscoring the delicate balance required for its proper functionality.

Concluding with the ankle, a uniaxial hinge joint, it facilitates the crucial movements of dorsiflexion and plantarflexion in the foot. The inversion and eversion movements, pivotal for lateral foot movements, are orchestrated by additional joints interposed between the tarsal bones of the foot. Collectively, these articulations within the lower limbs contribute to the orchestration of human motion, underscoring the significance of comprehending their anatomical and biomechanical intricacies

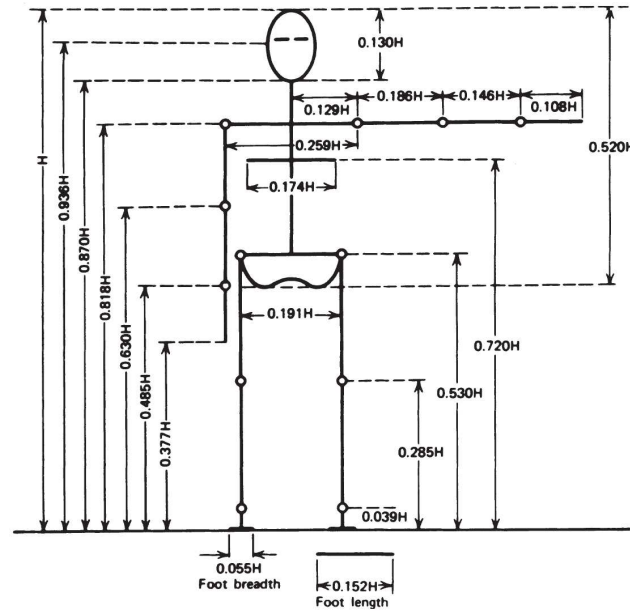


Fig. 1.3 Body segment lengths as function of body height [25]

for a holistic understanding of human locomotion.

1.1.1 Anthropometry

Anthropometry explores the intricate proportions concerning weight and size across various limbs of the human body. Employing systematic measurements, it endeavors to parameterize each limb, anchoring these parameters to either the individual's body weight (W) or height (H). Although primarily leveraged by healthcare professionals such as physiotherapists and orthopedists, anthropometry extends its utility to the design of rehabilitative and assistive devices, including robotic prostheses and exoskeletons. Notably, anthropometric data serves as a foundational basis for device design, allowing for a certain degree of independence from specific user profiles, while still necessitating a level of customization tailored to individual needs.

Considering anthropometric references, two significant sources merit consideration. The work by Tilley *et al.* [26] furnishes anthropometric data normalized to population percentiles, categorized across men, women, and infants. In contrast, Winter [25] contributes insights with data normalized to body height and weight. Figure 1.3 elucidates body segment lengths normalized to overall height, providing a visual representation of these anthropometric considerations. This multifaceted approach

to anthropometric data, encompassing diverse sources and normalization methodologies, underscores its pivotal role in enhancing our understanding of human physical characteristics and informing the design processes of biomechanically-oriented devices. The two datasets are slightly different, being [26] slightly underestimating with respect to [25].

1.1.2 Walking and Stairs Gait cycles

This section comprehensively delves into the walking and stairs gait cycles, as they emerge as prevalent and crucial sets of movements within the spectrum of locomotion [14, 27]. The walking gait constitutes a cyclical sequence of movements predominantly unfolding in the sagittal plane. It is pertinent to note that achieving perfect symmetry in locomotion, especially among individuals affected by lower limb amputations, remains a challenging endeavor [28]. Despite this inherent asymmetry, the prosthetic device's overarching objective remains the establishment of a symmetrical gait cycle, aiming to mitigate compensatory mechanisms and reinstate a healthy gait pattern.

With a focus on the walking gait cycle, a conventional categorization delineates two distinct phases: stance and swing, as elucidated in Figure 1.4. The demarcation between these phases is notably dictated by the loading condition. In the stance phase, the leg undertakes the weight-bearing responsibility, ideally accounting for approximately 50% of the total body weight. This phase commences at heel strike, characterized by the body in a double support stance, with the foot heel in contact

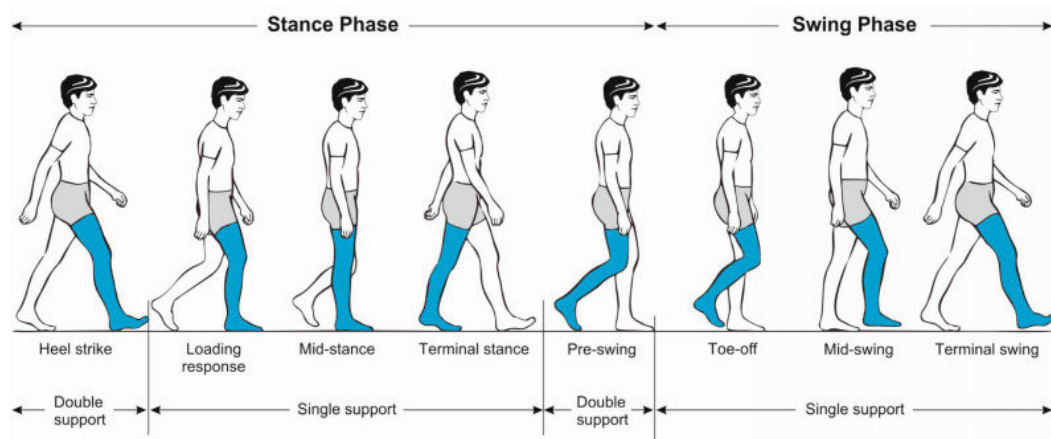


Fig. 1.4 Phases of the walking gait cycle [29]

with the ground. Transitioning into single support, the ankle undergoes plantarflexion, the knee assumes a partial flexed position under the body weight, and the hip initiates extension. Mid-stance marks the subsequent phase, with the ankle gradually dorsiflexing as both the knee and hip extend. Terminal stance concludes the single support phase, with the heel lifted off the ground and the leg positioned behind the trunk. The final portion of the stance, known as pre-swing, witnesses rapid ankle plantarflexion, concomitant with flexion of the knee and hip. On the other hand, the initiation of the swing phase unfolds at toe-off, marked by a plantarflexed ankle and simultaneous flexion of both the knee and hip. Progressing into mid-swing, the ankle maintains a neutral position, and the knee undergoes extension while the hip attains its maximal degree of flexion. The terminal swing phase is fundamentally characterized by the full extension of the knee, setting the stage for the subsequent stance phase.

In contrast, the gait cycle during stair negotiation introduces a heightened level of variability. Specifically, three distinctive gait cycles are discerned: (i) single-step step-to, (ii) double-step step-to, and (iii) step-over-step [30]. The step-over-step configuration demands notably greater effort, as the affected side is more actively engaged in the movement, aligning with the conventional gait pattern observed in healthy individuals. In the ascending phase, the requirement for substantial stance knee extension prevails, often necessitating the exertion of hip and knee torques. From the perspective of individuals with amputations, compensatory mechanisms may attenuate the challenges during the stance phase [31], while the swing phase predominantly relies on the inertial coupling between the thigh and shank, i.e., the prosthetic device. Notably distinct from the swing cycle in walking, the leg dynamics is smaller, rendering the achievement of ballistic movements challenging.

Conversely, the two other ascending gait cycles on stairs prove energetically more viable for amputees, as they predominantly rely on the healthy limb for negotiation. These cycles demand considerably lower stance knee extension, and conventional prosthetic devices typically achieve minimal flexion/extension of the knee joint. Descending stairs, on average, is an energetically passive task, characterized by a higher braking or damping power compared to walking. The conventional amputee gait during stair descent typically manifests as either single-step step-to or step-over-step. The distinctive biomechanical demands of stair negotiation underscore the necessity for nuanced considerations in the design and application of assistive devices for individuals with lower limb amputations. Targeting the knee joint, it

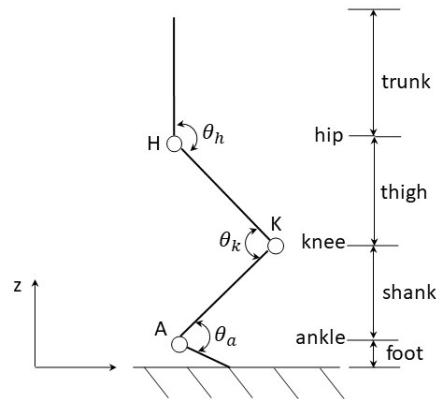


Fig. 1.5 Leg link model highlighting relevant joints and limbs.

becomes imperative to elucidate both its kinematic and kinetic behaviors, providing a foundational understanding essential for the replication of its functioning. To extrapolate pertinent knee joint quantities, a rigid-link model of the leg, as depicted in Figure 1.5, emerges as a valuable tool. Traditionally, the extraction of kinematic quantities, encompassing angles and angular speed, is facilitated through either direct measurements, employing tools such as goniometers and inclinometers, or sophisticated camera-based methodologies that leverage three-dimensional marker tracking in space. In parallel, the derivation of kinetic quantities, such as joint torque and power, necessitates direct force measurements, entailing the resolution of an inverse dynamic model. The dataset curated by Lencioni *et. al* serves as a valuable resource for gathering pertinent kinetic and kinematic measurements specific to the knee joint. Figure 1.6 elucidates the knee angle and external moment, delineating their variations over task duration for both walking in three distinct conditions and step ascending and descending in a step-over fashion. A discerning observation reveals noteworthy disparities in torque requirements at the knee joint between walking and stair negotiation gait cycles. Specifically, the extension torque demands during stairs ascending surpass those of normal walking by more than double. This contrast extends to the assessment of power output at the knee joint, as depicted in Figure 1.7, emphasizing the distinct biomechanical demands imposed during these diverse locomotor tasks. In essence, this comprehensive examination of knee joint kinematics and kinetics, backed by empirical measurements from Lencioni *et. al*, not only unveils the complexities inherent in different locomotor tasks but also lays a solid foundation for informed endeavors in replicating and comprehending

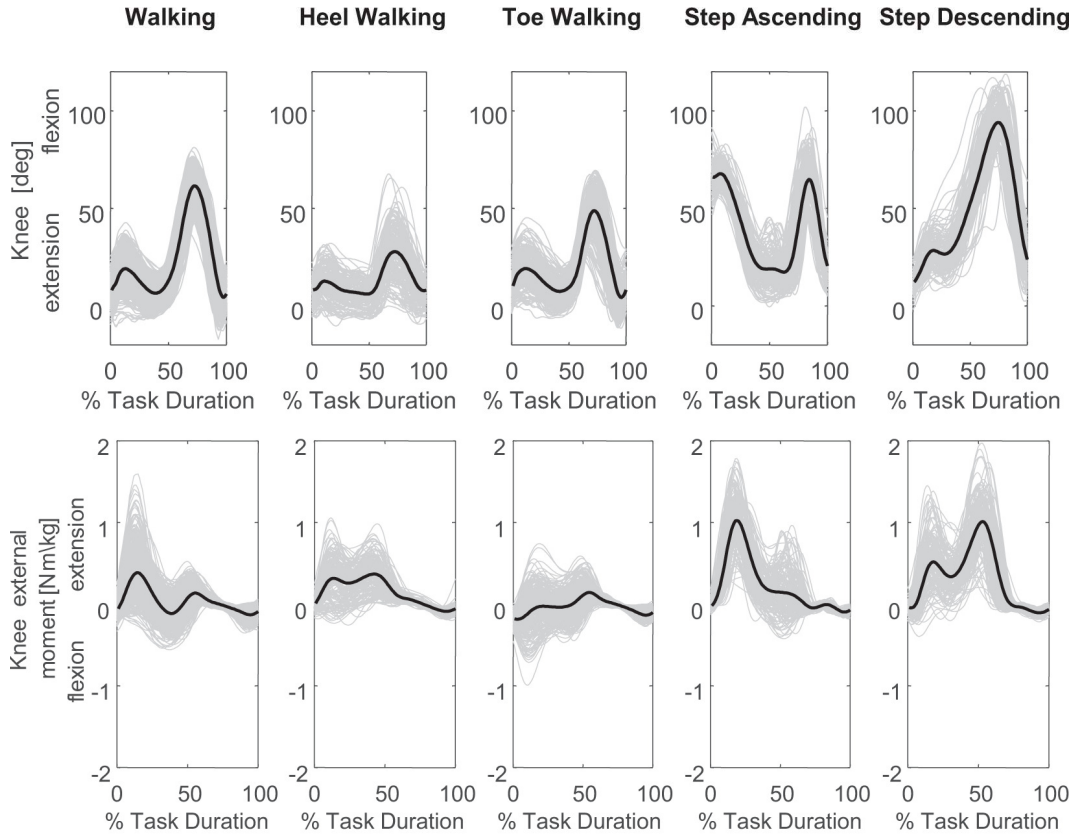


Fig. 1.6 Locomotion activities at the knee joint. (top) angle, (bottom) normalized moment. [32]

the intricate functioning of the knee joint. The intricate dynamics of the knee joint, acting predominantly as a brake or damper during level walking and exhibiting active power exertion only in the stance loading phase, introduce complexities in the design of a robotic prosthesis. Notably, the average power output during normal walking, measured at -0.18 W/kg with a peak active power of 0.27 W/kg, suggests the feasibility of navigating typical walking tasks without injecting active power at the knee joint. However, the scenario shifts during stairs descending, where the power dynamics also tilt toward the negative, with a peak power nearly four times greater than that observed in walking, despite maintaining a comparable knee angular speed.

Conversely, the biomechanics of stairs ascending pose a distinct challenge, constituting a predominantly active task with consistently positive joint power, except for the final phase, terminal extension. This multifaceted landscape of power dynamics across different locomotor activities underscores the diverse biomechanical demands

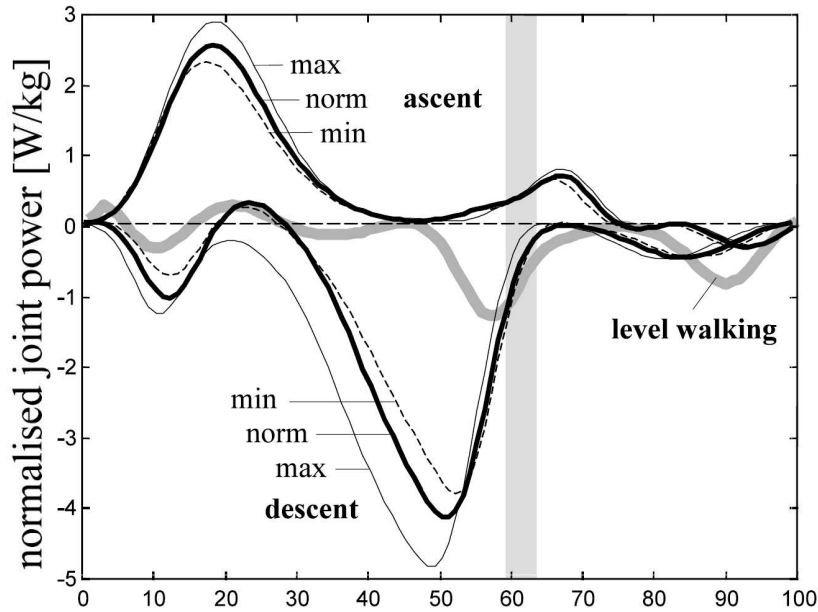


Fig. 1.7 Knee joint normalised power during walking, stairs ascending and descending as function of gait cycle [33]

imposed on the knee joint. Navigating the requirements of both level walking and stairs negotiation in the design of a robotic prosthesis proves to be a non-trivial endeavor. The wide-ranging biomechanical prerequisites, spanning power, torque, and angular speed at the knee joint, underscore the need for a sophisticated and adaptive design that can seamlessly accommodate various locomotor tasks. Addressing these challenges requires a meticulous and comprehensive approach to ensure the development of a versatile robotic prosthesis capable of meeting the diverse demands of real-world mobility scenarios.

1.1.3 User Needs

An imperative stride in the pursuit of effective prosthetic design involves user needs research, a pivotal focus aimed at translating these needs into tangible prosthetic design requirements. The significance of centering design efforts around the user has been underscored in previous studies [34–36], emphasizing that the integration of human and technical considerations can yield solutions markedly different from those prioritizing engineering requirements [35]. Within the prosthetic field, several domains, spanning medical, engineering, design, and psychological aspects, neces-

sitates a comprehensive understanding to inform design decisions. Initiating user research involves leveraging tools such as surveys and focus groups to define and explore user needs [35, 37, 38]. Notably, the acceptance of technology emerges as a critical aspect in prosthetic devices [39]. Numerous studies have illustrated that low prosthetic acceptance often leads to device abandonment, a phenomenon attributed to technical challenges impeding Activities of Daily Living (ADLs), issues related to reliability and responsiveness, and deficiencies in human-technology integration, particularly in terms of embodiment [40, 41]. Addressing these concerns and enhancing prosthetic embodiment becomes integral to the design strategy, aiming to boost technological acceptance and overall user satisfaction [42, 43]. Beyond these considerations, the literature has predominantly scrutinized user needs in the context of lower limb prostheses, as elucidated by [44]. Four distinct groups of needs have been identified: functional, psychological-cognitive, ergonomic, and others. The overarching desire expressed by users is an independent life with full reintegration into social activities. Prosthetic features deemed crucial for such enhancements include efficiency, versatility, gait stability, and support for various activities such as sitting and standing. Additional factors such as mobility level, age, and gender have also surfaced as influential elements shaping user needs. A noteworthy multidisciplinary study by Fanciullacci *et al.* [36] delves into the user-centered design of novel transfemoral prostheses. Their survey data, tailored for unilateral transfemoral amputees (TFAs), highlights user prioritization, with reliability, comfort, weight, and cost emerging in that order. Examining the desired functionality of the prosthesis, stability takes precedence, followed by adaptability to varying gait speeds and different terrain conditions, including uneven terrains, slopes, and stairs. Active assistance gains significance, particularly if it aids users in avoiding falls, especially during demanding tasks such as stair and slope ascent/descent, sit-to-stand maneuvers, and walking at elevated velocities. This multidimensional exploration underscores the complexity of user needs, advocating for a holistic and user-centric approach in the design of transfemoral prostheses.

1.2 Robotic Knee Prostheses

Considering transfemoral amputations, the prosthetic device constitutes a triad of essential components: the socket, the knee prosthesis, and the ankle-foot prosthesis.

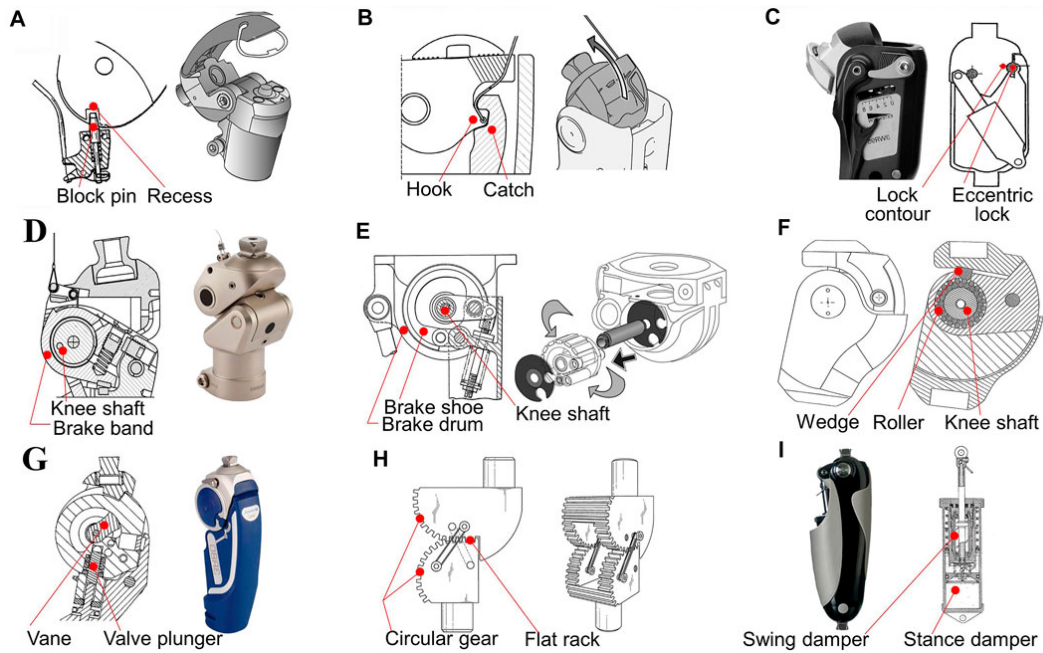


Fig. 1.8 Examples of monocentric knee designs [45]

The socket serves as the crucial interface between the residual limb and the robotic device. Due to the diverse spectrum of amputations, ranging from knee to hip disarticulations, and the different sizes of residual limbs, the socket demands high levels of customization. Typically, the design and manufacturing of the socket are delegated to prosthetists, recognizing the nature of tailoring it to the specific needs and anatomical nuances of individual users. Conversely, the knee and ankle prostheses emerge as focal points in the landscape of assistive robotics, where their design not only exhibits considerable diversity but also targets distinct functionalities within the realm of human locomotion. This section aims to shed light on knee prostheses, given their central significance to the doctoral work. A fundamental classification of knee prostheses hinges on three key aspects: (i) kinematic behavior; (ii) adaptability to different locomotion conditions, and (iii) the ability to exert active power toward the knee joint. Delving into the kinematic behavior, knee devices can be categorized as either monocentric or polycentric. Monocentric designs feature a fixed knee joint center of rotation (examples highlighted in Figure 1.8), a prevalent configuration adopted by many devices. Conversely, polycentric knees exhibit a variable instantaneous center of rotation (ICR). Although the polycentric design often involves a more complex and expensive 4-bar linkage mechanism, it confers inherent stability during loading phases, as the ICR is elevated and posterior when the knee

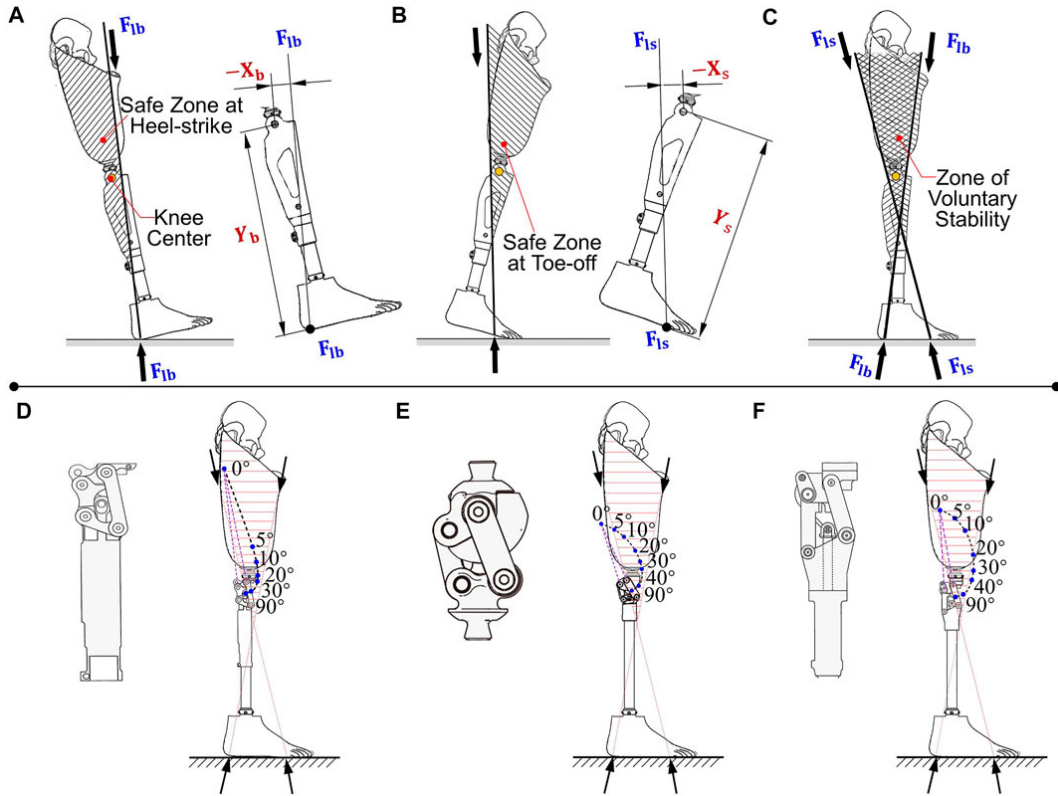


Fig. 1.9 Examples of polycentric knee designs and their inherent stability [45]. The instantaneous rotation center is posterior and above the conventional knee joint center.

is fully extended [46], as depicted in Figure 1.9. The second classification revolves around the controllability of the knee prosthesis. Mechanical knee prostheses, devoid of controllability, furnish a fixed response to ballistic leg movements initiated by the hip joint [47]. In contrast, controllability and adaptability are typically associated with the integration of an electronic control unit (ECU) in the prosthesis, endowing it with an "intelligent" attribute. The ECU's role is pivotal in tailoring the knee's behavior in accordance with the specific task at hand. Furthermore, the inclusion of an actuation unit augments the device's versatility. The final classification centers on the prosthesis's ability to provide active power at the knee joint. While the presence of an actuator doesn't guarantee the injection of active power, certain knee prostheses, like variable damping knee prostheses, employ a motor to alter the leg's passive behavior without the capability to provide active assistance. Navigating the rich terminology and classifications prevalent in the literature, it becomes imperative to establish a clear lexicon for subsequent discussions. Devices capable of actively assisting at the knee joint will be referred to as "active knee prostheses". In contrast,

if a prosthesis lacks this capacity, irrespective of actuation type or design, it will be termed a "passive knee prosthesis." This refined terminology sets the stage for a coherent exploration of the intricacies related to knee prostheses and their pivotal role in assistive robotics.

