Human-Centered Design Trade-offs for Semi-Powered Knee Prostheses: A Review

Human-Centered Design Trade-Offs for Semi-Powered Knee Prostheses: A Review

Andrea Berettoni, *Member, IEEE*, Josephus J. M. Driessen, *Member, IEEE*, Marco Puliti, *Student Member, IEEE*, Giacinto Barresi, *Member, IEEE*, Carlo De Benedictis, Carlo Ferraresi, and Matteo Laffranchi

***Abstract*—For many decades, developments of knee prostheses have shown a dichotomy regarding fundamental working principles. The industry has mainly emphasized on quasi-passive hydraulic solutions, whereas most research works have focused on powered devices, employing electric actuation. The former have an energetically passive effect at the knee joint, for which they often lack in providing versatility and movement robustness for the wearer. Powered prostheses can address these deficiencies, but are often rejected as they struggle to fulfill other user needs (e.g., weight and acoustic noise). Correspondingly, recent studies have emerged that attempt to significantly attenuate the deficiencies of fully powered prosthesis knees, partially sacrificing on device versatility. Recognizing the state-of-the art difficulties in balancing active assistance and user needs fulfilment, this work analyses human-centered design perspectives and their prospects for prosthetic development, in light of the often diverging user needs. We conclude that various types of both explored and yet unexplored semi-powered solutions may have the potential to provide the better trade-off between quasi-passive and fully powered prosthetic devices.**

***Index Terms*—Biomechanics, human-centered, semi-powered, transfemoral, prosthesis.**

I. Introduction

PROSTHETIC knees are commonly categorized according to the inclusion of electronics and their ability to exert positive power at the joint. Generally, they are categorized in passive, quasi-passive and powered prosthetic knees. Passive prostheses do not rely on any electronics and cannot exert positive work. Rather, they provide resistance or damping at the knee joint that is either constant or variable, depending on the mechanical architecture. Mechanical locks, linkage mechanisms, hydraulic or pneumatic dampers [1], [2] are often exploited to distinguish between load support phases (e.g., stance) and inertially-driven motion (i.e., swing phase), passively adjusting to the user intent. They are generally inexpensive but present limitations in locomotion activities other than level ground walking at fixed speed [3].

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Andrea Berettoni and Marco Puliti are with the Rehab Technologies IIT-INAIL Lab, Italian Institute of Technology, 16163 Genova, Italy, and also with the Department of Mechanical and Aerospace Engineering, Polytechnic University of Turin, 10129 Turin, Italy (e-mail: andrea.berettoni@iit.it).

Josephus J. M. Driessen and Matteo Laffranchi are with the Rehab Technologies IIT-INAIL Lab, Italian Institute of Technology, 16163 Genova, Italy.

Giacinto Barresi is with the Rehab Technologies IIT-INAIL Lab, Italian Institute of Technology, 16163 Genova, Italy, and also with the Bristol Robotics Laboratory, University of the West of England, BS16 1QY Bristol, U.K.

Carlo De Benedictis and Carlo Ferraresi are with the Department of Mechanical and Aerospace Engineering, Polytechnic University of Turin, 10129 Turin, Italy.

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Conversely, quasi-passive prostheses include electronics to actively control the amount of resistance or damping at the joint. In literature, they are also referred to as adaptive [4], semi-powered [5], variable damping [6], semi-active [7], [8], [9], [10], [11], or (passive) microprocessor-controlled prosthetic knees (MPKs) [3], [12], [13]. However, conversely, the terms semi-active, semi-powered and MPK are also occasionally used to refer to prostheses that include actuation that can deliver positive power, which is technically correct. In this work we want to make a clear distinction between several types of prosthesis that can provide a degree of positive power and those that cannot, so to avoid confusion we refrain from using these terms in the remainder of the manuscript. Nominally, quasi-passive prostheses are more versatile and robust than passive prostheses, as they can more easily adapt to different locomotion activities (e.g., walking speed adaptation, downward slopes, stair descent and sitting down), generally governed by energy dissipative or neutral behaviour. However, they are limited in energetically positive activities such as stair climbing and sit-to-stand [14].

Lastly, powered prosthetic knees are – in addition to being controlled electronically – also able to exert positive work at the knee joint, by including actuation units. Literature also defines them as actuated prostheses [15], active prostheses [16], [17], or active MPKs [18]. Classically, such prostheses are designed with the aim to address the full torque-speed spectrum of multiple types of human locomotion, including stair climbing and sit-to-stand, to which we refer to as fully powered prostheses. Generally, these devices allow for improved versatility and movement robustness. For the sake of clarity, we define fully powered prosthetic devices as those relying solely on a single permanently engaged actuator that is designed to provide the torque and power required for supporting and sustaining typical active tasks, such as stair climbing. These prostheses are not designed to operate