1.2.1 Passive prostheses

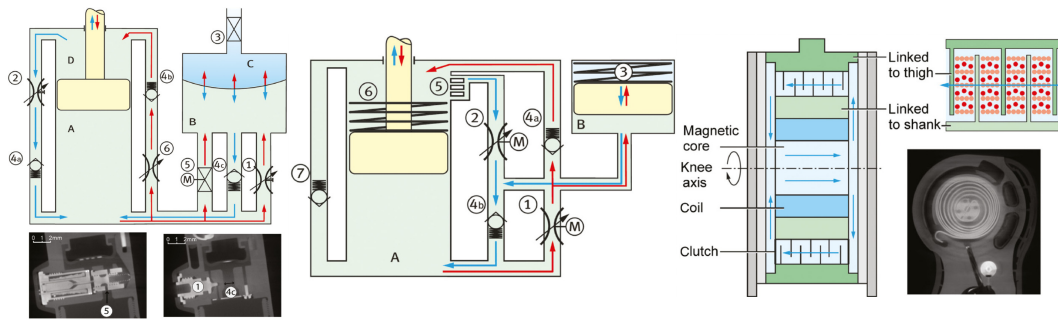


Fig. 1.10 *Left*: Plié 3 scheme of hydraulics and pneumatics (blue: flow during extension, red: flow during flexion) along with a computed tomography; *center*: C-Leg 4 scheme of hydraulics (blue: flow during extension, red: flow during flexion); *bottom*: Rheo Knee 3 functional principle (left, blue arrows: magnetic flux) with a magnetorheological clutch (top right, red: magnetic particles, orange: particle chains) and computed tomography of the torsion steel spring for extension assistance.[48]

Focusing on passive prosthetic devices, two distinct categories can be identified. Firstly, there are purely passive devices, referred to in the context of this work as "mechanical knees". These devices lack the ability to dynamically adapt their behavior throughout the locomotion gait. Their damping response remains fixed, typically tuned to accommodate the swing phase, while stance support is often achieved through locking mechanisms. Notably, these mechanical knees eliminate the need for electronic components, batteries, or motors, thereby contributing to reduced costs, weight, and maintenance of the prosthesis. Their resistive behavior is either constant or dictated by mechanical factors, such as the design principles underlying single-axis or polycentric prosthetic knees [49]. Some models within this category incorporate passive pneumatic or hydraulic cylinders to provide stance or swing control [50, 51]. While offering advantages in terms of weight, size, and cost, the resulting gait from these prostheses deviates significantly from a natural gait, proving suboptimal for activities beyond fixed-speed level ground walking [52]. On the other end of the spectrum, a more adaptive damping behavior character-

izes variable damping knees, commonly known as "microprocessor knees" (MPKs). These prosthetic devices integrate a microprocessor that actively controls the amount of resistance, thereby enhancing gait performance and adaptability. Unlike their purely passive counterparts, MPKs exhibit partial adaptability to different walking speeds, downward slopes, and activities governed by energy-dissipative or neutral behavior, such as stair descent and sitting down. The fluid modulation in MPKs can take the form of air or oil, distinguishing between pneumatic and hydraulic/magnetorheologic MPKs, respectively. Dominating the lower limb prosthetic market, examples include the *C-Leg* [53], *Genium* [54], *Genium X3* [55], and *Rheo Knee* [56]. The schematic of the most representative ones is presented in Figure 1.10. Beyond heightened controllability in movements compared to mechanical knees, MPKs offer advantages in terms of weight, size, and noise reduction. They contribute to decreased hip effort on the affected side, improved gait symmetry, and reduced metabolic expenditure, ultimately enhancing safety and support for the user [49]. However, despite the improved performance of MPKs, they encounter challenges in tasks requiring positive power at the knee joint or when facing gait perturbations such as stumbling or scuffing. Negotiating stairs and ramps ascending, as well as performing sit-to-stand maneuvers, critically test the capabilities of MPKs. In response to these challenges, partially or completely active robotic knee prostheses have emerged, aiming to address energetically positive locomotion tasks and enhance the overall robustness of the prosthesis. Leveraging the ability to inject active power at the joint, these advancements seek to further improve the versatility and adaptability of prosthetic devices.

1.2.2 Active prostheses

Active prosthetic knees, distinguished by their electronic control and the capacity to perform positive mechanical work through an integrated actuation unit geared to the knee joint [57], represent a pinnacle of versatility within the prosthetic landscape. Actuators geared to the knee joint allow to exert positive power, heightening the capabilities of robotic prostheses. The most adopted technologies are electro-mechanical (e.g., lead screws, ball screws, gearboxes) and electro-hydraulic (e.g., pumps) [58–61]. However, despite their potential, the current market predominantly embraces passive and variable damping prosthetic knees, underscoring the limited clinical acceptance of powered prostheses [62]. This reluctance may be attributed

to a perceived lack of maturity, wherein the advantages of active knees have yet to offset their drawbacks linked directly to existing technological constraints. In comparison to their passive counterparts, active prosthetic knees tend to be significantly heavier, bulkier, and exhibit shorter battery life. The elevated weight of these devices adversely influences both biomechanical [63] and clinical outcomes, impacting gait asymmetry [64], hip effort [65], socket stability, and increasing metabolic costs [66].

As a response to these challenges, a recent surge in research endeavors seeks to combine the strengths of passive and active technologies into a unified concept known as the passive-active knee (PAK) or, alternatively, the *partially powered knee*. These prosthetic innovations aim to mitigate the deficits associated with fully active knees, albeit at the cost of sacrificing some of their active power capabilities. The primary objective is to provide partial support to patients, either by powering only specific activities or by undersizing the system to offer a percentage of the required support. Consequently, partially powered knees tend to be lighter, more compact, and boast an extended battery life. However, they fall short of fully replicating the biomechanical functions of the missing biological limb. A conceptual representation of the hypothetical performance of prostheses across various categories is illustrated in Figure 1.11, with desired attributes depicted along the vertical axis and typical consequential negative attributes along the horizontal axis. This graphical representation, adapted from Adamczyk [67], has been expanded to incorporate the emergent category of partially powered prostheses. Positioned at the top left of the graph is the ideal prosthesis, exhibiting the highest performance and acceptance rate. Considering partially powered knee design, diverse passive-active paradigms have undergone exploration and experimentation. While complete decoupling of these features might not always be feasible, a crucial endeavor involves quantifying the distinctive contributions each component makes to the overall functionality of the prosthesis.

Within this paradigm, the passive side assumes a pivotal role, particularly in scenarios such as level ground walking, where the knee tends to exhibit energetic passivity on average. The control of stance knee flexion is achieved through the integration of resistant elements, encompassing a spectrum from springs and elastic rubber to robust bumpers. Additionally, absorbers featuring hydraulic, magnetorheological, or electric dampers contribute to the prosthetic functionality. This interplay between passive and active elements delineates the relevant design considerations integral to the optimization of partially powered knees for diverse locomotive tasks.

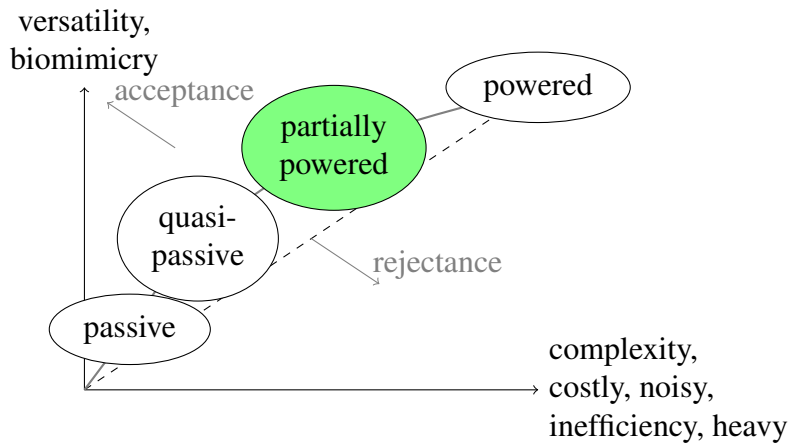


Fig. 1.11 Qualitative representation of prosthetic knee performance sorted by type.

In the context of prosthetic knee design, achieving stance knee extension through passive elements is not feasible, necessitating the incorporation of active power at the knee joint. Contrarily, the walking swing resistance exhibits a considerably lower magnitude compared to the stance phase, allowing for passive modulation of prosthetic swing behavior. This modulation is achieved by leveraging the primary influence of hip dynamic contribution on knee kinematics, facilitating passive adjustment to variations in walking speed, user height, and weight. In terms of damper technology, hydraulic components emerge as the preferred choice due to their inherent advantages. Hydraulic dampers, comprising pressurized reservoirs, servo-valves, and hydraulic cylinders, leverage incompressible hydraulic oil to produce substantial force, surpassing the capabilities of air-based systems. The damping force is finely tuned by adjusting the cylinder flow area, granting the ability to modulate the damping behavior at the knee joint, as represented by the C-LEG schematic of Figure 1.10. However, hydraulic dampers are not without challenges, with fluid leakages being a notable concern.

An alternative damping technology gaining traction is the use of magnetorheological (MR) dampers. Functioning as semi-active dampers, MR dampers employ magnetorheological fluid (MRF) that undergoes rapid transitions from liquid to semisolid states under the influence of an external magnetic field. This feature enables precise damping and force control. MR dampers exhibit simplicity in design, compact volumes, low energy requirements, and high resistance, making them a compelling choice.

Electric dampers, as actuation units employing electric motors and transmission systems, offer a different approach to energy dissipation at the knee joint. Distinguished by their fast dynamic behavior and broad control bandwidth, electric dampers are characterized by limited backdrivability compared to semi-active dampers. However, they introduce potential advantages, especially in hybrid prosthetic knee designs, where they can contribute not only to power dissipation but also to power generation.

On the active side, an examination of kinematic behavior and actuation topology provides valuable insights. Actuation units can be linear or rotary, each presenting distinct advantages. Linear actuation, often accompanied by a transmission system, facilitates knee angle-dependent transmission ratios, aiding in locomotion activities. Rotary actuation units, while not mandating a transmission system, frequently employ reduction stages to address torque and power requirements.

The transmission of motion between actuator and knee joint introduces three options: rigid, series elastic (SE), and parallel elastic (PE) motion transmissions. Rigid transmission is the most prevalent, while SE transmission incorporates a spring element to mitigate impacts and vibrations at heel strike and serve as a torque transducer at the knee joint. PE transmission integrates a spring element in parallel to aid actuation in one direction, emphasizing energy storage capabilities.

Actuation topology, specifically the transmission system, varies among mechanical, hydraulic, and pneumatic systems. Mechanical transmission systems, encompassing lead and ball screws, transform rotary motion into linear motion, necessitating additional transmission means. Hydraulic transmissions leverage a geared pump and hydraulic actuator, providing a quasi-rigid coupling that facilitates large bandwidth dynamics and precise motion control. Pneumatic transmissions utilize air for motion transmission, albeit with a higher susceptibility to leakages compared to hydraulic systems. A notable concern across all actuation systems is acoustic noise, primarily emanating from servomotors operating within the torque and speed ranges relevant to human ambulation. While measures such as limiting the transmission ratio, selecting low-noise transmission systems, and incorporating insulation coverings can mitigate noise emission, it remains a pertinent consideration in the design of powered prostheses. This comprehensive overview underscores the multifaceted considerations in the engineering of prosthetic knee designs, balancing passive and active elements to optimize performance across diverse locomotive tasks.

	Type	Name, Institute, Country	Year	Reference
1	K	C-Leg 4, Ottobock	2022	[53]
2	K	Power Knee II, Ossur	2012	[69]
3	K	ECT, Vanderbilt, USA	2022	[61]
4	K	CSEA, MIT, USA	2014	[70]
5	K	SA, Vanderbilt, USA	2018	[71]
6	K	SCSA, Vanderbilt, USA	2020	[72]
7	K	RIC, Chicago, USA	2015	[73]
8	K-KA	BerkeleyKnee, Berkeley, USA	2009	[74, 75]
9	K	AMKR, Minas Gerais, BRASIL	2018	[76, 77]
10	K	MREC, Inha, KOREA	2016	[78]
11	KA	CYBERLEGs, Belgium	2015	[79]
12	K	Dual Motor, IIT, Italy	2022	[80]
13	K	EHA actuator, IIT, Italy	2022	[60]
14	K	HybridUtah, Utah, USA	2018	[81]
15	KA	Bionic Leg, Utah, USA	2022	[58]

Table 1.1 Lower limb prosthetic knees

Literature review The literature review is based on one our works [68]. The search strategy identified 13 different knee prostheses, listed in Table 1.1. Commercial devices are also included in the table, as reference baseline for comparison. To obtain an overview and to compare the prostheses identified with the biomechanical requirements defined in Section 2.2, Table 1.2 summarizes the actuation technologies and their characteristics. The main attributes presented in columns are:

- *Legs*: Name of the devices;
- *Type*: this field presents which device is knee joint only (K) or ankle-knee (AK) devices;
- *Actuation*: this column describes whether the actuator is directly coupled to the knee joint (Stiff) or with Series Elastic actuators (SEA), variable stiffness actuator (VSA) and torque-sensitive joint (T-S);
- *Power supply*: if the system is tethered (T) or a standalone device (SA);
- *Actuator characteristics*: this field is constituted by four subcolumns concerning the hardware configuration of the actuation, the power rating of the

actuator, the moment rating at the knee joint, and the reflected inertia during swing if the motor is engaged. In case there is a nonlinear transmission between the motor shaft and the knee joint, the displayed transmission ratio is the highest during swing;

- *Prosthesis characteristics*: it is subdivided into six different attributes concerning the knee joint's range of motion (RoM), weight, height, distance between the pyramid-to-knee joint axis, the device's noise level during walking and cost. Noise column is estimated because generally, there is no standard characterization for prosthesis noise. Scores range from mere hydraulic component noise to motor backdriving while walking, culminating in using the motor for braking. Cost, like noise, is also only estimated and based on the authors' opinion. Low grade means less components, off-the-shelf components or not-elaborated actuators;
- *Active tasks*: this field highlights whether swing phase walking, swing flexion or stance extension during stair climbing or sit-to-stand transition are actively assisted by the prosthesis;
- *Semi-powered*: this field is constituted by two subcolumns based on the definitions of hybrid and partially powered prosthetic knees;
- *Status*: this field illustrates at which stage the prosthesis between commercially available (CA) or tested on a bench (TB);
- *Experiments*: this column presents, based on published data, whether the prosthetic knee has been already tested with an able-body adapter (AB) or amputees (A) or none of the above (N).

The presence of a dash means that the information could not be obtained neither from the respective publications nor calculated by the author. Moreover, a spectrum of colours ranging from green to red indicates the goodness of each feature. Cells marked with a background colour but represented with a dash signify insufficient information provided by the authors to quantitatively assess the feature goodness. However, it may still be feasible to make an estimate for some of them, based on best practice (e.g., Össur has not disclosed the gear ratio or the specific motor employed in the Power Knee. Nonetheless, it is conceivable to infer that the reflected inertia is

notably high, resulting in a red-colored cell, even if precise analytical computations are not feasible).

The majority of these devices are characterized by a stiff actuation technology, while just three feature serial elastic elements (SEA). Out of these, two of them comprise a variable stiffness actuator (VSA) or a torque-sensitive joint (T-S) in which the spring is able to vary the transmission ratio at the knee joint. Few transfemoral prostheses are tethered to either an external power supply or a laptop or require wearing a backpack. Whereas the majority are self-contained, it demonstrates the increasing trend of developing devices with higher Technology Readiness Level (TRL) instead of creating just platform for testing control strategies or use these devices as test benches to outline the pros and cons of the chosen actuator technology. The majority of the motor choices are Maxon BLDC motors that span power rating between 70 W to 260 W. Only one design considers a custom motor. Depending on the transmission choice, power ratings vary between 60 W to 360 W while knee active moment rating is widely spread between 5 N · m to 125 N · m. Four devices are able to provide active peak moments of less than 30 N · m, three in-between 30 N · m and 75 N · m, and five legs are capable of providing more than 75 N · m. Six legs have a mechanism capable of changing the actuation configuration with either clutches or actively variable transmission (AVT).

In reference to the technological criteria presented in Section 2.2, addressing versatility involves considerations such as braking torque, active torque, and the joint's high transparency. Although all devices excel in meeting requirements for braking and stance support, these standard features of quasi-passive prostheses have been excluded from the table for clarity. The comparison employs the 50th percentile of females to define prosthesis properties. Therefore, legs capable of exerting more than 68.75 N · m deliver the required torque for tasks like sit-to-stand and stair stance extension. Among these, five legs fall into the hybrid category, enabling motor disengagement or variable actuation configurations. Specifically, the *RIC* and the *Hybrid Utah* legs achieve full disengagement, resulting in zero reflected inertia and enabling passive walking, thereby fulfilling battery life requirements. The *Bionic Leg* although nearing the upper bound defined in Section 2.2 with a maximum swing reflected inertia of 356 kg · cm², can also walk passively. In contrast, the *Dual Motor*, due to its consistently engaged initial reduction stage, necessitates the motor for walking, impacting the device's battery life. Five legs are partially powered, with only the *MREC* leg focusing on partially support for both stairs and sit-to-stand.

Others just consider swing phase during stair climbing. The very high value of the *MREC* reflected inertia leads into the impossibility to walk passively. *SA* and *SCSA* have a reflected inertia around the 10 % of the upper bound previously defined ensuring the possibility to walk passively. *AMKR* provides more torque supporting also the heavier task while it disengages the motor at walking allowing to walk passively. *ECT* and *CSEA* focus on swing support while the motor is always engaged. Notably, while low impedance (high joint transparency) is a crucial requirement, only the *ECT* knee [61] characterized leg impedance with a value of 0.5 Nms/rad.

Additional user requirements for ensuring user satisfaction include possessing an anthropometric shape with a perceived weight that is low. Hence, five systems can fulfil the 1st female percentile, four between the 1st and the 50th, two between 50th and the 99th, and only one supports more than the 99th female percentile weight. Comparing the height of the 1st female percentile physiological shank to the devices shows that all the prosthetic devices are smaller except the *ECT* device (360 mm). Furthermore, the distance pyramid-to-knee joint cover an important role for the possibility to fit the prosthesis in a large TF amputees. This length is directly influenced by the design choice and the type of actuation since having the actuator at the knee joint shifts this distance at least half of the motor radius. Even if it is a relevant parameter, only the *Bionic Leg* claims that this distance is lower than the C-Leg segment (26 mm). For all the other systems the pyramid-to-knee joint distance is graphically estimated. three devices have this distance greater than the Power Knee (45 mm) while the others are between the two commercially available device measurements.

None of them are commercially available. However, most of the groups are clinically evaluating the devices. Subjects testing has been performed for 10 prostheses, eight of them including transfemoral amputees. Figure 1.12 displays the prostheses being investigated and indicates the specific clusters to which they belong.

Low noise, particularly in repetitive tasks such as walking is a user need not so much investigated among the authors. In fact, only *EHA actuator* provide some reference with a test bench in an anechoic chamber with a Sound Pressure Level of 22.5 dBA [82]. For all the other devices a score is given based on whether the walking task is addressed only hydraulically, or the actuator is in backdrivable mode or the actuator is actively used to brake depending on the transmission used.

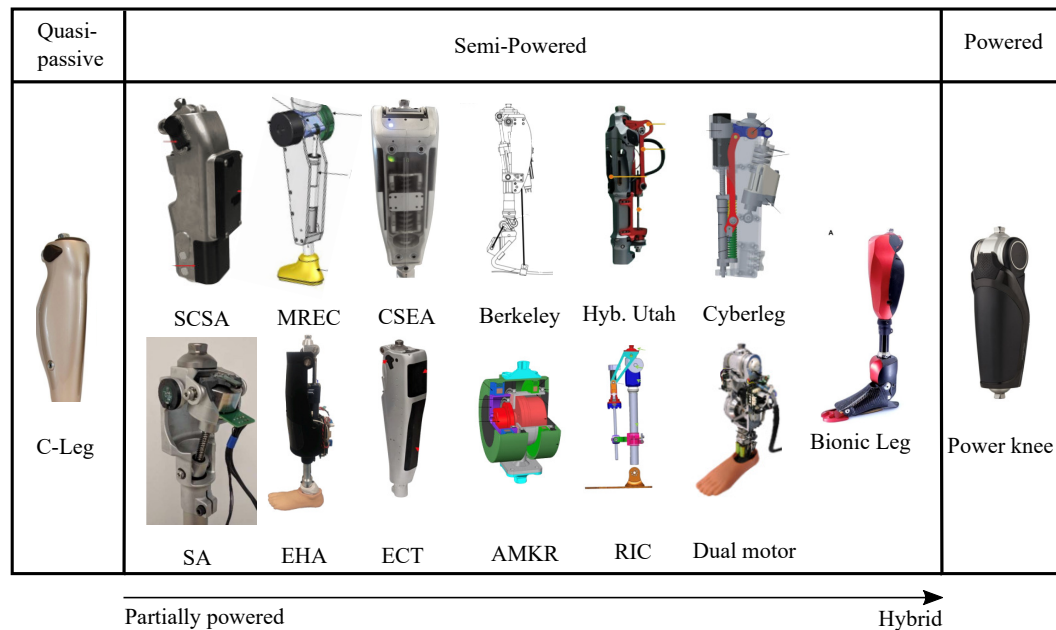


Fig. 1.12 Lower limb prosthetic knees and their clusters.

1.3 Research objective and study motivation

In the pursuit to enhance powered prosthetic legs, this study aims to blend the strengths of passive and active devices. The objective is to develop a knee prosthesis that combines versatility, compactness, and an improved walking experience for individuals with transfemoral amputations. In the scenario of semi-active prosthetics, this research seeks to demonstrate that strategic planning and a well-designed approach can effectively address common challenges faced by amputees.

To achieve this goal, a novel prosthetic layout is proposed. It involves a motor-driven hydraulic valve, emulating a passive damper, primarily intended for tasks requiring substantial braking torque. Additionally, a compact linear electro-hydrostatic actuator (EHA) is incorporated to handle tasks involving higher speed and lower torque. This approach retains the favorable characteristics of passive devices while introducing the capability to utilize active power when needed. The actuator design is focused on providing support during swing phase, aiming to reduce the weight and size of the actuator for improved prosthetic usability. The possibility to apply power at the knee joint offers advantages in various scenarios, such as negotiating stairs and slopes, adjusting walking speed, and facilitating sitting or standing maneuvers. After evaluating the strengths and weaknesses of passive and active prosthetics, it

appears that finding a middle ground is a promising approach. Thus, the central question of this study is, "Is there a trade-off between active power capabilities and prosthesis usability?". To address this question, a multi-objective design framework is employed to develop a partially powered prosthetic device that predominantly behaves like a passive device but can provide a controlled amount of active power as needed. Experimental tests are conducted to assess the added value of the active component in terms of movement robustness and adaptation to different activities of daily living.

				Actuator characteristics		Prosthetic characteristics					Active Tasks			Semi-powered							
Legs	Type	Actuation	Power Supply	Mot. & Trans.	Power Rating [W]	Active Moment Rating [N · m]	Reflected Inertia @ swing [kg · cm ²]	RoM [deg]	Mass [kg]	Height [mm]	pyramid-to-knee joint [mm]	Noisiness @ walking	Cost	Walking: Swing	Stairs: Swing Flexion	Stairs: Stance Extension	Sit to Stand	Hybrid	Partially Powered	Status	Experiments
C-Leg	K	Stiff	SA	-	-	-	-	0-130	1.24	289	26	+	+	-	-	-	-	-	-	CA	A
Power Knee	K	SEA	SA	DC motor, Harmonic Drive Gear	-	96	-	0-120	3.1	275	45	+++	+++	X	X	X	X	-	-	CA	A
ECT	K	Stiff	SA	BLDC Maxon EC-4pole 22 90W, Lead screw & planetary gearbox, ECT	75	16	25* [†]	0-130	1.9	360	30 [†]	++	++	-	X	X	X	X	X	TB	A
CSEA	K	SEA	SA	BLDC Maxon EC-4pole 30 200W, parallel clutch, ball screw	200	14	700 [†]	0-120	2.7	285	40 [†]	+++	+++	X	X	-	-	X	X	TB	A
SA	K	Stiff	SA	BLDC Maxon EC-8pole 45 70W, leadscrew	65	5	46* [†]	0-105	1.07	-	-	++	+	-	X	-	-	-	X	TB	A
SCSA	K	Stiff	SA	BLDC Maxon EC-4pole 22 90W, Spur gear & Lead screw	75	7.5	31* [†]	0-130	2.18	278	25 [†]	+	++	-	X	-	-	-	X	TB	A
RIC	K	Stiff	T	BLDC Maxon EC-4pole 22 90 W, Ball screw, motorized clutch	90	85	D	-	1.7	-	-	+	++	-	X	X	X	X	-	TB	AB
BerkeleyKnee	AK	Stiff	SA	BLDC Maxon EC-powermax 30, hydraulic pump	40	-	-	-	4	280	40 [†]	++	++	-	X	-	-	-	X	TB	A
AMKR	K	Stiff	SA	BLDC Maxon EC-7pole 60 100W, harmonic drive, MR clutch	254	40.4	D	-	2	117	45 [†]	+	+	-	X	X	X	X	X	TB	A
MREC	K	Stiff	T	BLDC Maxon EC-1pole 90 260W, planetary gearhead	90	43.4	9356 [†]	0-155	5	313	45 [†]	+++	++	X	X	X	X	-	X	TB	N
Cyberlegs	AK	VSA	T	Motor, ball screw	-	75	-	0-90	-	-	-	++	+++	-	X	X	X	X	-	TB	N
Dual Motor	K	Stiff	T	BLDC Maxon EC-8pole 45 & Koll-Morgen BMS 1712, 2 strain gears	320	75	1810 [†]	0-125	3	257	55 [†]	+++	+++	X	X	X	X	X	-	TB	AB
EHA actuator	K	Stiff	SA	Custom motor, gerotor pump	220	35	-	0-110	2.4	315	50 [†]	++	++	-	X	-	-	-	X	TB	N
Hybrid Utah	K	Stiff	SA	BLDC Maxon EC-4pole 22 120W, roller & timing belt, AVT	-	125	D	0-155	1.7	290	55 [†]	++	+++	-	X	X	X	X	-	TB	A
Bionic Leg	K	T-S	SA	BLDC Maxon EC-4pole 22 120W, Helical gears & ball screw, variable transmission	200	125	356* [†]	0-120	1.2	255	23	+++	+++	-	X	X	X	X	-	TB	A

* computed at the highest transmission ratio at swing

† data calculated/graphically estimated

Table 1.2 K, Knee; AK, Ankle-Knee; Stiff, Stiffness; SEA, Serial elastic Actuator; VSA, Variable stiffness actuator; T-S, torque sensitive actuator; SA, Standalone; T, tethered; PT, Prototype; TB, Testbench; CA, commercially available; N, No; AB, Able-body adapter; A, Amputee;

Chapter 2

Prosthesis Mechatronic Design

This chapter delves into the process of designing and prototyping a fully customized robotic knee prosthesis equipped with active power capabilities. The initial section of this chapter lays the groundwork by defining the biomechanical principles guiding the prosthetic device's design. Subsequently, dedicated sections delve into the designs of both the EHA and hydraulic valve. A specific focus is given to the optimization framework based on a genetic algorithm, used to size the EHA [83]. The chapter culminates in an exploration of the prototype assembly, with a particular emphasis on the design of the structural chassis and aesthetic cover. This meticulous consideration accounts for factors such as embodiment and acoustic noise optimization, ensuring a comprehensive approach to the device's construction.

2.1 Biomechanical Requirements

Weight The initial biomechanical criterion pertains to the intended user weight, with specific reference to the *50th* percentile for men, which coincides with the *63rd* percentile for women, amounting to 78.4 kg. By adopting this approach, the aim is to prevent the development of an excessively large prosthesis [21]. It's crucial to note that while this user weight is considered in the design of mechatronic components, it doesn't imply that the structural chassis of the prosthesis is exclusively tailored to accommodate this particular percentile.

Table 2.1 Gait cycle activities [84–90]

Task	Occurrence/day	Metabolic cost V_{O_2} [ml/(kg·min)]
Walking	1500 steps (TFA)	14.2 (TFA)
	5500 steps (healthy)	9 (healthy)
Stairs ascending	47-66 stairs (healthy)	33.5 (healthy)
Stairs descending	47-66 stairs (healthy)	17 (healthy)
Sit-to-Stand (StS)	33-71 StS (healthy)	3.8 (healthy)

Locomotion activities Table 2.1 summarizes the most relevant tasks and their daily occurrence for healthy and transfemoral amputees (TFAs). Additionally, Figure 2.1 presents the torque-speed requirements at the knee joint considering level walking and stairs ascending and descending. Level walking has the highest occurrence per day and therefore should be prioritized. The task of self-selected speed level walking primarily involves the knee joint exhibiting a dissipative behavior. Hence, it is envisioned that the passive aspect of the prosthetic device will handle this task efficiently. Specifically, the device needs to modulate its dissipative behavior to effectively address both the swing and stance phases. Achieving this modulation entails maintaining low impedance during the swing phase and high impedance during the stance phase, falling within the ranges of $1 \text{ N} \cdot \text{m} \cdot \text{s}$ and $100 \text{ N} \cdot \text{m} \cdot \text{s}$, respectively [91]. To facilitate this modulation, a flow control valve is proposed, serving to transition between these two conditions. The dynamic behavior of the valve can further be leveraged to manage the nonlinear transmission ratio. This approach operates under the assumption that the hip joint drives locomotion, steering the knee joint through a ballistic movement. Consequently, active knee extension during mid-stance becomes impractical. However, this limitation can be alleviated through strategies such as heel striking, loading the leg with a fully extended knee, and incorporating a slight pelvic tilt.

Nevertheless, there are instances where active power can enhance the passive behavior. For example, walking at slow speeds, where the ballistic motion initiated by the hip may be insufficient to achieve adequate knee flexion, could lead to reduced toe clearance. Introducing active power in such scenarios can make a crucial difference in preventing falls or scuffing [92]. Conversely, stairs ascent proves to be a challenging task to handle passively. As previously discussed in Chapter 1, stair

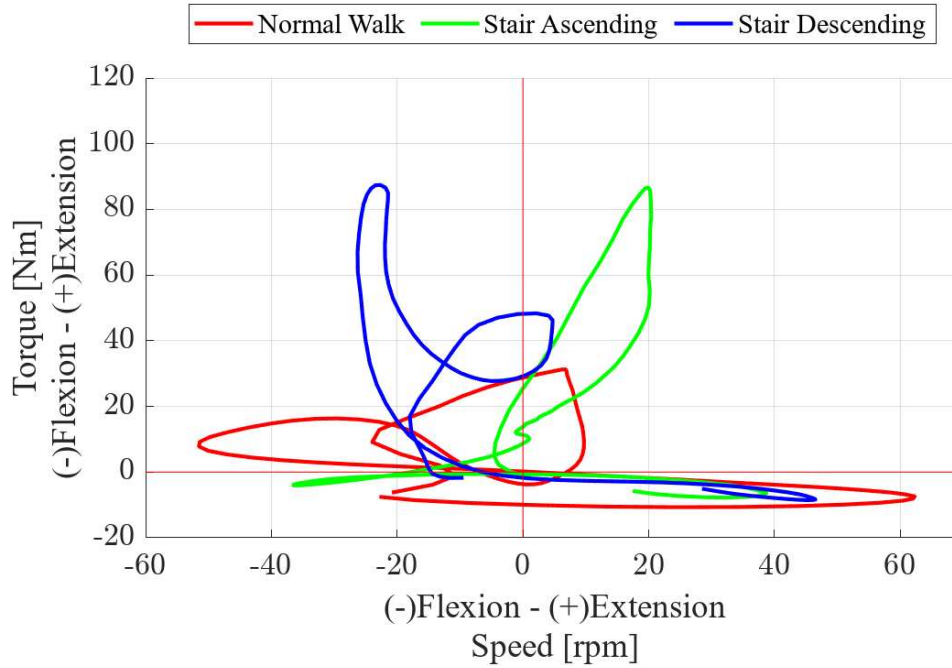


Fig. 2.1 Biomechanical torque-speed requirements for level walking (red) and stair ascent (green) and descent (blue), considering the 50th male percentile.

ascent necessitates substantial active power at the knee joint. While it is theoretically plausible to design an actuation unit capable of meeting these requirements, practical challenges arise. Specifically, designing for the required stance specifications would result in a heavy and bulky prosthetic device, compromising user embodiment. Alternatively, opting for a transmission system with a high power reduction ratio raises concerns about backdrivability, given that actuation reflected inertia is proportional to the square of the transmission ratio. Striking a balance between these considerations proves to be a substantial challenge in robotics. Essentially, during the stance phase of stairs ascent, the frictional constraint between the foot and the ground transforms the leg into a closed kinematic chain. Users can partially control movement through hip movement and handrail assistance [93].

In contrast, the swing phase heavily relies on the inertial coupling between the thigh and shank, primarily governed by the knee prosthesis. While this inertial contribution is generally sufficient for proper ballistic movement during regular walking, it tends to fall short when it comes to stair ascent. The hip motion fails to generate a knee trajectory that allows the foot to clear the subsequent step without resorting to compensatory behaviors like hip circumduction, pelvic tilt, or ankle

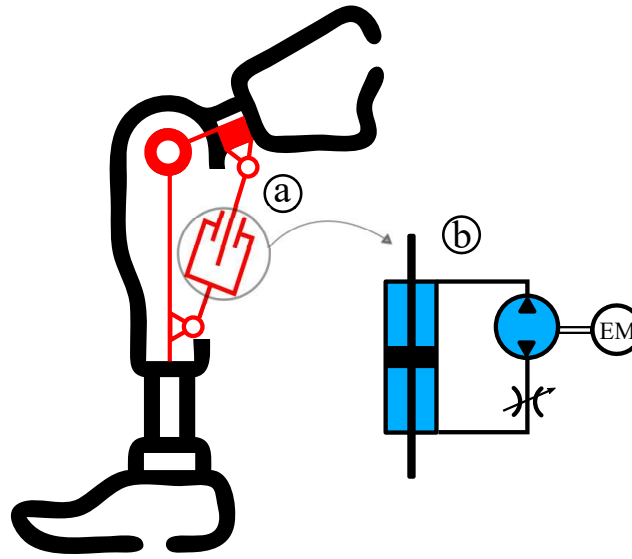


Fig. 2.2 Knee prosthesis architecture schematic: (a) inverted-slider crank mechanism to convert knee rotary motion into linear motion and vice versa, (b) mechatronic hydraulic unit comprising a linear hydraulic cylinder, a bidirectional gerotor pump, a control valve and an electric motor (EM).

vaulting. Consequently, the swing phase emerges as a deficient aspect in passive devices for stair ascent activities. Moreover, stance active knee extension torque proves most beneficial in a step-over-step gait, a movement seldom performed by transfemoral amputees [94].

In summary, the biomechanical rationale behind designing the actuation unit is to provide assistance during the swing phases of both stair ascent and level ground walking. Theoretically, this approach holds promise in substantially improving the gait of transfemoral amputees without impacting the size and weight of the prosthetic device. The mechatronic schematic of the proposed prosthesis is presented in Figure 2.2 and graphically summarizes the points discussed so far.

Knee kinematic behavior The subsequent biomechanical criterion to take into account pertains to the kinematic behavior of the knee joint. While 90° of knee flexion prove adequate for navigating both level walking and stairs ascent and descent, it is crucial to ensure that the knee's range of motion enables amputees to kneel and sit comfortably. Hence, the designated value for the knee range of motion is established at 110° . Alternatively, for a rapid walking scenario (i.e., $6 \text{ km} \cdot \text{h}$), the knee prosthesis is required to execute a walking gait, achieving a maximum angu-

lar speed of 84 rev/min (8.8 rad/s) during extension and -65 rev/min (6.8 rad/s) during flexion.

Knee kinetic behavior The final specification pertains to the prosthetic device's capability to generate and dissipate power at the knee joint. Power generation requirements can be determined by modeling the leg as a double pendulum with rigid links between the joints, as presented in Figure 1.5. The torque equilibrium at the knee joint is given by

$$\tau_h \approx \pm \tau_g + \tau_i + \tau_p \quad (2.1)$$

being τ_h the hip torque contribution at the knee, τ_g the gravitational torque (opposite to hip torque in swing flexion), τ_i the inertial torque and τ_p the prosthesis passive dynamics torque contribution. Since MPKs are unable to exert active torque at the knee joint, the hip has to counteract both inertial and gravitational terms, initiating an inertially-coupled, ballistic motion. To reduce the hip required effort, the actuation design rationale is to compensate for the inertial contribution. This term can be represented as

$$\tau_i = (J_{dev} + J_{act}N^2) \cdot \alpha_{k,max} \quad (2.2)$$

being J_{dev} the prosthesis rotational inertia, J_{act} the actuation rotational inertia, N the overall transmission ratio and $\alpha_{k,max}$ the maximum angular acceleration of the knee joint during the swing phase. The center of mass of the biological shank, situated at approximately 45 % of the total length from the proximal end [95], has been taken into account to calculate the leg's rotational inertia. Its length corresponds to approximately 0.5 m, considering the 50th percentile for men. The second term in the inertia equation (Eq. 2.2) represents the reflected inertia of the actuator, a value that significantly influences the backdrivability properties of the prosthesis. Previous studies have indicated that achieving desirable swing-phase dynamics is possible with a reflected inertia of up to 10 % of the overall rotational inertia, based on data from the biological shank and foot [61, 96]. Considering a fast walking condition, corresponding to an angular acceleration $\alpha_{k,max}$ of 120 rad/s² [97], the

inertial torque estimate can be rearranged as

$$\tau_i = 1.1 \cdot (m_{target} \cdot (0.45L_{k-f})^2) \cdot \alpha_{k,max} \quad (2.3)$$

The only variable in Equation 2.3 that remains unknown is the targeted mass of the prosthesis. Examining the well-established microprocessor-controlled knee prostheses available in the market, such as the Ottobock C-Leg and Ottobock Genium X3, with weights of 1.5 kg and 1.8 kg, respectively, provides a basis for estimation. By averaging these values and incorporating a 30% mass increase due to the actuation unit, the mass target for the knee prosthesis is set at 2.15 kg, excluding the ankle and foot components. In totality, the targeted leg mass, including a conventional passive ankle and foot, is approximately 3.1 kg, notably lighter than a biological limb's weight [26]. The estimate of Equation 2.3 results in a torque of 20 N · m at the knee joint, yielding an active power of up to 110 W, considering an angular speed of up to 7 rad/s at the knee joint. This actuation target aims to provide assistance during various activities, with a projected assistance level of 61 % during level walking stance, 22 % during stair ascending, and 19 % in sit-to-stand maneuvers relative to the torque requirements of healthy subjects [32], as graphically presented in Figure 2.1. Recent studies have suggested that partial assistance during these tasks can be highly beneficial for transfemoral amputees [98].

In contrast, the goal for power dissipation is to align with the characteristics of existing commercial microprocessor-controlled knee prostheses. However, obtaining specific information on the torque, power, and damping values required for such robotic devices has proven challenging. Therefore, insights from studies involving healthy subjects were evaluated to extrapolate the design requirements for the modulated hydraulic damper. The highest resistive behavior, associated with knee yielding during stair descent, necessitates a maximum resistive torque of 135 N · m, corresponding to a resistive power of 400 W and a damping constant of approximately 100 N · m · s/rad, considering a target user weight of 100 kg [33]. However, the knee should also facilitate the required passive ballistic motion during swing-related activities, requiring a minimum damping of around 1 N · m · s/rad [91]. Ultimately, the prosthesis should exhibit the capability to promptly control damping within this range and switch between the two extremes in no more than 100 ms, an average duration between heel strike and foot flat [25]. To summarize, Table 2.2 presents the relevant biomechanical requirements for the mechatronic design.

Table 2.2 Biomechanical requirements

Variable	Symbol	Values
Maximum swing torque	$T_{act,max}$	20 N · m
Maximum knee speed	ω_{max}	7 rad/s
Maximum active power	P_{act}	140 W
Knee range of motion	$\Delta\theta_k$	110°
Prosthesis mass target	m_{target}	2.15 kg
Maximum resistive torque	$T_{res,max}$	100 N · m
Maximum damping	β_{max}	100 N · m · s/rad
Minimum damping	β_{min}	1 N · m · s/rad

2.2 Electro-hydrostatic Actuation

The mechatronic schematic of a linear electro-hydrostatic actuator (EHA) is illustrated in Figure 2.3. This system operates on a closed-circuit transmission setup, featuring an electric motor that propels a bidirectional gerotor hydraulic machine. This hydraulic machine plays a crucial role in regulating pressure differentials and fluid flow within a hydraulic actuator. The design of EHAs combines the advantageous features of both hydraulic and mechanical actuation. Similar to their conventional hydraulic counterparts, EHAs deliver a commendable power-to-mass ratio, ensuring efficient power delivery. Furthermore, they maintain a quasi-rigid coupling between the actuation chain and the load, facilitating large bandwidth dynamics and precise motion control [99]. In scenarios where factors like maintainability and reliability take precedence, EHAs emerge as a robust alternative to electro-mechanical technologies. This is primarily because EHAs offer the dual benefits of decreased weight and size, as supported by research findings [100]. The integration of EHAs in systems not only enhances performance but also addresses key engineering considerations. The closed-circuit configuration, driven by an electric motor and hydraulic machine, exemplifies a harmonious synergy of components. This design allows EHAs to navigate diverse engineering challenges, making them a compelling choice in applications where both reliability and compactness are pivotal factors. The concept behind EHA systems is hydrostatic transmission, in which the medium for converting motion between rotary and linear domains is the fluid. Therefore, achieving an advantageous placement within the prosthesis layout is

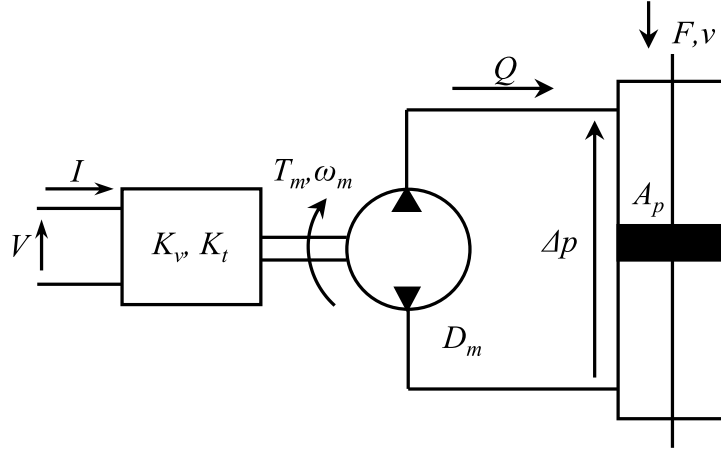


Fig. 2.3 EHA system schematic in actuation mode

feasible. The rotor shaft of the electric machine is quasi-rigidly linked to the fixed displacement hydraulic pump. Here, the rotational speed undergoes conversion into fluid displacement, resulting in a specific volumetric flow rate Q .

Conversely, the electric motor torque translates into a pressure differential Δp across the cylinder chambers. This pressure differential is proportional to the pump volumetric displacement D_m . Finally, the cylinder's cross-sectional area A_p is instrumental in converting both the volumetric flow rate Q into linear cylinder rod speed v (exhibiting inverse proportionality) and the pressure differential Δp into force F (displaying direct proportionality). This interplay of variables within the electro-hydraulic actuator design ensures effective and versatile functioning within the prosthetic device. To summarize the qualitative relationships expressed, it is possible to write the following equations

$$\begin{aligned} \omega_m &= \frac{V}{K_v}, \quad T_m = K_t I, \\ Q &= D_m \omega_m, \quad \Delta p = \frac{T_m}{D_m}, \\ v &= Q A_p, \quad F = \Delta p A_p \end{aligned} \tag{2.4}$$

in which each component is considered in its ideal and static conditions. Overall, the transmission ratio N_{EHA} [rad/m] between rotary and linear domains is given by the following relationships:

$$N_{EHA} = \frac{\omega_m}{v} = \frac{A_p}{D_m}, \tag{2.5}$$

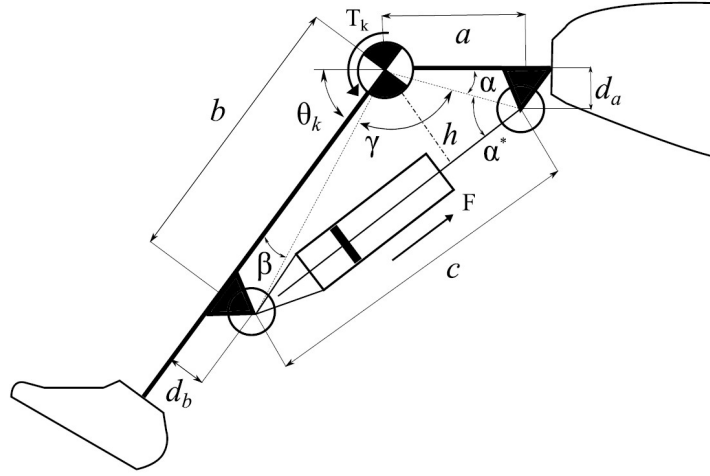


Fig. 2.4 Kinematics of actuation realizing an inverted slider-crank mechanism [101]

Additionally, another transmission contribution is given by the conversion of linear cylinder motion into knee joint rotation. Figure 2.4 presents a schematic of the actuation kinematics.

2.2.1 Hydraulic cylinder

The conversion of linear cylinder motion to rotary motion involves the utilization of the parameter h , which acts as a lever arm. This lever arm's behavior is essentially characterized by an inverted slider-crank mechanism, defined by parameters such as a , d_a , b , and d_b . The selection of these parameters requires a careful consideration of a trade-off between transmission ratio variability and adherence to geometrical constraints, ensuring the avoidance of a bulky prosthesis. Additionally, a hydraulic cylinder experiences both mechanical and hydraulic losses attributed to mechanical friction and hydraulic leakages. These losses are considered through an overall efficiency parameter η_c , which is the combination of mechanical η_{cm} and hydraulic η_{ch} losses and that, in a first approximation, can be regarded as constant. The methodology employed for designing the actuation unit adheres to a top-down approach, commencing with the biomechanical requirements outlined in Table 2.2. The optimization process involves leveraging a multi-objective, evolutionary genetic algorithm [83]. Figure 2.5 illustrates the workflow for the optimization framework to design the Electro-Hydrostatic Actuator (EHA). The modeling phase establishes analytical models for each EHA sub-component. Subsequently, the Multi-Objective

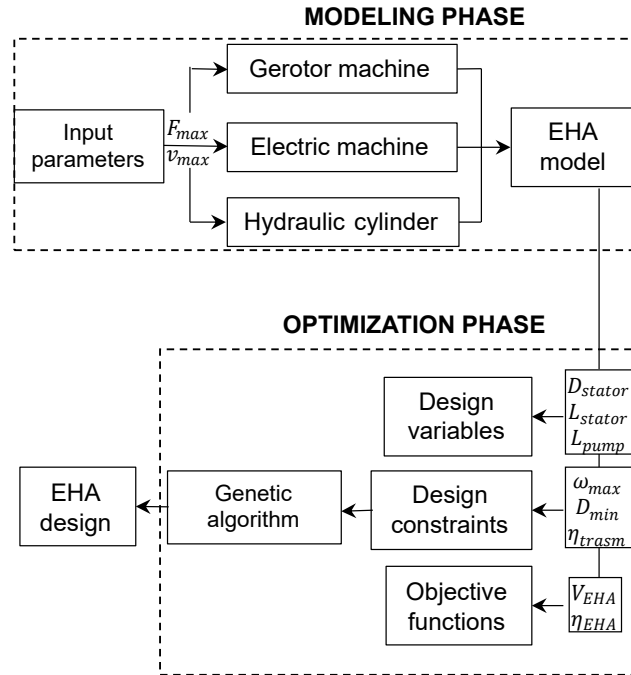


Fig. 2.5 Workflow of the EHA design optimization framework

Optimization (MOO) phase involves defining design variables, constraints, and objective functions crucial for the optimization algorithm. This systematic approach ensures the actuation unit is meticulously designed to meet biomechanical requisites while considering various design constraints and objectives.

First, the analytical model of each component is defined. To achieve actuator compactness, a passing-shaft, double-rod cylinder is chosen. This design eliminates the need for a hydraulic accumulator to counteract flow rate differences among the cylinder's chambers. The omission of an accumulator streamlines the device's hydraulic and mechanical complexity. Additionally, ensuring pressure preload necessitates either pressurized gas or a preloaded spring, introducing further intricacies to the hydraulic circuit. On the other side, adopting a passing-shaft architecture mandates a larger housing. This passing-shaft cylinder can be characterized by two fundamental parameters: the piston area A_p and the piston stroke s . The former is linked to the force exerted by the actuator and its load, while the latter determines the maximum travel of the actuator, delineating its operational range. Regardless of the application, sizing this component requires defining at least two operating points in the force-speed plane.

2.2.2 Gerotor hydraulic machine

Gerotor machines, integral to hydraulic systems, encompass internal gear units distinguished by inner and outer rotors that rotate eccentrically. Primarily applied in low-pressure scenarios, typically up to 10 MPa [102], gerotor machines are favored for their enhanced compactness and resilience to wear and noise compared to their external gear counterparts. The outer gear, featuring N teeth, exceeds the inner gear by one, resulting in a relative angular speed between them [103]. Functioning as positive displacement components, gerotors induce flow by orchestrating periodic suction and delivery actions within the N chambers formed during gear meshing [104]. While previous works have extensively explored optimizing the 2D section of gerotors for enhanced performance and reduced leakages [105, 106], a comprehensive analysis of their analytical design falls beyond the scope of this doctoral work. These prior studies, serving as benchmarks, contribute to the analytical modeling of the gerotor component. Defining the 2D geometry of the gerotor enables the computation of the specific volumetric displacement A_m of the component. This calculation, based on the emptying or filling functions of the device, reflects the average fluid area during a complete revolution of the outer gear. With such parameter, the volumetric displacement D_m can be obtained as follows

$$D_m = A_m(D_g)L_g \quad (2.6)$$

where D_g and L_g are the gerotor external diameter and axial length, respectively. Combining Eq.2.6 with the hydraulic cylinder requirements presented in Section 2.2.1, it is possible to derive the physical model of a gerotor machine. First, the maximum flow rate Q_{max} and pressure differential Δp_{max} on the gerotor side are computed starting from the system requirements and the hydraulic cylinder characteristics. Namely,

$$Q_{max} = \frac{A_p v_{max}}{\eta_{ch}}, \quad \Delta p_{max} = \frac{F_{max}}{A_p \eta_{cm}} \quad (2.7)$$

in which v_{max} and F_{max} represent the cylinder speed and force requirements, respectively; whereas, η_{ch} and η_{cm} are the hydraulic and mechanical efficiencies of the cylinder. Then, the theoretical, gerotor maximum torque $T_{g,max}$ and power $P_{g,max}$ can be computed as follows

$$T_{max} = D_m(D_g, L_g) \Delta p_{max},$$

$$P_{gmax} = T_{max} \omega_g = \frac{P_{max}}{\eta_c} \quad (2.8)$$

where ω_g is the inner gear angular speed and D_m is the machine volumetric displacement, proportional to gerotor external diameter D_g and axial length L_g . The second row of Eq.2.8 represents the gerotor power in nominal conditions, neglecting losses. To account for them, Harrison et al. proposed a 1D orifice model to analytically quantify viscous and friction losses in gerotor devices [107]. Within the presented work, losses due to axial and radial clearances between gerotor and its housing have been considered. The power loss due to axial clearances can be modeled according to

$$P_{f1} = 2\mu \frac{A_g \omega_{avg}^2 R_g^2}{c_{ax}} \quad (2.9)$$

in which μ is the fluid dynamic viscosity, A_g is the surface area of the film between gears and gerotor housing, ω_{avg} is the average angular speed between gears, R_g is the mean radius of friction surface and c_{ax} is the axial clearance. Then, the power loss due to radial leakages is accounted by considering the interaction between gear and roller bearing. Its effect can be modelled as follows

$$P_{f2} = 2\pi\mu L_g \frac{\omega_{avg}^2 R_{ogr}^3}{c_{rad}} \quad (2.10)$$

being R_{ogr} the outer gear radius with respect to the inner gear rotation center and c_{rad} the radial clearance value. It is also possible to model the power loss related to the radial clearance between gerotor envelope and bearing outer ring. The formulation is the same as Eq. 2.10, changing axial thickness and radius accordingly. Combining Eq. 2.8 with the power loss contributions, the gerotor overall efficiency η_g can be quantified as

$$\eta_g = 1 - \frac{P_{f1} + P_{f2}}{P_{gmax}} = f(D_g, L_g) \quad (2.11)$$

2.2.3 Brushless, permanent magnet electric machine

Brushless DC electric machines, particularly permanent magnet synchronous machine (PMSM), are prevalent in Electro-Hydrostatic Actuator (EHA) systems. The fundamental operational principle hinges on the interaction between a current-carrying coil and a magnetic field, giving rise to an electromagnetic force. Due to the intricate coupling of magnetic, electric, and mechanical domains, analytical modeling of these machines becomes inherently complex.

Guidelines for analytically modeling PMSMs have been delineated by Hanselman [108], providing valuable insights. However, when the 2D section of the machine assumes a level of complexity beyond certain thresholds, finite element-based models emerge as the preferred design paradigm. In the scope of this research, the electrical machine underwent an initial modeling phase based on the characteristic equations outlined in [108]. Subsequently, a more refined understanding of its performance was attained through a parametric Finite Element Analysis (FEA). A pivotal parameter in this analysis is the machine's external diameter, denoted as D_s , which assumes paramount importance for normalizing PMSM performances relative to a standardized size. The chosen topology adheres to a radial flux electric machine configuration, characterized by a slotted stator and surface-mounted permanent magnets. The main constraints from magnetic and electric domains are expressed in Table 2.1. The FEA outcome is the peak torque per unit meter T_{pm}

$$T_{pm} = \frac{k_t I_{ph}}{L_m} \quad (2.12)$$

in which I_{ph} is the phase current and L_m is the machine axial length, normalized to 1 meter. Considering the electric machine topology, the torque constant, k_t , is related

Table 2.3 Permanent Magnet electric machine main parameters

Symbol	Value	Description
N_{ph}	3	Number of phases
B_r	1.4	Magnetic remanence [T]
B_{max}	1.5	Maximum magnetic flux density [T]
J_p	20	Peak current density [A/mm ²]
J_c	6	Continuous current density [A/mm ²]

to the electric constant k_e by the following relationship

$$k_t = \frac{N_{ph}}{2} k_e = \frac{N_{ph}}{2} n_s N_m L_m B_g R_o \quad (2.13)$$

where n_s is the number of turns per stator slot, B_g is the air gap flux density, R_o the rotor outer radius and N_m the number of magnetic pairs. Overall, combining Eq.2.12, the parameters of Table 2.1 and the geometrical and electrical relationships provided in [108], it is possible to model the 2D section of the PMSM. Two potential approaches can be employed to integrate the electric machine's performance into the Multi-Objective Optimization (MOO) problem. One avenue involves invoking the numerical model of the electric machine for each set of design parameters, executing Finite Element Analysis (FEA), and subsequently extrapolating the torque per meter. While this approach would yield more precise outcomes, the computational time is heavily influenced by the numerical simulations. On the contrary, an alternative strategy involves conducting this analysis in advance, encompassing a diverse range of machine diameters (D_s , ranging between 20 and 150 mm). This allows for the direct inclusion of a torque-per-meter fitting function in the MOO problem. Consequently, the computational time is substantially reduced compared to the former case. Figure 2.6 illustrates the FEA torque per meter concerning the electric machine diameter and the corresponding fitted curve. Notably, the fitting function adequately represents the

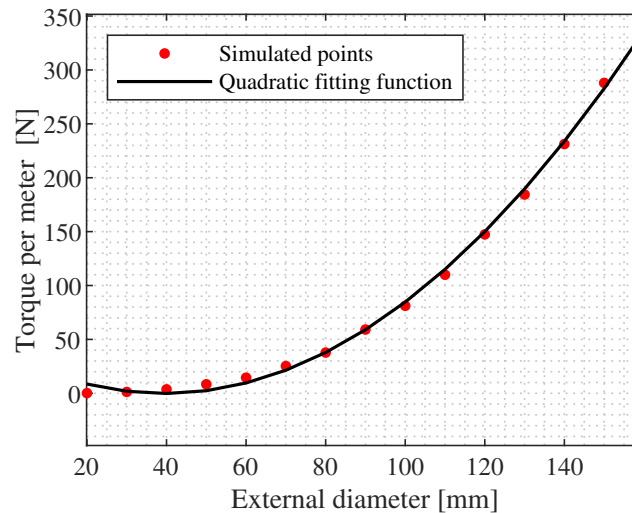


Fig. 2.6 Numerical characterization of the torque per meter T_{pm} as function of the machine external diameter D_s

numerical results, making it suitable for integration into the optimization problem formulation. Lastly, the inclusion of electric machine efficiency η_m is recommended, and it can be reasonably considered constant in the initial approximation, given that the MOO focuses on the EHA's nominal operating conditions.

2.2.4 Multi-objective optimization problem formulation

An MOO problem is formulated considering the proposed design model and variables, objective functions and constraint conditions [109]. Mathematically, it can be written as

$$\begin{aligned}
 & \min \{f_1(\mathbf{X}), f_2(\mathbf{X}), \dots, f_m(\mathbf{X})\} \\
 & \text{s.t. } \{g_i(\mathbf{X}) \leq 0, i = 0, 1, \dots, p\}, \\
 & \quad \{h_j(\mathbf{X}) = 0, j = 0, 1, \dots, q\} \\
 & \quad \mathbf{X}_L \leq \mathbf{X} \leq \mathbf{X}_U, \mathbf{X} = [x_1, x_2, \dots, x_n]^\top
 \end{aligned} \tag{2.14}$$

being $f_m(\mathbf{X})$ the m_{th} objective function, $g_i(\mathbf{X})$ the i_{th} inequality constraint, $h_j(\mathbf{X})$ the j_{th} equality constraint and $\mathbf{X}$ the vector of design variables. Additionally, $\mathbf{X}_L$ and $\mathbf{X}_U$ represent the vectors of lower and upper bounds, respectively. Such constraints define the variables space, $\mathbf{R}^n$, in which the algorithm can span to search for optimal solutions. Various strategies exist for solving multi-objective optimization (MOO) problems, each serving distinct purposes [110]. Broadly speaking, there are three primary methods. Firstly, one can combine the objective functions into a composite function using a weighted sum method, commonly known as *scalarizing*. This approach results in a unique optimal solution, and by adjusting the weights of individual objective functions, the Pareto front can be systematically generated. However, a drawback of this method is the lack of a definitive criterion for assigning weights, introducing additional uncertainty into the results.

The second approach involves treating all but one objective function as constraints, effectively transforming the problem into a mono-objective one. By iteratively applying this procedure with different combinations of objectives, the Pareto front can be established.

The third approach considers all objective functions simultaneously, aiming to identify the complete set of solutions in one iteration. This method is often employed in heuristics and evolutionary strategies, such as genetic algorithms (GAs). The methodology described in [111] leverages this criterion.

Selecting appropriate design variables is a non-trivial task, as they should be independent and minimized without restricting design possibilities. In the presented optimization framework, the design variables include the permanent magnet synchronous machine (PMSM) diameter D_s , its length L_m , and the axial length of the gerotor machine L_g . Additionally, relevant parameters like the gerotor diameter D_g and the hydraulic cylinder diameter D_p are not explicitly included but derived within the optimization process. To achieve compact yet high-performing electro-hydrostatic actuator (EHA) systems, the initial assumption is that the gerotor diameter matches the electric machine diameter D_s , including a bearing support D_{brg} . Similarly, the piston diameter of the cylinder D_p is correlated with the electric machine diameter. Exploiting data from a commercial bearing manufacturer [112] and considering an EHA application in assistive devices [60], it is assumed that

$$D_{brg} = D_s, \quad D_g = 0.65 D_s, \quad D_p = 0.3 D_s \quad (2.15)$$

The first objective function is the EHA volume, in favor of compactness and weight reduction. The consideration is confined to the volumes of the electric and hydraulic machines, as the shape and dimensions of the hydraulic cylinder are subject to significant variation. The objective function is expressed as:

$$V_m = \frac{\pi}{4} D_s^2 L_m, \quad V_g = \frac{\pi}{4} D_g^2 L_g - D_m, \quad (2.16)$$

$$f_1(\mathbf{X}) = V_{EHA} = V_m + V_g = f(D_s, L_g, L_m)$$

where the electric machine volume V_m is calculated as if it was full of material, whereas the fluid volume D_m is subtracted to the gerotor volume. The second objective function concerns the EHA efficiency. Namely, combining gerotor and electric machine efficiencies in nominal conditions, the formulation is as follows

$$f_2(\mathbf{X}) = -\eta_g \eta_m = f(D_s, L_g, L_m) \quad (2.17)$$

in which the minus sign is inserted so that the algorithm can perform an appropriate minimization. Finally, it is essential to establish design constraints to delimit the algorithm's search space. These constraints encompass both geometrical and physical aspects of the system and can be classified into two categories: boundary and performance constraints. Boundary constraints confine the design variables to specific domains and are articulated as upper and lower bounds. Conversely, performance constraints are expressed as either equalities or inequalities, outlining the physical limitations in the design. First, the maximum angular speed of the electric machine is taken into account. Accounting for the cylinder specifications of Section 2.2.1 and the definition of transmission ratio N , the constraint is formulated as

$$\tau = \frac{D_m}{A_p} = f(D_s, L_g), \quad (2.18)$$

$$g_1(\mathbf{X}) = \frac{v_{max}}{\tau} - \omega_{max} \leq 0$$

in which the first term is derived from the cylinder linear speed requirement; whereas, the second is its upper bound. Eq.2.18 takes into account that the angular speed of gerotor and electric machines is the same since they are quasi rigidly coupled. The second constraint is related to the gerotor volumetric displacement D_m . Recalling Eq.2.8 and the maximum power at the gerotor side P_{gmax}

$$\begin{aligned} P_{gmax} &= \frac{P_{max}}{\eta_c} = D_t \Delta p_{max} \omega_n, \\ D_t &= \frac{P_{max}}{\Delta p_{max} \omega_n} \\ g_2(\mathbf{X}) &= D_t - D_m \leq 0 \end{aligned} \quad (2.19)$$

in which ω_n represents the base speed, corresponding to the maximum torque condition for the electric machine and D_t is the target volumetric displacement. Eq.2.19

Table 2.4 Power generation requirements

Description	Symbol	Value
Maximum torque	T_{max}	20 N · m
Maximum angular speed	ω_{max}	7 rad/s
Maximum power	P_{max}	110 W

establishes a lower limit, ensuring that the power requirement is met at the very least. The subsequent step involves determining the algorithm used for resolution. Genetic algorithms (GAs) stand out as parallel, efficient, and robust search and optimization techniques inspired by the principles of genetics and evolution in nature. They demonstrate particular efficacy when dealing with complex, large, and multimodal search spaces. While traditionally designed as single-objective optimization algorithms, real-world problems seldom conform to this condition, given the need to satisfy diverse criteria for each problem. Consequently, GAs yield a set of solutions, representing a trade-off among objectives. Each solution vector $\mathbf{X}$ is encoded into a chromosome, a string-like data structure. Evaluation is performed using fitness values, indicating the quality of the encoded solution. Key operators in GAs include mutation, introducing random diversities to prevent local minima, and crossover, working between distinct chromosomes. In the crossover process, two parent chromosomes merge to create an offspring chromosome, aiming to iteratively combine those chromosomes that best align with the objective functions, facilitating rapid and accurate algorithm convergence. A solution vector $\mathbf{X}_1$ attains the status of Pareto optimal if and only if there is no other vector $\mathbf{X}_2$ that could reduce one objective function without simultaneously increasing another. The algorithm concludes its search for solutions upon meeting a specified termination condition, often expressed as a threshold in the objective and constraint functions. This concludes the methodology employed in designing the actuation component of the prosthesis, with the objective of providing a set of optimal EHA designs.

The actuation requirements and initial assumptions are stated in Table 2.4, derived from the biomechanical requirements definition of Section 2.1. Concerning the genetic algorithm, the initial population size is set to 200 chromosomes, 80 % of them is subject to crossover, the remaining to mutation, according to [113]. The maximum algorithm iterations are 100, whereas the tolerance in objective and constraints functions has been selected as 10^{-4} and 10^{-3} , respectively. Within the

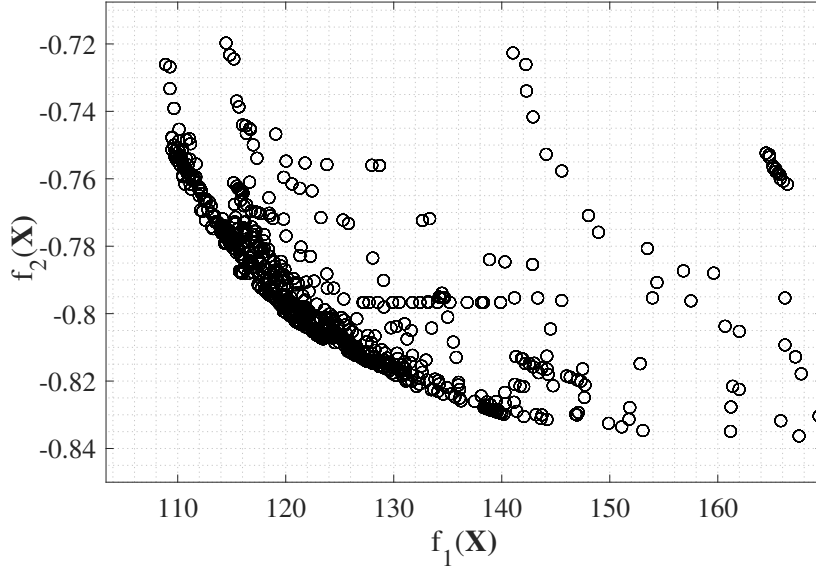


Fig. 2.7 Solutions space in the objective functions plane: Actuator volume $f_1(\mathbf{X})$ in cm^3 and global efficiency $f_2(\mathbf{X})$ in negative percentage.

optimization framework, the geometrical constraints are

$$\begin{aligned} 10 &\leq D_s \leq 50 \text{ [mm]} \\ 5 &\leq L_g \leq 20 \text{ [mm]} \\ 5 &\leq L_m \leq 40 \text{ [mm]} \end{aligned} \quad (2.20)$$

In Figure 2.7, the optimization results are depicted in the objective functions plane. $f_1(\mathbf{X})$ signifies the EHA volume, while $f_2(\mathbf{X})$ denotes the overall EHA efficiency with a negative sign. Given the contrasting objectives, where one is to be maximized and the other minimized, the algorithm strives to discover an optimal trade-off between the two, considering the imposed constraints. Figure 2.8 illustrates the Pareto front generated by the algorithm in relation to the design variables: electric machine diameter D_s , gerotor length L_g , and electric machine active length L_m . The blue projection of solutions reveals an inverse correlation between L_m and D_s combinations. This behavior can be rationalized by considering the quadratic relationship of the volume objective function with the diameter and its linear dependence on axial length. The algorithm seeks to minimize the diameter to reduce the overall EHA volume. Conversely, the efficiency objective promotes an increase in axial length at the expense of the electric machine diameter. Red dots

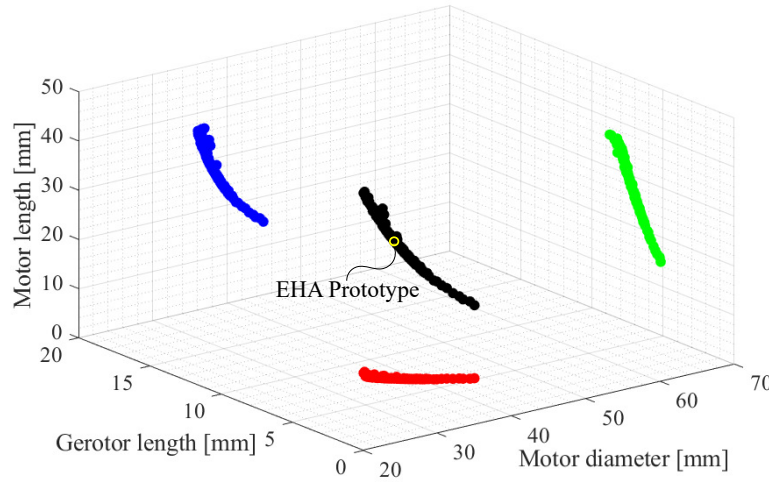


Fig. 2.8 Pareto front of the MOO problem. Black dots denote the solution space, whereas colored dots indicate projections over two of the three design variables. The optimal solution is highlighted in yellow.

signify the interplay between stator diameter D_s and gerotor length L_g , where the latter approaches a lower bound, imposing a constraint on the algorithm's search space. In addition, green dots illustrate the relationship between gerotor length L_g and electric machine active length L_m . Changes in L_m do not substantially impact gerotor length L_g , as it approaches its geometrical lower limit. Notably, there is a consistent direct proportionality between L_m and L_g . Furthermore, Figure 2.8 highlights the optimal solution's location with a yellow dot, which was chosen to design the EHA prototype. Table 2.5 provides details on the optimization parameters.

To validate the multi-objective optimization (MOO) framework numerically, a Computational Fluid Dynamics (CFD) analysis is conducted, focusing on the gerotor component. This involves simulations to examine fluid volume behavior in two distinct configurations: operating as a motor and as a pump.

In pumping operations, the electric machine imposes torque, and the gerotor pressure differential resists this load, corresponding to the II and IV quadrants of the torque-speed plane. Conversely, in scenarios where the gerotor pressure differential generates torque, and the electric machine resists the load (typical during backdriving of the EHA), the gerotor functions as a hydraulic motor, exploring the I and III quadrants. The CFD analysis is set up to consider positive torque, varying the gerotor angular speed from negative to positive values, given the symmetrical energetic

Table 2.5 EHA prototype design parameters

Description	Symbol	Value
Electric machine diameter	D_s	38 mm
Electric machine length	L_m	26 mm
Gerotor diameter	D_g	28.5 mm
Gerotor axial length	L_g	8 mm
Gerotor volumetric displacement	D_m	1.25 cc/rev
Linear cylinder piston diameter	D_p	22 mm
Transmission ratio	N_{EHA}	1.9 rad/mm

behavior of the gerotor with respect to force. Simerics Pumplinx, a specialized computational fluid dynamics software package for valves and hydraulic machinery, is employed for simulation. The rotating grid interface in Pumplinx facilitates fluid motion without engaging fluid-structure interaction, minimizing computational costs. Fluid domains are imported from an STL file generated in the source CAD software, incorporating trochoidal profiles. The volumes are meshed with approximately one million elements, and clearances are meshed with at least three-layer elements to simulate leakages. The simulation utilizes the laminar flow and cavitation modules in Pumplinx, considering fluid parameters such as density ($\rho = 840 \text{ kg/m}^3$) and dynamic viscosity ($\mu_f = 9.4 \text{ mPa/s}$). The simulation input grid comprises six equally spaced angular speed values (ω_g) between -10 krev/min and 10 krev/min and six pressure differentials (Δp) ranging from 0.5 to 5 MPa between ports. Each simulation involves solving the 3D fluid model for an angular motion transient equivalent to four hydraulic chamber cycles, with a maximum of 100 iterations per time step. Both radial and axial clearances are considered in the simulation. Radially, the clearance is set to $c_{rad} = 10 \mu\text{m}$ to account for viscous losses between the gerotor and its envelope. Axially, the clearance is set to $c_{ax} = 20 \mu\text{m}$ on each side, assuming a uniform gap distribution. The simulation does not consider axis misalignment or gear deformation effects, and mismatched grid interfaces are implemented to connect meshed volumes. In each simulated point, the efficiency is calculated according to

$$\eta_{gm} = \frac{T_g \omega_g}{\Delta p Q_g}, \quad \eta_{gp} = \frac{\Delta p Q_g}{T_g \omega_g} \quad (2.21)$$

in which η_{gm} is for gerotor motoring operations (i.e., I quadrant) whereas η_{gp} for pump behavior (i.e., II quadrant). The analysis outcomes are torque and flow rate, extrapolated considering a gridded interpolation. Efficiency results are post-processed

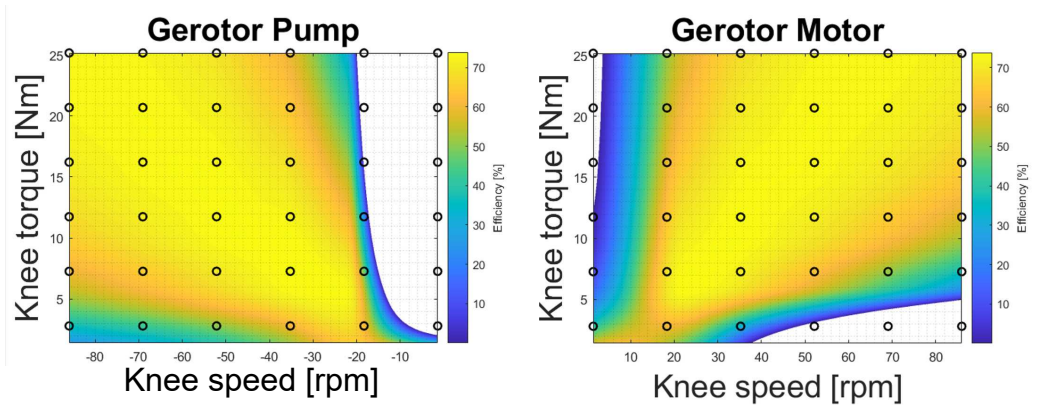


Fig. 2.9 Gerotor unit efficiency maps in two operating conditions: hydraulic pump (left), hydraulic motor (right). Black dots represent simulated points whereas the colormap is the result of a gridded interpolation.

with a piece-wise cubic interpolation function and presented in Fig. 2.9. Note that map values are reported in the knee torque-speed plane to gather an understanding at a prosthesis level. This conversion is a proportional scaling that employs the pump displacement, piston cross section and inverse slider-crank nonlinear transmission ratio. The maximum efficiency is approximately 75% in both operating conditions. Efficiency maps reveal distinct hydraulic (volumetric) and mechanical losses. Hydraulic losses manifest as a decline in efficiency at elevated torques, while mechanical losses become apparent at high speeds, as depicted by blank regions in Figure 2.9. Additionally, the sealing performance of the gerotor, when operating as a pump, surpasses that during motoring operations. The efficiency in both scenarios is influenced by clearances; tighter gaps favor volumetric efficiency over mechanical efficiency, while larger gaps reduce viscous drag but result in increased hydraulic leakages.

Similarly, Finite Element Method (FEM) models are utilized for a preliminary assessment of electric machine efficiency, with simulations conducted under motoring conditions. Comparable outcomes can be anticipated when the machine operates as a generator. Figure 2.10 illustrates the electric machine efficiency map, derived through a shape-preserving piece-wise cubic function. The peak efficiency reaches approximately 78%, aligning with the machine's nominal operating condition. However, the map indicates that the electric machine struggles to operate efficiently under specific conditions. Low-speed, high-torque, and high-speed, low-torque operations inherently exhibit inefficiencies, where the generated electric power falls short of

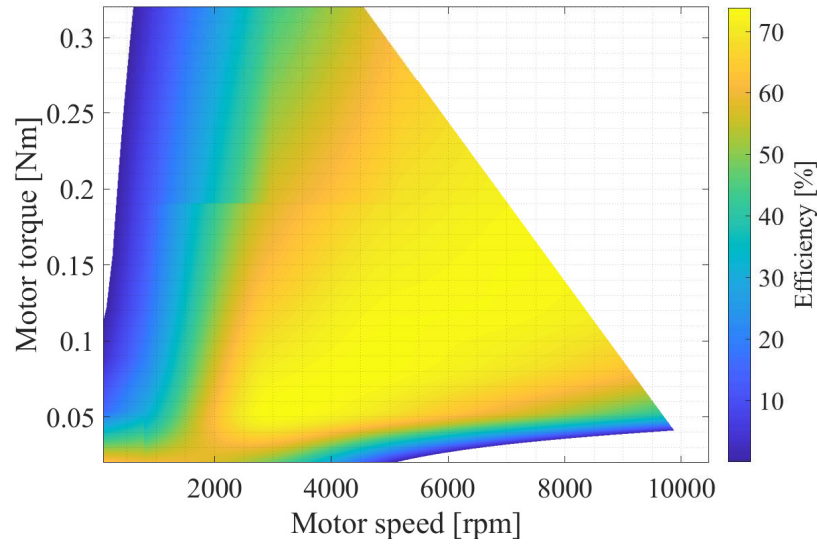


Fig. 2.10 Electric machine efficiency maps.

overcoming core and Joule losses. In order to circumvent the necessity for a fully customized machine while fulfilling the power generation prerequisites, the actuator is equipped with a 110 W frameless, brushless motor (TQ RoboDrive ILM38x06). This motor provides a maximum continuous torque of $100 \text{ mN} \cdot \text{m}$ and a peak torque of $320 \text{ mN} \cdot \text{m}$. It achieves a maximum efficiency of 88 % and a 10.5 krev/min angular speed, if operated at no-load.

2.3 Rotary, Flow Control, Hydraulic Valve

Focusing on the passive aspect of the prosthesis, the rotary hydraulic valve consists of three integral elements, as illustrated in Figure 2.11. A valve shaft, driven by a direct current (DC) electric motor, is equipped with two end stops, defining a 75° opening range. Additionally, the design incorporates a valve cap and an outer sleeve seamlessly integrated into the prosthesis casing. The functionality of the hydraulic valve is contingent upon the relative rotation of shaft and sleeve, governing the flow passage area and subsequently influencing the damping torque at the knee joint. The variable damping characteristic is linked to the change of the flow passage area. In a closed configuration, the valve is designed to withstand a significant 20 MPa pressure differential Δ_p with minimal leakage. This capability is crucial for executing

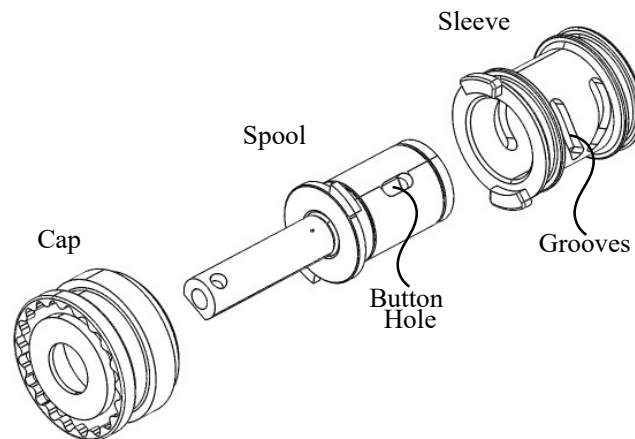


Fig. 2.11 Valve exploded view

tasks such as negotiating stairs or ramps descent and securing the prosthetic device in a locked position. Conversely, in a fully open state, the valve accommodates a maximum flowrate of 10 l/min, resulting in 0.15 MPa Δp , associated with the hydraulic resistance of the valve. This low pressure differential proves advantageous during the swing phase, where minimal damping is often desired. The valve behavior can be controlled, resulting in a continuous modulation of pressure differentials and flowrates within predetermined boundary conditions. Depending on the operating condition, the fluid path is different. Specifically, during knee flexion, hydraulic fluid enters axially through the valve spool and exits radially through the groove formed by the sleeve and valve housing. Conversely, during knee extension tasks, the fluid flow pattern is reversed. Considering these distinct fluid pathways, it is anticipated that the characteristics of the valve will exhibit slight variations between swing flexion and extension, with the latter likely displaying a higher impedance profile. Notably, the valve cap assumes the crucial role of axially aligning the valve spool and sleeve. Additionally, an axial thrust bearing, situated between the valve cap and spool, bears the axial load of the fluid on the sleeve.

The aim of the rotary hydraulic valve is to enable substantial braking torques, a characteristic unattainable or inefficiently achieved by the motor-gerotor unit alone. This valve facilitates the regulation of hydraulic system resistance or damping by governing valve closure through a compact electric actuator. Given that the valve predominantly influences the behavior of the prosthesis, it is relevant to define its design rationale. Specifically, the passive impedance behavior of the knee prosthesis is defined by the intersected area between the openings of the spool and sleeve.

Empirically, the objective is to establish a linear damping characteristic as the valve transitions from a fully open to a fully closed position, encompassing the maximum to minimum fluid passage area. This envisioned linearity is crucial to facilitate a broad spectrum of control over the valve, allowing continuous modulation of the damping behavior at the knee joint. To model the volume flow rate, an orifice plate model can be employed. This modeling approach contributes to a comprehensive understanding of the dynamic behavior of the hydraulic system within the prosthesis. The volumetric flow rate can be defined as

$$Q_v = \frac{C_d}{\sqrt{1-\beta^4}} \varepsilon A_o \sqrt{2\Delta_p \rho}, \quad (2.22)$$

in which C_d is the discharge coefficient, β is the ratio between orifice and pipe diameters, ε is the expansibility factor, Δ_p is the pressure differential, ρ the fluid density and A_o the intersected area. Rearranging the equation unifying constant terms, the volumetric flow rate is proportional to

$$Q_v \propto k A_o \sqrt{\Delta_p}, \quad (2.23)$$

Considering that the volumetric flow rate Q_v is the product of intersected area A_o and fluid velocity v , it is possible to write the hydraulic resistance as follows:

$$R_{hyd} = \frac{\Delta_p}{Q_v} \propto k \frac{A_o v}{A_o^2} = k \frac{v}{A_o}, \quad (2.24)$$

Assuming no velocity variations, the hydraulic resistance is proportional to the inverse of the orifice area. No turbulence and cavitation effects are considered at this initial design stage. The hydraulic resistance is largely determined by the valve's orifice cross-sectional area A_o , which corresponds to overlapping grooves or holes in the inner and outer walls of the valve, and which is directly related to the controlled valve rotation θ_v . For a small orifice, the valve dominates the losses that occur in the damper, so that this relation also applies to the damper force F and speed v , which are the reciprocates of pressure difference and flow rate:

$$F \propto \frac{v^2}{A^2} \quad (2.25)$$

The non-linearity of eq. (2.25) is unfortunate for controllability: phases governed by high braking torques and low speed require the valve to be operated in the

almost closed range, for which the damping behavior is extremely sensitive to valve rotation. Conversely, phases governed by low resistance and high speed require the valve to be fully open and fast. This undesired behavior is further exacerbated if A increases progressively as function of θ_v . This was found to be the case for an early design of the component that was available to test in the lab. In particular, it was experimentally found that roughly $F \propto v^2/\theta^3$, which posed controllability issues in an early prototype design.

In response, I have designed a custom valve with reduced non-linearity of damping force as a function of valve rotation. In particular, eq. (2.25) suggests that A should regressively change as function of valve rotation according to $A = \theta_v^{-1/2}$, in order to achieve the linear relation $F \propto v^2\theta$. This rationale manifests as a hyperbolic trend in the intersected area, characterized by the following pattern:

$$A_o = \frac{k_1}{\theta_v + k_2}, \quad (2.26)$$

being θ_v the relative position of the spool with respect to the sleeve. From a mathematical point of view, the intersected area is represented by the integral of the valve intersecting trajectory between spool and sleeve. Hence, the geometrical trajectory of the intersected area can be computed as follows

$$y(\theta_v) = \frac{dA_o}{d\theta_v} = \frac{k_3}{(\theta_v + k_4)^2}, \quad (2.27)$$

in which k_3 and k_4 should be tuned according to the application. In a qualitative analysis, Figure 2.12 illustrates a potential geometric configuration of the valve ports alongside a reconstructed 3D fluid model used for simulations. The geometric profile of the ports can be established based on the trajectory of the intersected area. Two viable options emerge: (i) configure both of them in a manner aligning with the theoretical behavior of the system, or (ii) maintain one port fixed as a straight line and solely define the shape of the other port, as depicted in Figure 2.12. In the presented design, the latter option is favored for its simplification of manufacturing processes. Consequently, the spool port geometry adopts the form of a *buttonhole*, where the surface defining the intersection remains a straight line. Conversely, the sleeve port geometry adheres to the theoretical analysis, as illustrated in Figure 2.11. However, incorporating a hyperbolic geometric relation into the mechanical design is both impractical and grounded solely on an approximation of hydraulic losses (i.e.,

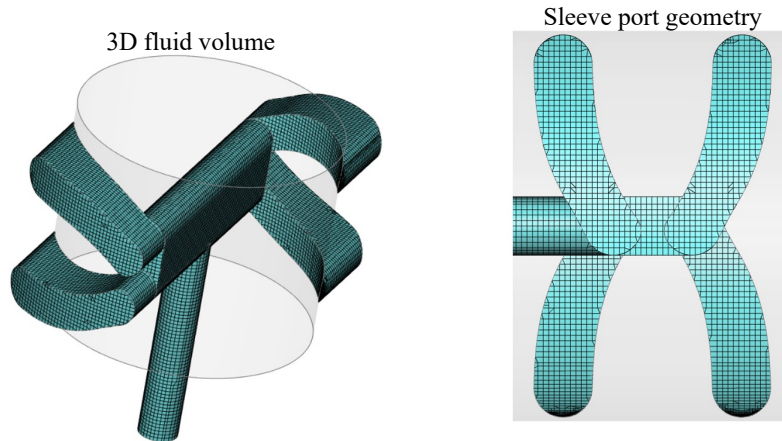


Fig. 2.12 Valve fluid volume in fully open condition

eq. (2.25)). Consequently, the decision is made to pursue a parametric optimization of a feasible design using computational fluid dynamic analyses. Specifically, the groove in the outer wall of the valve, which influences $A(\theta_v)$, is characterized as a machinable polynomial. The coefficients of this polynomial are then optimized to enhance the linearity of the pressure difference concerning the valve rotation θ_v for a given flow rate within a rotary range of $10 \leq \theta_v \leq 50$ %. For practical considerations, a cubic curve has been chosen. Figure 2.12 showcases the 3D model of the fluid volume in a fully open state.

To validate the theoretical analysis and assumptions associated with the port geometry, computational fluid dynamics (CFDs) analyses were conducted using the Simerics Pumplinx environment. Specifically, the fluid volume is extrapolated from the CAD model, incorporating a thin extruded cut to simulate the radial clearance between the spool and sleeve. This clearance is set to $10 \mu\text{m}$, following best practices in valve design. Opting for a tight clearance tends to enhance hydraulic performance over dynamic friction. In situations demanding secure valve locking under weight-bearing conditions, a tight clearance is often preferred to ensure proper resistance behavior. To replicate the impact of the interface between the spool and sleeve, a 3D annulus volume was generated, accounting for the entire radial interface between the two components, as emphasized in Figure 2.13. The rotating grid interface in Pumplinx facilitates fluid motion without necessitating fluid-structure interaction, a process associated with a high computational cost. The fluid volumes undergo meshing with a grid comprising approximately one million elements. Clearances are meshed with a minimum of three-layer elements to accurately replicate leakages. The

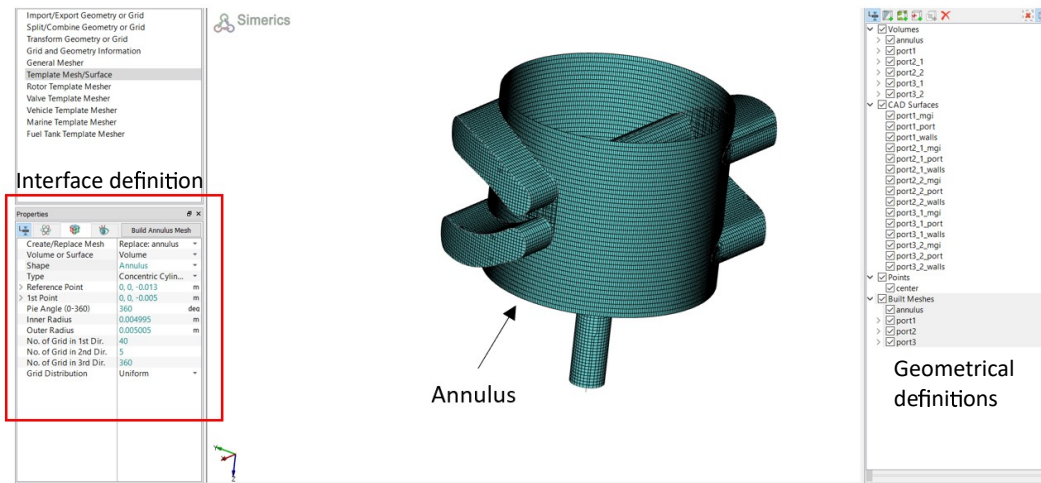


Fig. 2.13 Valve simulation environment

model is then simulated utilizing the turbulent flow and cavitation modules within Pumplinx, considering a fluid density of $\rho = 840 \text{ kg/m}^3$ and a dynamic viscosity of $\mu_f = 9.4 \text{ mPa}$ in the fluid domains. Then, the interface surfaces between the annulus and valve ports should be carefully selected. This procedure is paramount for simulating the fluid behavior at the component interface, a region that delineates the damping characteristics of the hydraulic component. Figure 2.14 provides an illustrative example of the interface definition. The simulation framework does not consider axis misalignment or mechanical deformations. The simulations were executed by imposing volumetric flow rates within the range of $-10 \div 10 \text{ L/min}$, concurrently measuring the pressure differentials between the valve input and out-

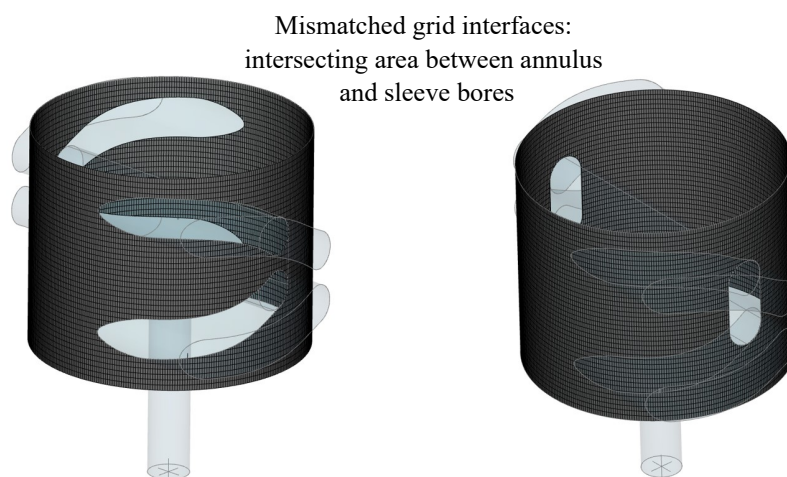


Fig. 2.14 Valve surface interfaces

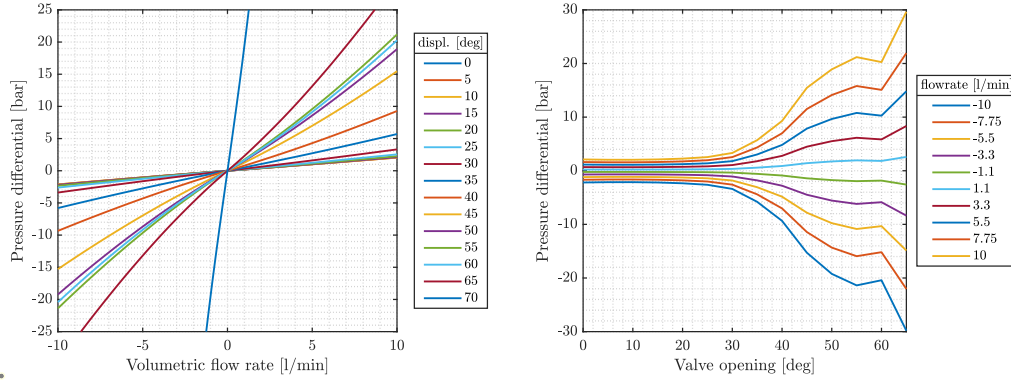


Fig. 2.15 Valve numerical characteristic. Left) Pressure differential as function of flowrate, for different valve apertures. Right) Pressure differential as function of valve opening, for different flowrates.

put. A preload pressure of 1 MPa was factored into the analysis, a value chosen in accordance with the Electro-Hydraulic Actuator (EHA) specifications. This choice aimed at preemptively averting cavitation concerns when actively applying power at the knee joint. It is worth to highlight that simulations enforcing pressure differentials and those gauging volumetric flow rates should ideally converge to identical results. Figure 2.15 presents the outcomes of the simulations in the pressure-flowrate plane, as function of valve opening angle θ_v on the left. Notably, for each valve opening, the valve characteristic aligns with a first-order polynomial. Overall, the trend hovers between linear and quadratic. This is likely due to the exclusion of velocity variations, and by neglecting nonlinearities, such as boundary effects and the turbulent regime. Despite these intricacies, the overall behavior of the valve emerges as highly controllable as it progressively opens, theoretically ensuring a wide spectrum of discrete damping behaviors at the knee joint. This adaptability hints at the potential for versatile control over the prosthesis's damping characteristics. The right side of Figure 2.15 presents the pressure differential trend as function of valve aperture, for different flowrates. The trend is between linear and quadratic, the latter predominant towards valve closure. If compared to valve available to test in the lab, the presented design shows a reduce nonlinearity, as presented in Figure 2.16 in terms of equivalent turbulent damping (F/v^2). Especially towards valve closure, the behavior is rather linear, improving the controllability of the device in this region. Finally, the valve spool is driven by a 9 V DC motor (Faulhaber 1016 SR) with a 64 : 1 gear head and embedded encoder for position control. The valve components are manufactured in stainless steel, surface tempered and lapped to

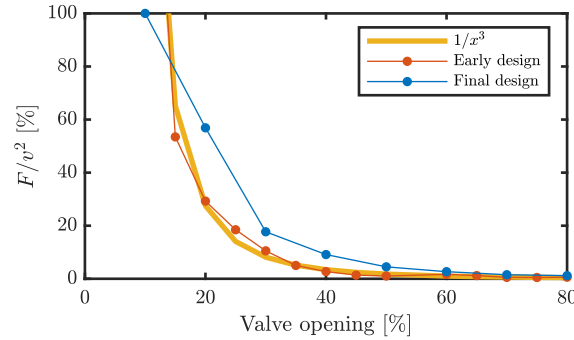


Fig. 2.16 A characterisation of an off-the-shelf valve design shows a strong nonlinearity ($1/x^3$) with valve position, which is undesirable for control purposes. The final design reduces this nonlinearity. In this graph, the equivalent quadratic damping coefficient F/v^2 is computed as a least-square fit.

guarantee proper fluid dynamic behavior as well as reduced dynamic friction. The radial clearance between spool and sleeve has been set to $10\text{ }\mu\text{m}$. Chapter 4 will deal with the experimental characterization of the component in isolation and its integration within the prosthesis.

2.4 Prototype Assembly

In Figure 2.17, a comprehensive model of the mechatronic system, encompassing the actuation unit and the controllable rotary valve, is presented. Cross-section b highlights the fluid path within the valve, indicated in red; whereas cross-section c illustrates the fluid path within the Electro-Hydraulic Actuator (EHA) in blue. The sequence of fluid entry serves as an indicator of the knee's mechanical action: initial fluid entry into the valve implies knee flexion, whereas a sequence of fluid entering the EHA first and then the valve corresponds to knee extension. It is worth to remark that the EHA and the valve operate in series. Moreover, the electric machine operates submerged in fluid, being its rotor rigidly coupled to the gerotor inner gear.

Figure 2.18 presents the 3D model of the leg prosthesis, not including embedded electronics and battery pack. The pressure differential either defined by active or passive side is sensed through two pressure transducers (Kulite ETL-76A-190 (M) series). The structure connects to the knee joint through a spherical coupling and attaches to the structural chassis via a revolute joint. These linkages create an inverted slider-crank mechanism. Importantly, the transmission ratio in this mechanism varies

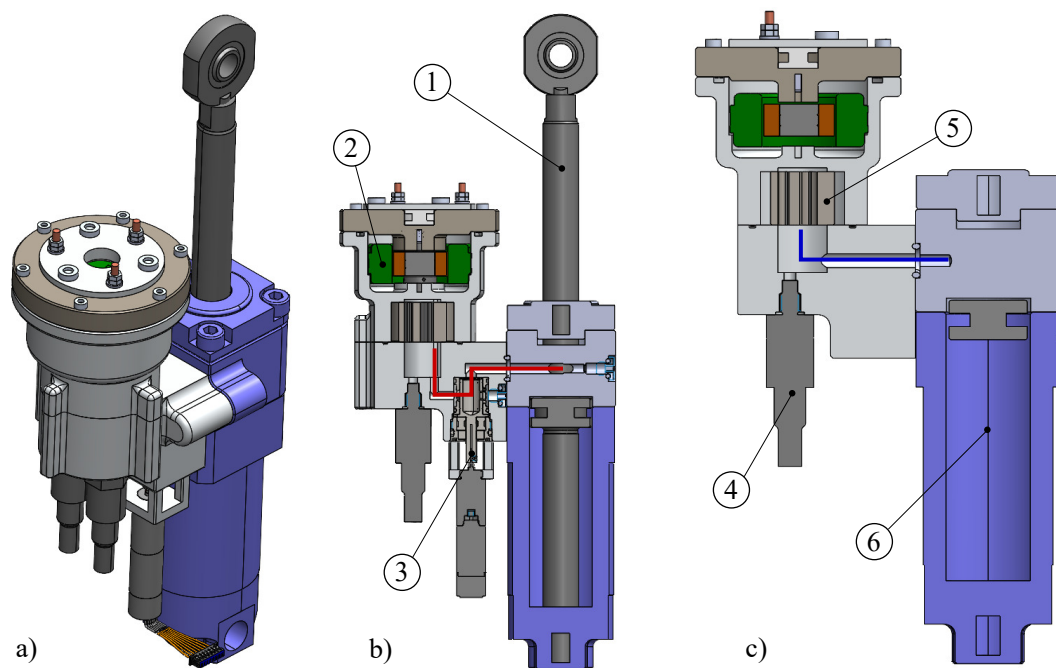


Fig. 2.17 Mechatronic system solid model. a) Overall assembly; b) Cross-section highlighting fluid flexion path (red); c) Cross-section highlighting fluid extension path (blue). (1) Cylinder rod; (2) Electric machine; (3) Motor-controlled valve; (4) Pressure sensor; (5) Gerotor machine; (6) Hydraulic cylinder.

with the knee angle, a characteristic illustrated in Figure 2.19. In terms of the prosthesis itself, it includes a passive foot/ankle unit connected to a pylon. Inside this pylon, there is a load cell embedded to measure both axial force and torque. The latter is integral to the structural chassis by means of a clamp mechanism. Finally, the knee angle is sensed by means of an off-axis, rotary, absolute, magnetic encoder (RLS AksIM-2) and a steel rod serves as end-stop, defining the minimum flexion angle of the prosthesis. Furthermore, the electronic components include an inertial measurement unit and a current sensor, incorporated for control purposes that will be elaborated upon later. Given the emphasis on the prosthesis assembly in this section, it is valuable to direct attention to the prosthesis chassis, considering its structural, acoustical, and ergonomic aspects.

2.4.1 Chassis Design

The prosthesis chassis assumes a pivotal role in the functionality of the device. It is imperative that the chassis effectively bears the load during the stance phase,

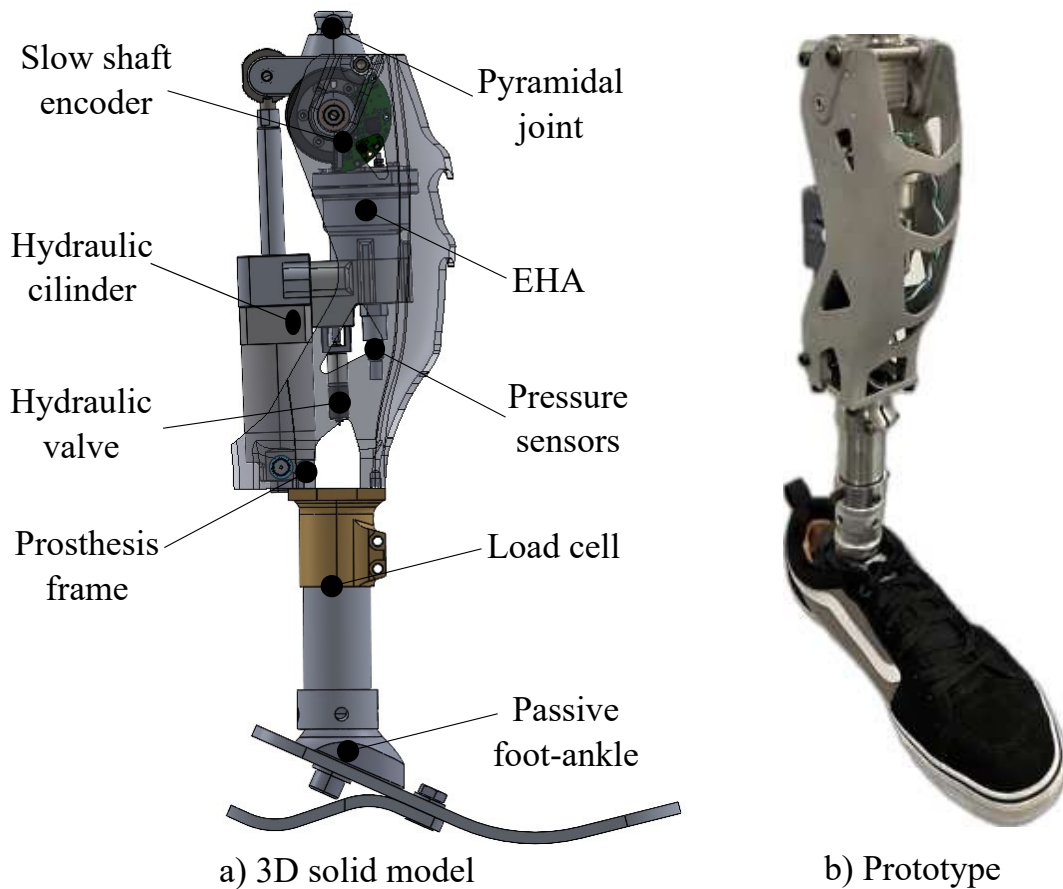


Fig. 2.18 Knee prosthesis prototype with structural chassis

emphasizing its structural significance. Furthermore, the chassis is designed to be ergonomically sound, ensuring user comfort, and concurrently exhibits aesthetic appeal. Additionally, in the realm of (semi)-powered knee prostheses, the chassis endeavors to mitigate acoustic noise emissions from the actuation unit, underscoring its acoustical considerations.

This section consider three aspects of the prosthesis frame. Namely structural integrity, ergonomic design, and acoustical considerations within the prosthesis chassis. It is noteworthy, however, that while these aspects contribute significantly to the overall prototype, they do not constitute the primary focus of this research endeavor. For a more exhaustive elucidation of these themes, interested readers are directed to refer to the comprehensive explanations provided in the references [114, 115].

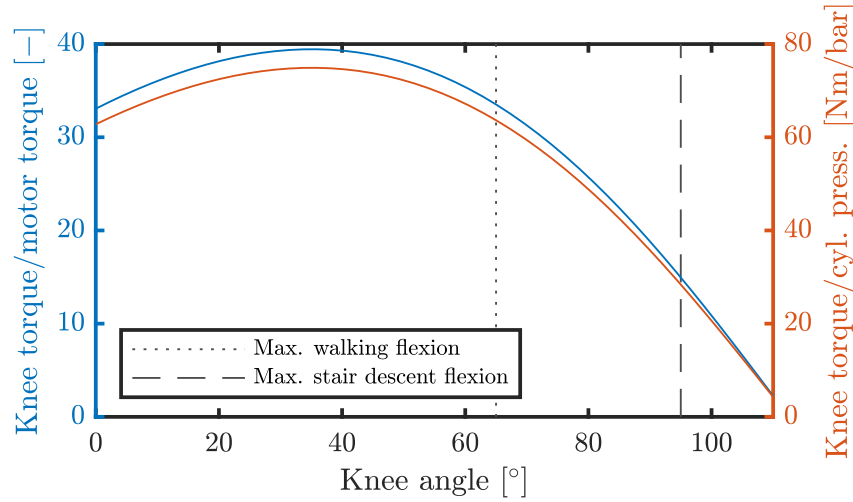


Fig. 2.19 Overall prosthesis transmission ratio between knee joint rotation and electric motor rotation (blue) and knee joint rotation and hydraulic damper (orange).

Structural argument Guaranteeing appropriate structural properties of a prosthesis chassis is a non trivial problem. On one facet, the prosthesis chassis necessitates structural robustness to cyclically endure the applied user load, while concurrently adhering to the constraints of being lightweight and conforming to the dimensions of a natural limb on the other. This duality in objectives renders the design of this component a compelling optimization problem. Within the scope of this doctoral investigation, the chassis undergoes topological optimization to ensure commendable loading capabilities, weight reduction, and dimensional compactness.

The topological optimization process, executed in Ansys, initiates with a voluminous chassis to facilitate optimization. The loading conditions are defined in accordance with ISO 10328 [116], considering the weight of a standard user at 100 kg, corresponding to the loading condition P5 (i.e., 3980 N at ISO interface and respective transportation moments, considering toe loading). Considering a density-based optimization algorithm, the Solid Isotropic Material with Penalization (SIMP) method [117, 118] is employed. The optimization objective function pertains to the minimization of the mechanical compliance of the chassis, thereof maximizing stiffness. Performance and physical constraints delimit the algorithm's search space, permitting a maximum volume reduction of 60 % to facilitate integration of the actuation system, embedded electronics, and overall mass reduction.

Subsequently, Von Mises stress constraints are enforced, wherein the yield strength limit is applied to critical interaction regions. Conversely, a safety factor

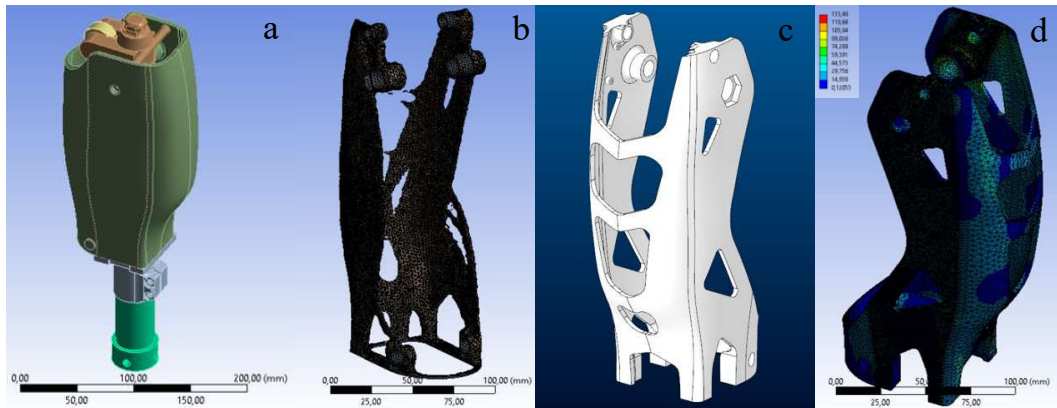


Fig. 2.20 Chassis structural optimization. a) Initial chassis. b) Optimization chassis. c) Optimized chassis. d) FEA numerical validation.

of two on the yield strength governs the remaining frame. The algorithm yields a preliminary optimized geometry, necessitating refinement for manufacturability assurance. The optimized chassis is then incorporated into the complete assembly, subjected to finite element analysis under the same loading conditions to numerically validate the proposed topological optimization. Figure 2.20 delineates the four stages leading to the final optimized chassis. The physical realization involves metal 3D printing in AlSi10Mg aluminum alloy, resulting in a noteworthy 47 % reduction in weight (reducing from 542 g to 285 g). It is imperative to acknowledge certain limitations: first, a 3D printed chassis may exhibit suboptimal mechanical fatigue characteristics, attributed to irregularities and voids inherent in the manufacturing process. However, compliance with the fatigue specifications of ISO 10328 show adequate life up to the P3 loading condition, roughly corresponding to a 50th male user. Second, the consideration of manufacturing cost, notably the expense associated with metal 3D printing remains a pertinent factor impacting the overall cost of the prosthetic device. However, a chassis with a shell-type design remains costly, even when using traditional manufacturing methods. 3D printing could offer a viable solution in this context, as it eliminates material wastage during the manufacturing process. Finally, Figure 2.21 shows the developed chassis.

Acoustic argument In adherence to the prosthesis design rationale, which incorporates a motor for assistance as needed, the issue of acoustic noise becomes a pertinent consideration. The introduction of active power at the knee joint often entails a device that is not only heavier and bulkier but also exhibits increased noise



Fig. 2.21 Metal 3D printed, prosthesis structural chassis

levels [119], thereby impeding an optimal usability of the prosthesis. Despite its significance as a user need [34, 36, 120], acoustic noise remains largely overlooked in the literature, with only a few works addressing it, either through characterizations [60] or by integrating noise reduction considerations in the design and selection of the actuator and its gearing [121]. These components represent the primary sources of noise in (semi-)powered prostheses.

Recognizing the imperative to tackle this issue, the objective of this section is to present an initial endeavor aimed at mitigating prosthesis acoustic noise through the design of a purpose-built chassis with a focus on noise reduction. By concentrating solely on the casing, this approach can be extrapolated to address audible noise concerns in any prosthetic design. The initial step involves conducting an acoustic characterization of both the prosthesis and the actuation mechanism to ascertain whether the chassis functions as a sound box. Figure 2.22 illustrates the experimental setup employed for this purpose. Five measuring points are considered: two on the prosthesis frontal plane and three on the sagittal plane, at different heights. Acquisitions are performed with three omnidirectional microphones (PCB Piezotronics 378B02 and two Roga AMV MI-17) and two sound pressure level meters (Bruel & Kjaer 2270). Raw microphones data are sampled at 48 kHz and filtered with an 'A-weighting' filter to account for the relative loudness perceived by the human ear. Each test consisted of 25 swing cycles in which the prosthetic leg reaches a

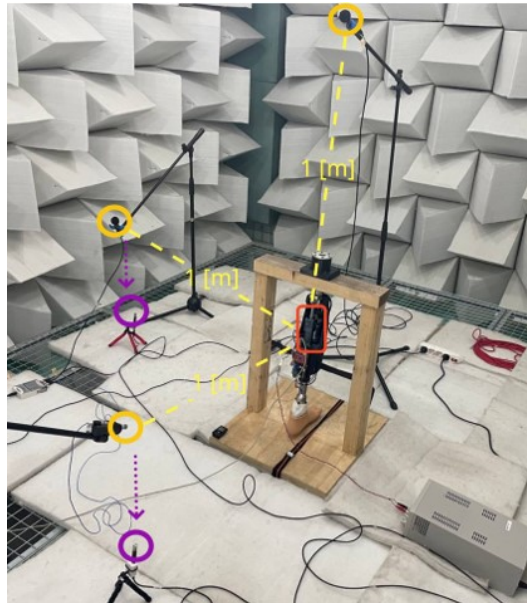


Fig. 2.22 Anechoic chamber experimental setup: measuring microphones (yellow), sound pressure level meters (purple) and actuation unit (red).

45° knee angle. The equivalent range of movement is scaled and used when testing the actuator outside of its prosthesis housing. The sound descriptors pertaining to the prosthesis compound are much higher than the one related to the actuator, with a maximum difference of around 27.5 dBA in the sagittal plane. By performing a frequency analysis, it is experimentally found that frequencies between 100 and 300 Hz are the most excited of the spectrum. Overall, these two results justify the redesign of an aesthetic chassis to account for noise reduction.

Within the domain of acoustic reduction techniques, passive solutions are favored over active alternatives due to their simplicity and the absence of additional circuitry requirements. Particularly, the preferred method entails the utilization of a metamaterial panel comprising arrays of Helmholtz resonators. This approach is chosen for its effectiveness in attenuating acoustic noise. The acoustic chassis, as depicted in Figure 2.23, is realized through 3D printing in thermoplastic polyetherimide resin. A parallel experimental campaign is executed to evaluate the noise reduction capabilities of the chassis. Experimental findings reveal that, owing to the resonators panel effect, the prosthesis becomes approximately 33 % and 18 % quieter than the original configuration in the frontal and sagittal planes, respectively. Furthermore, within the designated frequency range for the design, an average difference of 6 dBA is observed. These outcomes serve as a validation of the proposed solution, offering

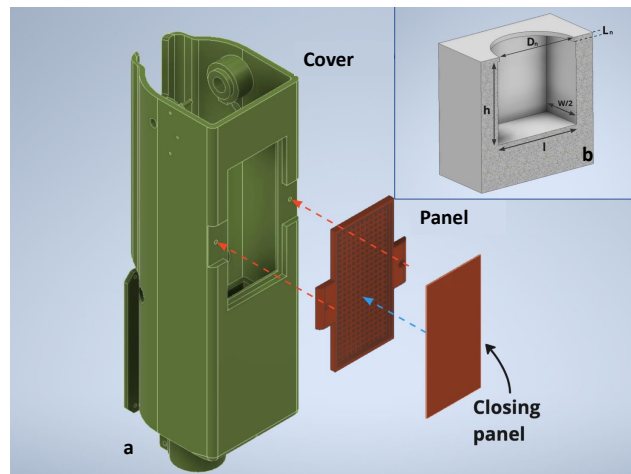


Fig. 2.23 Exploded view of the improved casing (a) and resonator cavity (b)

an effective mean to diminish acoustic noise emission without impeding the functionality of the actuation unit. A complete overview of the methodologies and results obtained is available in [114].

Aesthetic argument Finally, a crucial aspect of a prosthesis chassis is its aesthetic appearance. The visual appeal of the chassis can hold multifaceted implications for the embodiment of the device and may exert psychological influence on the overall acceptance of the prosthesis [36]. Hence, considerable effort has been invested in defining the geometric shape of the aesthetic cover for the prosthesis, aiming to achieve ergonomic resemblance to a natural limb and facilitate easy donning and doffing. A visual representation of the design cover is depicted in Figure 2.24. The cover is fabricated through 3D printing, utilizing a combination of thermoplastic polyurethane (TPU) for the flaps and acrylonitrile-butadiene-styrene (ABS) for the cover body. An HP MultiJet Fusion 3D printer is employed, offering the versatility to use different materials and colors. The cover comprises two half shells, with one securely attached to the prosthesis mechanically, and the other magnetically connected to form a cohesive unit.

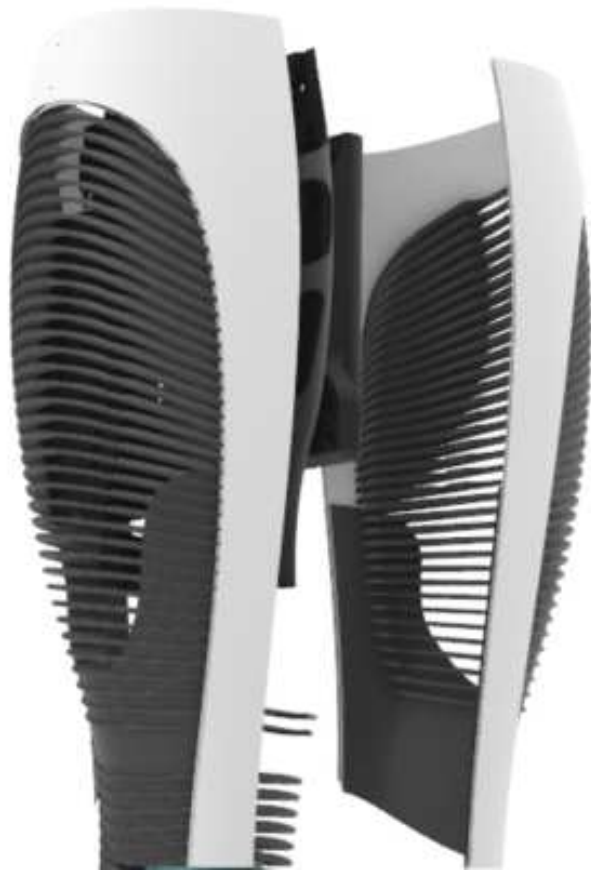


Fig. 2.24 Aesthetic cover

Chapter 3

Prosthesis Control Architecture

This chapter addresses the control architecture implemented within the prosthesis, incorporating two distinct layers of control: (i) low-level control, primarily responsible for managing current/torque control of the Electro-Hydraulic Actuator (EHA) and position control of the valve, and (ii) high-level control, where a finite state machine facilitates transitions between states and tasks (e.g., level ground walking and stairs ascent and descent), defining actuation control strategies.

The low-level control operates on the embedded electronics, while Simulink Real-Time is employed for the high-level control, establishing communication through the Controller Area Network (CAN) with the prosthesis. Before delving into the intricacies of each control layer, a brief overview of the embedded electronics is provided.

3.1 Embedded Electronics

The low-level control runs on a custom-made board, represented in Figure 3.1. A 32-bit MCU (TI C2000 TMS320F28379DPTPS) deals with the sensors and high-level communication and sends reference signals to the motors gate drivers. The EHA motor employs a Trinamic TMC6200 3-halfbridge gate driver with embedded current sensing, whereas a TI DRV8881 dual h-bridge motor driver is used for the valve motor. The board embeds an IMU (Bosch BN055), communicating via I^2C communication protocol with the micro controller. Both motor encoders communicate through SPI protocol, whereas the knee joint encoder is communicating

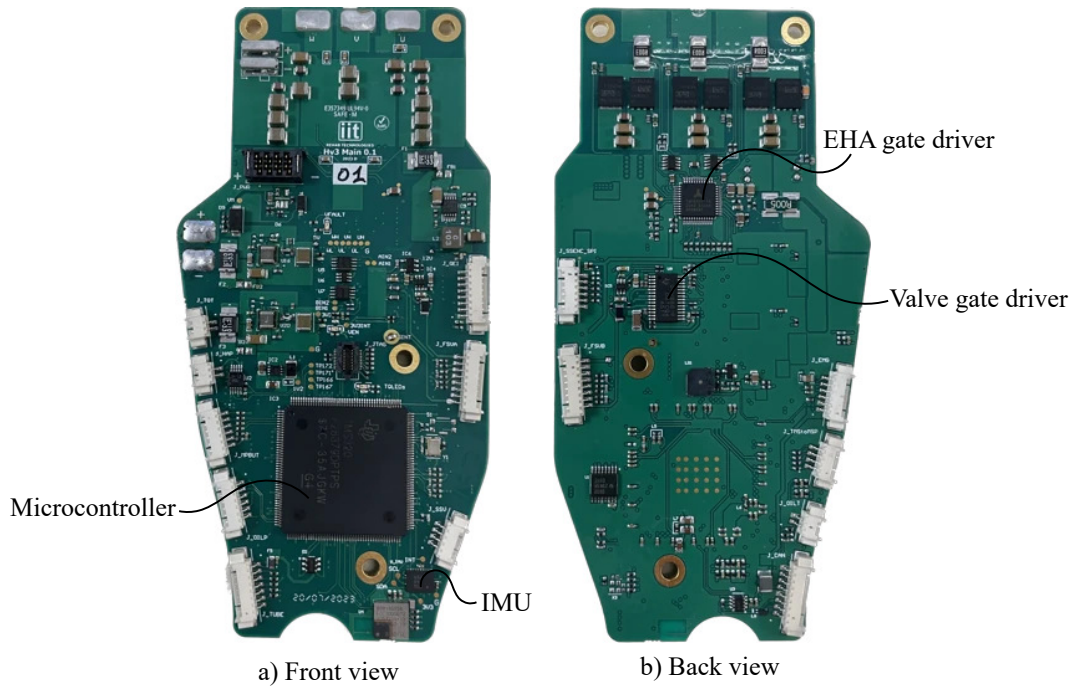


Fig. 3.1 Embedded electronics board

through UART. Finally, both pressure sensors and load cell are read as analog signals, with their respective amplifying circuits. Lastly, a 21.6 V, 3350 mAh battery pack supplies the electronics, with a 15.5 A maximum discharge current.

3.2 Low-level Control

The low-level control layer is programmed onto the embedded electronics, featuring two parallel loop, one dedicated to the valve DC motor control and the other to the brushless motor of the EHA. The valve control loop is relatively straightforward, mapping valve position directly to voltage Pulse Width Modulation (PWM) for the motor. Consequently, the position control loop is a Proportional-Integral (PI) controller operating at 1 kHz. The high-level control communicates the desired position through CAN. While the loop frequency could be increased, it is considered adequate, given that the time to transition from fully open to fully closed valve positions is approximately 100 ms (equivalent to 10 Hz).

In contrast, the low-level control for the EHA motor entails two cascaded loops. The outer loop is a torque-based controller accounting for the nonlinear dynamics

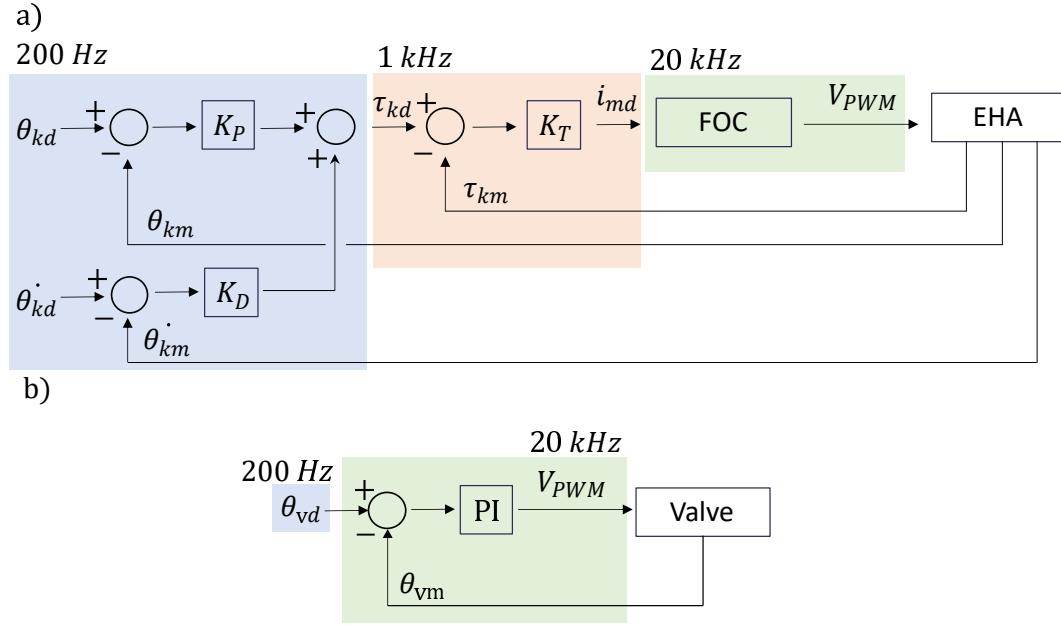


Fig. 3.2 Low level control architecture. a) EHA control strategy. b) Valve control architecture.

at the knee joint and compensating for friction. Feedback torque is computed from the pressure differential measurement, considering the overall transmission ratio for conversion into torque. The desired torque is communicated through CAN by the high-level control. This loop operates at 2 kHz, which is ten times faster than the high-level control. Its output is a motor current command, serving as input to a Field-Oriented Control (FOC) loop [122]. The latter considers current measurements, commands, motor electric angle, and outputs a Pulse Width Modulation (PWM) voltage signal for the motor phases. To ensure proper motor torque production, an initial phasing aligns phase A with the direct axis, setting the desired current on the direct axis to zero. This inner loop operates at 20 kHz, a conventional frequency for motor current control. Figure 3.2 schematically summarizes the low-level control architecture.

3.3 High-level Control

The high-level control system is structured around three Finite State Machines (FSMs), guided by heuristic principles. The outermost FSM serves the purpose of transitioning between tasks, specifically delineating the shift between walking and

stair ascending functionalities. Conversely, the two inner FSMs are dedicated to managing state transitions within their respective tasks. In Figure 3.3, a comprehensive schematic delineates the control architecture, offering a visual representation of the coordinated interplay between the three FSMs, both considering transitions between states and tasks.

By default, the prosthesis operates in walking mode, with two possible avenues for transitioning to stair ascending. One involves an automatic transition (1-2a in Figure 3.3), while the other is intentionally triggered by the user (1-2b in Figure 3.3). In the case of condition 1-2a, the heuristic employed to differentiate between walking and stair ascending relies on observing the prosthesis unloading condition. Specifically, during walking, the prosthesis experiences unloading with a partially flexed knee, whereas during stair ascending, the knee flexion angle is significantly lower, if not zero. To prevent false positives in the transition, the condition is defined by specific thresholds, as follows

- Normalized load $> 15\%$ body weight;
- Knee angle $< 5^\circ$;
- Shank angle $\leq -10^\circ$ (i.e., leg behind user);
- Walking speed > 0.3 m/s;
- Thigh angular speed ≤ -100 deg/s.

Notably, all four thresholds must be met within a 20 ms time window, and the thigh threshold is sequentially checked only after all the others are found to be true. The thigh angular speed heuristic serves as a reliable indicator that the user intends to either ascend a staircase or navigate an obstacle. This automated transition between walking and stair ascent is particularly advantageous, allowing the prosthesis to approach a staircase without the need for a pause at its base. Despite the effectiveness of the automatic transition, there are circumstances where the inclusion of an intentional stair task transition, as exemplified by transition 1-2b, proves beneficial. In this scenario, the user has the ability to independently trigger the transition by performing two stomps on the affected side. The detection of a double stomp involves an analysis of the rate of change of the load and the frequency at which this change occurs. The double stomp is considered intentional if performed within

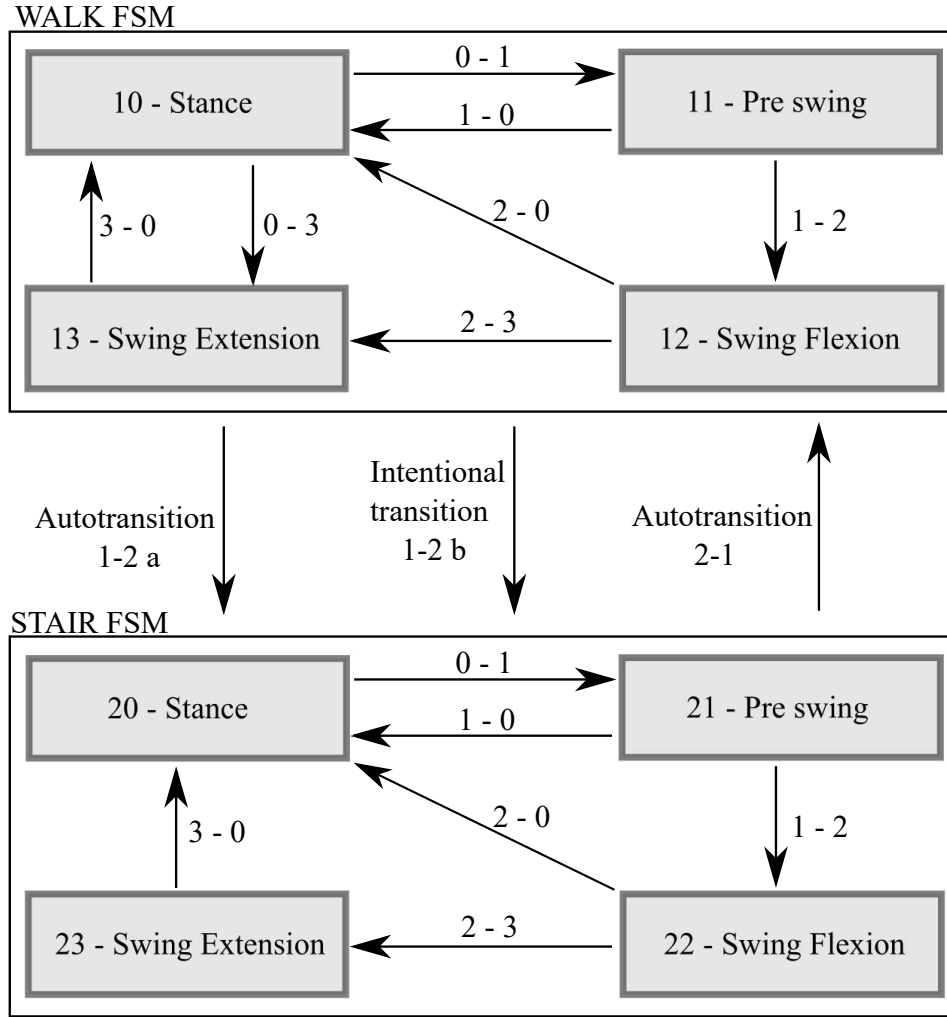


Fig. 3.3 High-level control loop architecture

the frequency band from 2.5 Hz to 10 Hz. While this intentional trigger is robust, precautions are taken to minimize the potential for false positives, necessitating the fulfillment of additional conditions to enhance the overall reliability of the transition. In the context of transitioning from walking to stair ascending (2-1), the process involves meeting two additional conditions. Firstly, the knee angle (θ_k) should approximate straightness ($\theta_k < 5^\circ$), and secondly, the prosthesis should be positioned behind the user ($\theta_s < -10^\circ$).

Conversely, when automatically transitioning from stair ascending to walking (1-2), the decision-making process is informed by the prosthesis unloading condition, applying the same heuristic rationale as observed in transition 1-2a. Namely, the conditions to be met are

Table 3.1 Walking Controller State Transitions

Transition	Condition
0 - 1	Thigh behind torso ($\theta_t < 0^\circ$) Prosthesis load $< 15\%$ body weight Knee angle $< 10^\circ$
1 - 0	Prosthesis load $> 15\%$ body weight
1 - 2	Prosthesis load $< 7.5\%$ body weight
2 - 3	Knee speed sign change ($\omega_k < 0$)
2 - 0	Prosthesis load $> 40\%$ body weight
3 - 0	Prosthesis load $> 40\%$ body weight Knee angle $< 2^\circ$
0 - 3	Prosthesis load $< 10\%$ body weight Knee angle $> 10^\circ$

- Normalized load $< 10\%$ body weight;
- Knee angle $> 10^\circ$;
- Shank angle $< -20^\circ$ (i.e., shank behind user);

The key distinction between this transition and transition 1-2a lies in the knee angle requirement, necessitating partial flexion when unloading the affected side. Once within a given task, the Finite State Machine (FSM) governs the transition of prosthesis behavior. In the walking task, the FSM comprises four discrete states: stance (ST), preswing (PS), swing flexion (SF), and swing extension (SE). The controller cyclically progresses through these states, starting from state 0, based on heuristic conditions outlined in Table 3.1. Notably, a preswing state is introduced to facilitate valve opening during the transition from stance to swing phases. Within each state, the prosthesis behaves as follows:

- *Stance*: The hydraulic valve is 80 % closed to guarantee proper stance flexion and extension damping (i.e., 0.75 Nms/radkg) while the actuation unit is backdriven;
- *Pre swing*: The hydraulic valve is moved to 20 % closed, to initiate the swing phase while the actuation is backdriven;
- *Swing flexion*: The hydraulic valve is fully open to provide minimal damping while the actuation unit is actively controlled;

- *Swing extension*: The hydraulic valve is fully open to provide minimal damping while the actuation unit is actively controlled.

Within the SF and SE states, the actuation controller features an impedance loop of the form

$$\tau_d = \beta_d \omega_k^n + \tau_{st} \text{sign}(\omega_k) \quad (3.1)$$

in which τ_d is the desired torque, β_d the desired damping to cancel, ω_k the knee angular speed, $n \geq 1$ represents the damping model considered (e.g., linear or turbulent) and τ_{st} is the static friction torque, given by the compound effect of hydraulic cylinder and knee joint friction. In contrast to a conventional impedance control law, this control methodology does not account for stiffness contributions, and the desired angular speed is set to 0. The selected damping is expressed as a percentage of the minimum impedance characterization of the mechatronic system, specifically when the Electro-Hydraulic Actuator (EHA) is back-driven, and the hydraulic valve is fully open. This characterization is identified during swing states, allowing for a modification of the prosthesis dynamic behavior without mandating a predetermined robotic movement at the knee joint. Complete damping and friction cancellation would render the prosthesis behavior akin to that of a free hinge. Initiating an inertially-coupled movement through the hip empowers the user with comprehensive control over the prosthesis, perceived as lighter and more easily ballistically swung than when in a passive mode. Termed as *homogeneous* dynamic compensation, this control philosophy ensures the absence of theoretically imposed exogenous inputs. Consequently, at the knee joint, there is a partial cancellation of the internal dynamics of the prosthesis. From a control perspective, this law is straightforward to implement and requires the tuning of just one parameter (β_d), aligning with user biomechanics and walking speed. Moreover, stair descent can be aptly executed using the same Finite State Machine (FSM). Given that the leg is loaded while the knee flexes, the FSM bypasses both preswing and swing flexion states, transitioning directly between stance (ST) and swing extension (SE).

In contrast, the Finite State Machine (FSM) governing stair ascent leverages increased power generation. The states mirror those of the walking FSM, and the conditions guiding transitions between them are detailed in Table 3.2. Within each state, the heuristics are:

- *Stance*: The hydraulic valve is 80 % closed to guarantee proper stance flexion and extension damping (i.e., 0.75 Nms/radkg). The EHA is backdriven;
- *Pre swing*: The hydraulic valve is moved to 20 % closed, to initiate the swing phase while the actuation is backdriven;
- *Swing flexion*: The hydraulic valve is fully open to provide minimal damping while the actuation unit is actively controlled to compensate for part of the prosthesis homogeneous dynamics and gravity;
- *Swing extension*: The hydraulic valve is fully open to provide minimal damping while the actuation unit is actively controlled to compensate for part of the prosthesis homogeneous dynamics and land with a vertical shank.

During swing flexion (SF) and swing extension (SE), two distinct control laws are applied, deviating from the walking FSM. In swing flexion, the controller seeks to partially cancel the homogeneous dynamics of the prosthesis while supplementing it with partial gravity compensation. The control law is of the form

$$\tau_d = [\beta_d \omega_k^n + \tau_{st} \text{sign}(\omega_k)] + [m_l g_d h_c \sin(\theta_s)] \quad (3.2)$$

in which m_l is the leg mass, g_d is the desired percentage of gravity to compensate, h_c is the center of mass longitudinal location with respect to knee joint center and θ_s the shank angle. The key disparity in Equation 3.2, relative to Equation 3.1, lies in the inclusion of the second term. This supplementary contribution is integrated to address

Table 3.2 Stair Ascent Controller State Transitions

Transition	Condition
0 - 1	Thigh behind torso ($\theta_t < -2^\circ$) Prosthesis load < 35 % body weight Knee angle < 10°
1 - 0	Prosthesis load > 50 % body weight Thigh in front torso ($\theta_t > -4^\circ$)
1 - 2	Prosthesis load < 7.5 % body weight
2 - 3	Thigh in front torso ($\theta_t > 0^\circ$) Thigh angular speed change of sign ($\omega_t < 0$)
2 - 0	Prosthesis load > 20 % body weight
3 - 0	Prosthesis load > 15 % body weight

the absence of ballistic leg motion during stair ascent. As elucidated in Chapter 1, initiating a ballistic swing in stair ascending presents a greater challenge owing to the lower leg dynamics compared to walking. Conversely, in swing extension, a virtual spring is introduced relative to the vertical shank position, further refining the control strategy during this phase. Namely the control law is as follows

$$\tau_d = \beta_d \omega_k^n + \tau_0 \text{sign}(\omega_k) - K_p(\theta_s) \theta_s \quad (3.3)$$

In this setup, $K_p(\theta_s)$ is gradually increased within 25 ms after the shank angle θ_s falls below 5° . This added term prevents the prosthesis from impacting the step during stair ascent. In contrast to the walking controller, a higher percentage of homogeneous dynamic cancellation is applied since the hip's dynamic contribution is lower than in walking.

Chapter 4

Testbench Experimental Results

The experimental validation of both the mechatronic system and the complete prosthesis involves a series of diverse testbench experiments. To facilitate discussion and categorization, the results are presented with a specific emphasis on power dissipation and power generation. Within each category, the analyses cover characterizations of individual subsystems as well as comprehensive evaluations of the overall prosthesis.

4.1 Power dissipation

Controllable damper Figure 4.1 presents the schematic of the testbench employed for evaluating the behavior of the valve. To prevent fluid cavitation, the circuit is preloaded with a hand pump (a). The pressure differential across the valve is measured by two pressure sensors (c, Kulite ETL-76A-190) and rectified through a hydraulic accumulator (b). The flow rate, generated by a commercial motor-pump driver (e, Kollmorgen BMS-1712A and Bucher APR0.75 pump), is measured using a piston flow meter (d, Max Machinery P214). The experiments involve imposing a range of flow rates between $-5 \div 5$ L/min, considering 10 discrete valve opening positions. Figure 4.2 illustrates the results within the hydraulic cylinder working space. Each dynamic behavior (dots sequence) is fitted with a second-order polynomial. Notably, for smaller openings, the quadratic term significantly dominates over the linear term, consistent with the methodology presented in Equation 2.25. The light grey area represents the power dissipation requirement outlined in the biomechanical specifications.

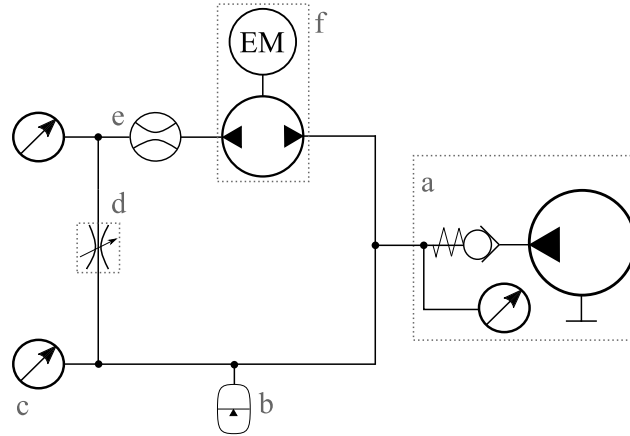


Fig. 4.1 Scheme of the testbench circuit. (a) Preload pump, (b) Gas loaded accumulator, (c) Pressure sensor, (d) Flow meter, (e) Motor-Pump driver.

The valve behavior is consistent up to 70 % open, aligning with the lower bound of the damping requirement (equivalent to 1 Nms/rad at the knee joint). Below 60 % closed, the valve exhibits appropriate controllability, with the damping force displaying an almost linear dependence on the valve opening position. Consequently, a broad spectrum of damping responses at the knee joint can be replicated, meeting the requirements of both stair and walk gait cycles. For further evaluation of its controllability, Figure 4.3 illustrates the equivalent quadratic damping coefficient (i.e., F/v^2) as a function of the valve position. This variable is determined through a least-square fit for a range of applied pressures or forces F and resulting flow rates or cylinder speeds v . It is inversely proportional to the intersected area between valve grooves and spool. The observations derived from Figure 4.3 reveal that, from fully open to around 30 % open, the valve demonstrates poor controllability, with minimal changes in damping within this range. However, ensuring precise control in this segment of valve openings is not crucial, as the valve is predominantly maintained fully open. Following this, the quadratic damping increases in a second-order polynomial fashion within the range of 30 % to 70 % open. While this behavior may not be optimal for control purposes, given that a slight change in valve opening (e.g., a positional control error) leads to a substantial damping alteration at the knee joint, it is worth noting that the valve is essentially inactive in this opening region. Therefore, it does not pose a significant issue at the prosthesis level.

Additionally, it is worthwhile to explore the damping behavior as the valve approaches closure. As evident, the damping response in relation to valve closure

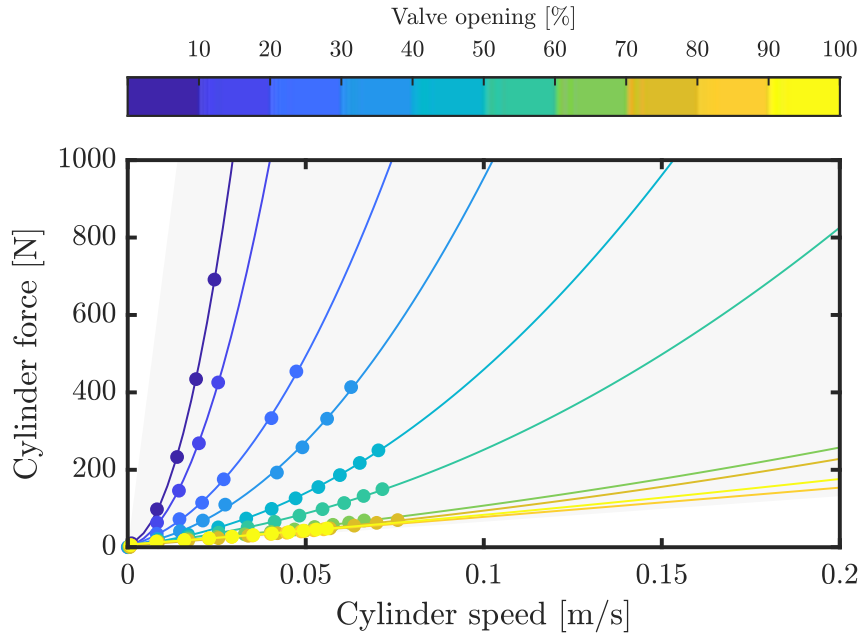


Fig. 4.2 Damping characteristic of custom, rotary, hydraulic valve as function of angular speed and valve closing position, presented in the knee operating space. Dots represents experimental tests, fitted with a quadratic curve. The green area represents the damping range defined from the biomechanical requirements.

follows a relatively linear trend, consistent with the theoretical analysis presented in Chapter 2. This linear behavior translates into improved controllability of the valve as it nears closure. At the prosthesis level, this outcome corresponds to more accurate damping control during tasks such as descending stairs (where the valve is commanded to slightly close to accommodate the change in transmission ratio) or precise adjustment of stance damping tailored to different users. Comparing the experimental results presented in Figure 4.2 with the numerical results presented in Figure 2.15, the valve behavior is more quadratic than linear. This is mainly related to the fact that the numerical analysis did not consider all the nonlinear effects. Nevertheless, both numerical and experimental results show an adequate controllability as function of valve aperture, objective of the custom design.

Overall Prosthesis In the context of the prosthesis's functionality, it is important to characterize the minimum impedance of the mechatronic system to ascertain the feasibility of ballistic walking solely with the damper or whether additional actuation assistance is necessary. Referring to Equation 2.1, the prosthesis is mounted to its pyramidal joint, restricting movements to a plane orthogonal to the gravity vector.

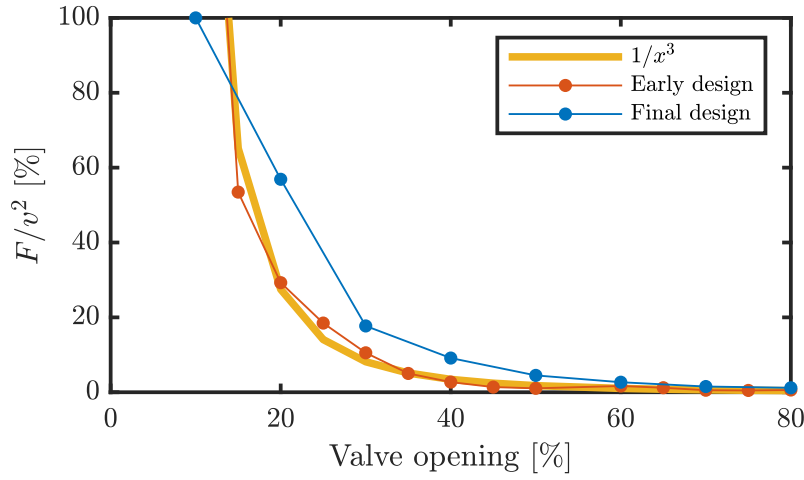


Fig. 4.3 Equivalent quadratic damping coefficient F/v^2 is computed as a least-square fit for a range of applied pressure differences or forces F and resulting flow rates or speeds v .

Notably, contributions from hip torque and gravitational forces are disregarded in this setup.

The experimental setup involves commanding the EHA to track a trapezoidal speed profile at the knee joint, exploiting an impedance controller, integrated with torque and current control loops. Each test iteration encompasses 10 swing cycles, conducted at four distinct angular speeds (5, 10, 15, and 20 rpm). Focusing solely on steady-state speed data points enables the exclusion of inertial contributions from both the prosthesis and the actuation unit, isolating effects related to the internal dynamics, including damping and friction. The prosthesis is equipped with a 28 cm passive foot prosthesis, designed for a 50th percentile built height. Considering the nonlinear transmission ratio, knee joint variables are translated into hydraulic cylinder variables. Namely, the hydraulic cylinder force is directly measured through pressure sensors, whereas the knee joint encoder is used to sense angular speed and then converted into linear speed considering the transmission ratio.

Figure 4.4 illustrates the extracted data points and the corresponding first-order polynomial fittings for both flexion and extension cycles. Although both behaviors are effectively approximated by a linear model, subtle differences exist, indicating that extension is inherently more damped than flexion. The static friction values are quantified as 31.5 N and 38.5 N for flexion and extension, respectively. In terms of minimum damping, the system exhibits values of 1874 N·s/m and 1946 N·s/m for flexion and extension. Extrapolating these results and considering an average

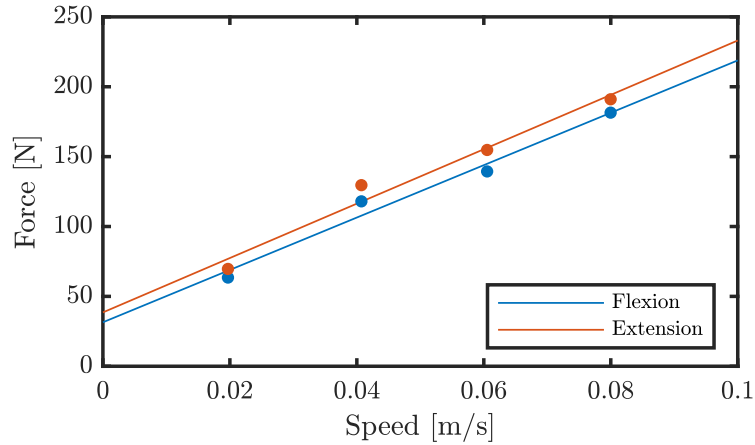


Fig. 4.4 Minimum impedance characterization in flexion and extension cycles. Dots represent data points, whereas lines their respective fitting.

transmission ratio of 37 mm, implies a minimum damping of $2.6 \text{ N} \cdot \text{m} \cdot \text{s}/\text{rad}$ at the knee joint.

4.2 Power generation

EHA tracking capabilities

Prosthesis tracking capabilities To evaluate the actuator tracking capabilities, a controlled motion is induced in the prosthesis along the sagittal plane by commanding a sinusoid at the knee joint. The prosthesis setup and mounting is presented in Figure 4.5. The waveform has an amplitude of 60° at a frequency of 1 Hz, implemented

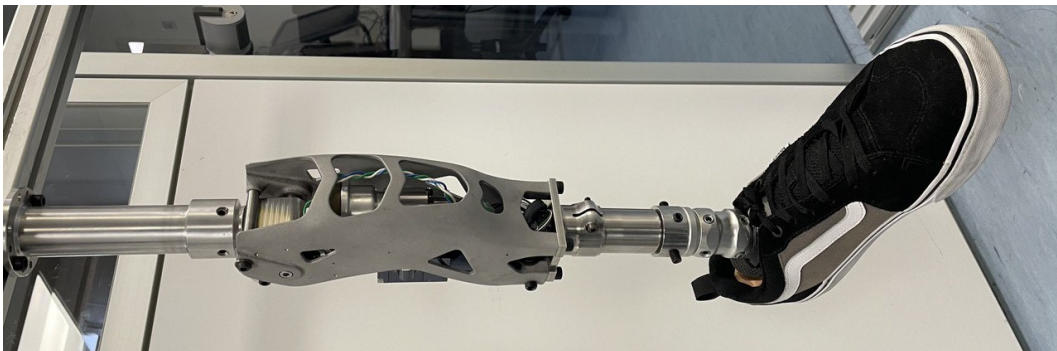


Fig. 4.5 Testbench mounting of the prototype prosthesis.

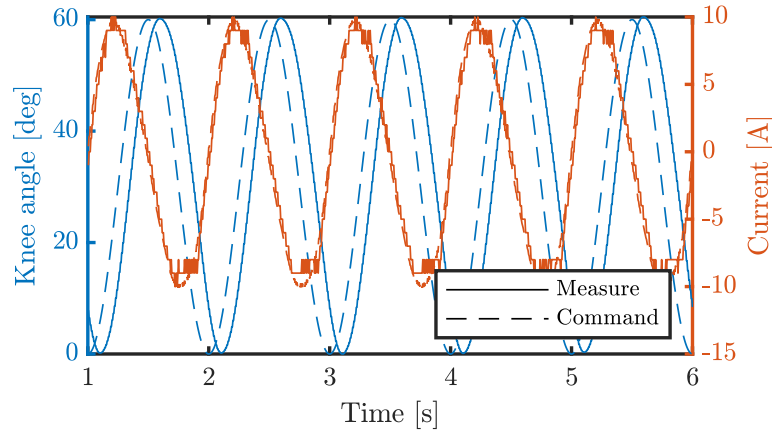


Fig. 4.6 Prosthesis active tracking, knee angle (blue), motor current (orange).

through an impedance controller. To accurately mimic a swing cycle, specific conditions are enforced. The hydraulic valve is maintained fully open, and the motor current is restricted to 11 A to prevent potential damage to the motor windings.

Figure 4.6 provides a visual representation of the angle and current tracking for five representative cycles. Notably, an average delay of 100 ms is observed between the commanded and measured angles. This delay, while present, is within an acceptable range and does not significantly impact the overall tracking performance of the actuator, as 100 ms is the average time elapsed between heel strike and foot flat. Nonetheless, the experimental testing aims at evaluating the prosthesis active capabilities. In real case scenarios, the EHA is not intended to impose a trajectory; rather, it compensates for part of the prosthesis passive dynamics.

The assessment of current tracking reveals more effective performance during flexion (characterized by positive current) compared to extension. This distinction is likely due to higher impedance encountered during the extension phase. Importantly, this experimental setup represents a worst-case scenario as it does not account for any hip contribution. The absence of hip torque might contribute to the challenges observed in accurately tracking current during the extension phase. Further investigations and optimizations could be explored to enhance performance under these specific conditions.

Prosthesis active capabilities To thoroughly evaluate the prosthesis's maximum power generation capabilities, an isolated test is conducted. The prosthesis is di-

Table 4.1 Prototype max. power generation capabilities

Knee variable	Motor max. current [A]	
	5.5	11
Angle, θ_k	58.36°	89.9°
Flexion speed, ω_k	21.93 rev/min	32.81 rev/min
Extension speed, ω_k	24.48 rev/min	48.53 rev/min
Flexion torque, τ_k	8.2 N · m	13.47 N · m
Extension torque, τ_k	6.03 N · m	10.82 N · m
Flexion power, P_k	8.4 W	43.4 W
Extension power, P_k	14.6 W	51.9 W

rected to track a trapezoidal angle trajectory with a 90° amplitude at a frequency of 1 Hz. This assessment is performed under two distinct current limit conditions: 11 A, representing challenging scenarios, and 5.5 A, simulating nominal conditions. The detailed results of these experiments are systematically presented in Table 4.1. As evident, the device can achieve a 90° angle when limiting the motor current to 11 A. In this configuration, the maximum angular speed is reached during extension at 48.53 rev/min, with a peak flexion torque of 13.47 N · m. The maximum exerted power is 51.9 W, occurring during knee extension, likely due to the added contribution given by gravity. Conversely, tests conducted under nominal current constraint reveal that the prototype is unable to reach the target angle, with a maximum of 58.36°. Extension occurs at a reduced angular speed of 24.98 rev/min, accompanied by lower torque production at 8.2 N · m during flexion. These results demonstrate that the prototype prosthesis can actively perform a swing cycle even without hip contribution. This scenario serves as a worst-case representation, highlighting the actuation unit's maximum power generation capabilities.

Chapter 5

Prosthesis Experimental Results

Experimental trials are conducted with a healthy individual wearing an able-body adapter, designed to emulate the dimensions associated with a 50th percentile prosthetic height. Figure 5.1 showcase the setup to experimentally test the developed prototype.



Fig. 5.1 Setup to experimentally evaluate the performance of the developed prosthesis.

5.1 Treadmill walking

The evaluation of the walking performance involves comprehensive tests conducted with both a widely used commercial MPK prosthesis, specifically the Ottobock C-Leg 4, and the developed prototype. These assessments take place on a AMTI split-belt, force-instrumented treadmill, covering a spectrum of representative walking speeds, including 0.5, 0.8, and 1.1 m/s. To capture lower limb kinematic data, an infrared motion capture system (Vicon, Oxford, UK) is employed.

The experimental protocol unfolds over a duration of 90 s for each walking condition, with the sequence starting initially with the use of the commercial prosthesis. This a systematic and comparative analysis between the established commercial device and the newly developed prototype, across a range of walking speeds, offering

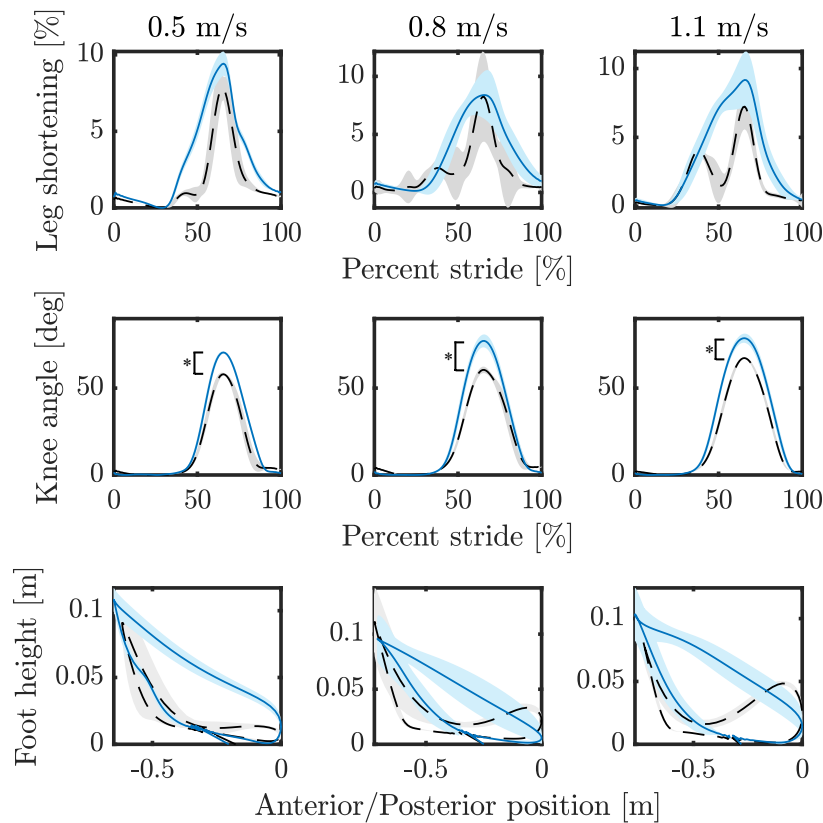


Fig. 5.2 Prosthesis walking trials at three representative walking speeds. Black dashed curves and gray bands represent commercial prosthesis results. Blue solid curves and light blue bands represent prototype results. Bands correspond to $\pm$ a standard deviation from the average for both commercial and prototype prostheses.

valuable insights into the performance characteristics of the proposed prosthesis. Figure 5.2 displays collective results for both the commercial (black) and prototype (blue) prostheses, segmented by strides, across three walking speeds. The depicted data represent averages over approximately 20 strides, with colored bands indicating a standard deviation around the mean. The top row of Fig. 5.2 illustrates the percentage of leg shortening, representing the vertical distance between pelvis and toe markers throughout the stride. A null percentage indicates the maximum distance between the markers. Therefore, a higher leg shortening corresponds to higher knee flexion on the sagittal plane. In all three walking speeds, this metric is statistically greater in the developed prototype than in the commercial device, with very similar peaks at 0.8 m/s. This is essentially reflected in the knee angle, mid row of Figure 5.2. In these row of plots, the peak knee angle is consistently higher in the developed prototype than in the commercial device. Since, the leg shortening peaks are similar in the two devices, the combination of the two highlights that the commercial prototype is subject to greater frontal plane movement than the prototype. Unfortunately, data on the frontal plane movement are not presented as the able-body adapter strongly affects the measure. Finally, the bottom row of Figure 5.2 showcase the foot height as function of the anterior/posterior position of the foot marker. Essentially, these plots represent the sagittal plane movement of the foot during the stride. In accordance with leg shortening and knee angle trends, the foot height peaks are consistently higher in the prototype than in the commercial device, apart from 0.8 m/s in which the peaks are similar. Then, it is qualitatively possible to highlight that the foot path in all three cases is on average higher in the prototype than in the commercial device. This translates in an improved toe

Table 5.1 Walking results. Mean values of the analyzed metrics.

Variable	Walking speed [m/s]					
	0.5		0.8		1.1	
	C	P	C	P	C	P
Peak knee angle [°]	58	70.5	60.5	77.2	67.3	78.7
Peak hip angle [°]	36.7	46.6	35.6	51.2	37.1	55.2
Stride length [m]	0.62	0.65	0.72	0.71	0.73	0.77
Stride time [s]	1.24	1.3	0.9	0.89	0.67	0.7
Peak leg short. [%]	7.8	9.4	8.2	8.4	7.2	9.2
Autonomy [strides]	—	9350	—	5400	—	4260

C—commercial device, P—prototype.

clearance for the prototype that could potentially lead to an avoidance or a reduction of falls [92]. Table 5.1 provides a summary of key findings for both commercial and prototype prostheses at various walking speeds. Three pair-wise t-tests are conducted to assess any significant difference in the peak value of the knee angle between the commercial and prototype versions, at each walking speed. The significance value was set to $\alpha = 0.05$. The results reveal a p-value lower than 0.001 in all three walking conditions. Moreover, it is crucial to comprehend the power and current consumption of the prototype in the three distinct walking conditions. Specifically, Figure 5.3 illustrates the trends in power and current per stride, encompassing the three investigated walking speeds. As discernible from the two prominent power peaks, the prototype utilizes active power during the swing phase, while the stance phase is tempered by passive hydraulics. The most substantial average power peak is associated with swing extension, amounting to 42.5 W with a corresponding current peak of approximately 10 A.

In addition to kinematic and kinetic considerations, an energetic analysis is undertaken to characterize the prosthesis autonomy. Accounting for Joule, gate driver, and electronics power losses, the average energy consumption per stride is quantified at 27.62 W, 47.5 W, and 60.58 W for walking speeds of 0.5, 0.8, and 1.1 m/s, respectively. Delving into practical implications, with a 21.6 V battery pack with a capacity of 3350 mA · h, the prosthesis demonstrates the capability to execute approximately 9350, 5400, and 4260 strides at the corresponding walking speeds. Translating these figures into daily usage scenarios suggests a usage span ranging from a maximum of 3 to a minimum of 1.5 days, aligning with the average daily step count reported for transfemoral amputees [123]. This comprehensive analysis not only sheds light on the biomechanical performance but also emphasizes the practicality and life span of the developed prosthesis.

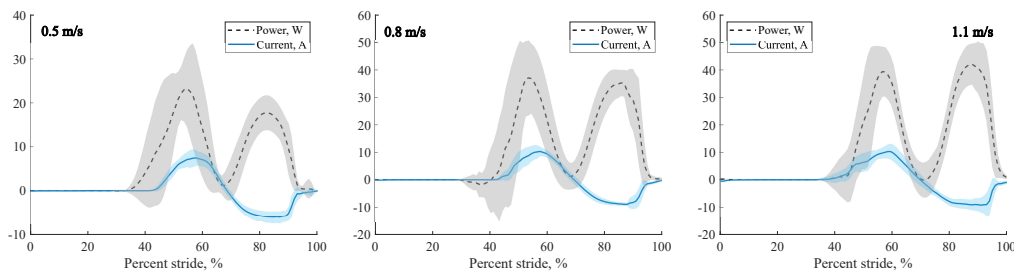


Fig. 5.3 Average generated power (black) and current (blue) per stride for three walking speeds (i.e., 0.5, 0.8 and 1.1 m/s)

5.2 Stairs Ascending

Stair ascending experiments are exclusively conducted with the prototype prosthesis, as obtaining kinematic data for the commercial device was not feasible. However, various studies have presented stair ascent kinematic data for commercial MPKs [124, 125, 58]. Experiments are carried out on a six-step staircase, considering two distinct ascending gaits: single step to (ST) and step over step (SO). The former represents the conventional gait for transfemoral amputees, while the latter is the common gait observed in able-bodied individuals. Results for knee angle and exerted torque are presented in Fig. 5.4. In both ascending gaits, significant peak knee flexion is attained, reaching 35° for ST and 39° for SO. Step clearance is achieved by applying torque at the joint, as detailed in section 3.3. The peak torque values, $5.4 \text{ N} \cdot \text{m}$ for ST and $7.2 \text{ N} \cdot \text{m}$ for SO, correspond to knee swing flexion, with no assistance provided during the stance phase.

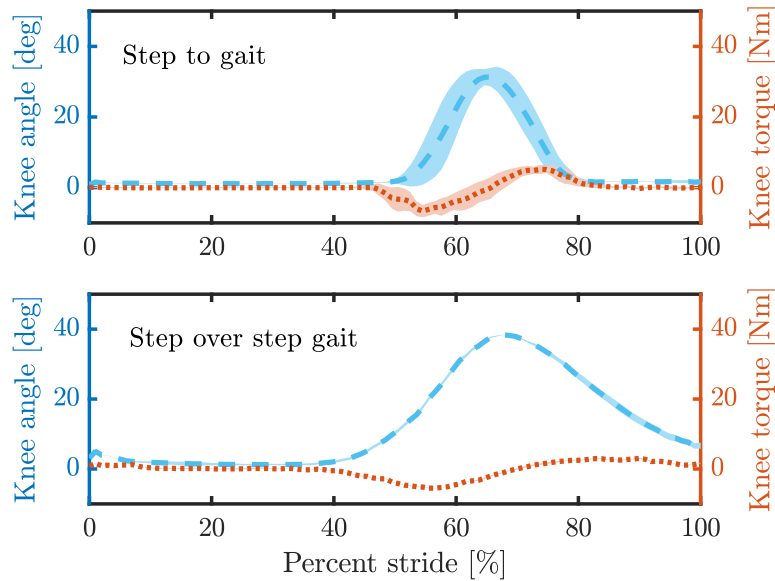


Fig. 5.4 Prosthesis stairs results considering two ascending gaits, step to (top) and step over step (bottom). In blue, average, parsed knee angle and band of $\pm$ a standard deviation. In orange, average, parsed, exerted torque and band of $\pm$ a standard deviation.

Chapter 6

Discussion

6.1 Prototype Features

The device is a compact and lightweight (2.15 kg) partially powered knee prosthesis designed to provide active power during the swing phase of various locomotion tasks. This choice results from a balance between power generation capabilities and a compact size to reduce inertial backdrivability. From a biomechanical perspective, focusing on the swing phase is reasonable, as it is the locomotion stage most influenced by external perturbations. Therefore, incorporating a small actuation unit effectively enhances the robustness of the swing movement, reducing the risk of scuffing and falling. Conversely, during load support phases, the user retains some control over the prosthesis movement, as the combination of device and user form a closed kinematic chain. For instance, during stair ascent, hip torque exertion and handrail assistance can compensate for the lack of active power at the knee joint.

The system utilizes a customized Electro-Hydraulic Actuator (EHA) in conjunction with a flow-control valve. The valve is specifically designed to enhance controllability as function of the actuating angular motor position (see Fig. 4.2). In experimental comparison with a commercial device, the specially designed valve demonstrates superior controllability within the range of 0 % to 30 % open, a range often requiring modulation. For instance, during stair descent, the valve position is adjusted to accommodate the varying transmission ratio between linear and rotary motions. Furthermore, characterizations on the testbench indicate that the device can operate within the entire damping region defined by the locomotion require-

ments of healthy subjects (see Section 2.1). The dynamic behavior of the valve aligns well with a second-order polynomial, consistent with theoretical expectations. Specifically, for small aperture angles, the quadratic term predominates over the linear one, though still within acceptable limits. It is crucial to account for radial clearances in the design of this component. Notably, experimental findings reveal that a radial clearance of $10\text{ }\mu\text{m}$ represents an optimal trade-off between mechanical friction and valve sealing. From a manufacturing standpoint, achieving this clearance proves challenging, especially for small valve dimensions (i.e., spool diameter of 7 mm, axial length of 15 mm). Therefore, I recommend tempering and grounding the relevant surfaces, leaving sufficient machining allowance. Subsequently, reworking and honing down the critical diameters by hand will ensure proper functionality.

The Electro-Hydraulic Actuator (EHA) has undergone a top-down design approach, utilizing a multi-objective optimization problem solved by a genetic algorithm. The optimization objectives included minimizing its volume for compactness and maximizing its efficiency for optimal performance. The design refinement involved finite element analyses for the electric motor and computational fluid dynamic analyses for the gerotor pump. Clearance definition for the hydraulic unit is a critical consideration. While increased thigh clearances generally enhance hydraulic efficiency, they concurrently increase friction, diminishing the power generation capabilities of the device. Another crucial aspect is the impact of clearance on backdrivability. In addition to inertial backdrivability, a metric directly related to actuator dimensions, a tight clearance reduces velocity-based losses, thus lessening the backdrivability of the device. This characteristic is specific to hydraulic actuators, as electro-mechanical ones primarily experience torque-based losses associated with backdrivability. Numerical validations indicate a maximum efficiency of approximately 75 % for the gerotor, with motoring operations exhibiting higher losses than pumping operations. The electric machine's numerical characterization reveals a maximum efficiency of around 78 %. Despite the motor manufacturer guaranteeing a maximum efficiency of 88 %, a decrease is anticipated due to the frameless nature of the motor, which is susceptible to specific losses associated with mounting and casing conditions. Nevertheless, the motor operates within a fluid environment, enabling it to withstand higher currents than in conventional applications.

The system is highly integrated, being that active and passive components share the same hydraulic cylinder for power generation and dissipation. Simultaneously, the actuation unit operates entirely submerged in fluid, enhancing performance by

eliminating the need for sealing between the motor and pump. Considering an average transmission ratio, the device provides a minimum damping impedance of $2.6 \text{ N} \cdot \text{m} \cdot \text{s}/\text{rad}$ at the knee joint and a maximum static friction of approximately 39 N. Under power generation conditions, specifically during a sinusoidal trajectory tracking task, the prototype can generate up to 52 W of power. It achieves a peak flexion of up to 90° and a maximum rotational speed of 49 rev/min. This indicates that the actuation unit can effectively replicate the swing phase of a healthy limb without requiring hip effort. However, it's essential to note that the experiment represented a worst-case scenario, as hip torque exertion always introduces dynamic behavior at the knee joint. This dynamic behavior is less pronounced during stair ascent compared to walking. Finally, active tracking tests demonstrated that the developed prosthesis consistently tracks a sinusoid up to 1 Hz with an average delay of 100 ms. This delay is within acceptable limits, as it is lower than the average time between heel strike and foot flat. While the current consumption is somewhat high, with peaks reaching 10 A, it remains entirely manageable by the motor. To assess winding thermal capabilities, a constant supply of 20 A has been applied to the motor phases, revealing no excessive temperature increase.

Walking tests were conducted on an healthy subject using a dedicated able-body adaptor, evaluating three different speeds (0.5, 0.8, 1.1 m/s) with both the developed prototype and a commercial Microprocessor Knee (MPK). In all scenarios, the prototype consistently exhibited a higher peak knee flexion compared to the commercial prosthesis, indicating greater leg flexion, as depicted in the middle row of Fig. 5.2. The bottom row of Fig. 5.2 illustrates foot movement on the sagittal plane. At both speeds of 0.5 and 1.1 m/s, the prototype shows a higher maximum foot height compared to the commercial prosthesis, while maintaining a comparable stride length. Results at 0.8 m/s reveal no significant differences, aligning with observations from the leg shortening metric. On average, the prototype's foot height is greater than that of the commercial prosthesis, indicating improved toe clearance. Enhanced toe clearance reduces the probability of scuffing and falling, a highly desirable feature for amputees. Furthermore, an energetic analysis was conducted, considering a 21.6 V battery pack with a capacity of 3350 mA · h. The prosthesis can sustain 9350, 5400, and 4260 strides at the respective speeds, translating to a usage range from a maximum of 3 to a minimum of 1.5 days. This accounts for the average number of steps per day performed by Transfemoral Amputees (TFAs) [123]. Finally, supplementing an MPK prosthesis with a small actuation unit can

enhance the robustness of the swing phase during walking and reduce the likelihood of falling.

The prototype underwent testing in a stair ascent task, considering two conventional ascending gaits: step to and step over. Step to, primarily used by Transfemoral Amputees (TFAs), involves the affected side being trailed by the contralateral one. On the other hand, step over is the conventional gait of healthy subjects. The experiments yielded promising results, demonstrating a peak knee flexion of up to 39° during the step-over-step gait. Comparative literature on Microprocessor Knee Prostheses (MPKs) often reports minimal knee flexion during step to [125] and the inability to perform a step-over-step (SO) gait, except for the Ottobock Genium [126], which relies on hip movement initiation. This accomplishment suggests that the inclusion of a small actuation unit also provides significant added value during stair ascending, enabling the negotiation of stairs in a step-over fashion—an essentially impossible feat with conventional MPKs. This result enhances the versatility of the device by unlocking new functionalities. For example, the actuation unit can also be utilized for obstacle avoidance, a perturbation difficult to navigate with commercial MPKs.

Lastly, a crucial innovation of the prosthesis lies in its control strategy. Specifically, the actuation unit is commanded to cancel part of the homogeneous dynamics of the prosthesis. This theoretically results in no imposed robotic movements, as no exogenous inputs are commanded to the prosthesis. In terms of integration and biomechanical embodiment, this feature translates to complete prosthesis control for the users, allowing them to retain agency over the device.

6.2 Comparison with Other Prosthesis Types

The presented prosthesis concept closely resembles a hybrid actuation scheme incorporating a variable hydraulic damper in parallel with a continuously engaged electric actuator. Notably, if the actuator acts on the damper rod, as demonstrated by Lee et al. [72], a similar joint-based nonlinearity in the gear ratio is achieved. Both designs share the common principle of sizing the motor and gearing to allow for power injection without compromising the natural dynamics of the passive leg. The variable damper regulates passive activities and provides high-torque braking.

A notable distinction lies in the placement of the motor within the hydraulic circuit in the presented prototype, in contrast to the electro-mechanical device proposed by [72]. A key difference is the non-rigid transmission inherent in the presented prototype due to fluid leakage paths inside the actuator, as opposed to the inherently rigid mechanical (e.g., toothed) transmissions. While a non-rigid transmission might be perceived as less efficient than a rigid one, it contributes to a natural impedance that results in smooth motion and greater adaptability to the user's perception. Unlike a series elastic actuator (SEA), this impedance is velocity-based rather than position-based, offering a smooth, controllable, and forgiving experience for the user. A future study focusing on efficiency optimization and user perception could provide insights into the preferable technology for integration into a partially powered lower limb prosthesis.

A limitation of the device relies on its minimum passive impedance. Although its value is contained (i.e., $2.6 \text{ N} \cdot \text{m} \cdot \text{s}/\text{rad}$), it may not be perceived as low as needed. In fact, the biomechanical requirement for healthy subjects suggests a minimum impedance of $1 \text{ N} \cdot \text{m} \cdot \text{s}/\text{rad}$, to guarantee appropriate swing natural dynamics. To improve this value, different options are available. First, the actuation reflected inertia can be decreased, either considering a lower transmission ratio or reduce the actuation power output. Both options can be implemented as experimental results showcased the ability of the actuation unit to perform a healthy subject movement without considering the hip contribution. Another possible avenue is to reduce the impedance improving fluid paths and the hydraulic resistance of the passive hydraulics compound. This second option seems more difficult to realise both for manufacturing limitations and highly nonlinear behavior of the hydraulic valve. Another limitation or rather a potential optimization of the device is the inclusion of a fixed orifice, extension check valve, to bypass the proportional valve. In fact, by looking at the knee biomechanics during the most relevant locomotion activities, the impedance behavior during extension does not substantially change and the lower the better. To further improve this locomotion phase, a spring biased towards extension can be inserted in parallel to the mechatronic system. Nevertheless, a trade-off must be accepted as such spring will work against knee flexion.

Chapter 7

Conclusion

Robotic knee prostheses play a pivotal role in restoring essential functionalities for individuals with transfemoral amputations. Currently, the prevalent devices in the market are microprocessor knees (MPKs), offering the ability to modulate the prosthesis' passive behavior but lacking the capability to inject active power at the knee joint. Although considerable research has been devoted to active prostheses, aimed at emulating the functionalities of a healthy knee joint through actuation units, the advantages of active assistance have yet to outweigh the drawbacks associated with increased size, weight, noise, and asynchronous movement between the prosthesis and user. While some literature explores the integration of passive and active elements in prostheses, determining the optimal trade-off between power generation and the natural dynamics of a healthy leg remains unclear. Addressing this gap, the presented doctoral work sought to enhance the versatility and robustness of conventional MPKs by supplementing it with a small hydraulic actuation unit. The design rationale focuses on swing activities due to their potential to limit the size and weight of the actuation unit, increased susceptibility to external perturbations, and the preservation of user agency by leveraging the beneficial natural dynamics of a passive leg.

To achieve this goal, I proposed a novel design for a partially powered knee prosthesis, combining a linear electro-hydrostatic actuation with a flow-control valve. Through a fully custom mechatronic design process, a compact, lightweight, and backdrivable device was developed, demonstrating sufficient active power capabilities to support the swing phase during both walking and stair ascending. Ex-

perimental characterizations on a testbench were complemented by validation on a healthy subject using an able-bodied adapter, showcasing superior performance compared to conventional commercial solutions. During walking tests, the prosthesis exhibited enhanced robustness in swing motion, increasing toe clearance, thus potentially reducing the likelihood of scuffing, stumbling or falling. This feature translates into improved stability, a critical factor for transfemoral amputees. In stair ascending tasks, the developed prototype not only displayed substantial knee flexion compared to conventional MPKs but also demonstrated the ability to negotiate stairs in a step-over fashion. This is particularly advantageous for amputees, aligning with the conventional gait of healthy subjects and enabling faster stair climbing compared to a step-to gait.

Future endeavors will focus on validating these promising results within the transfemoral amputee population, contributing to the ongoing evolution and refinement of advanced prosthetic technologies.

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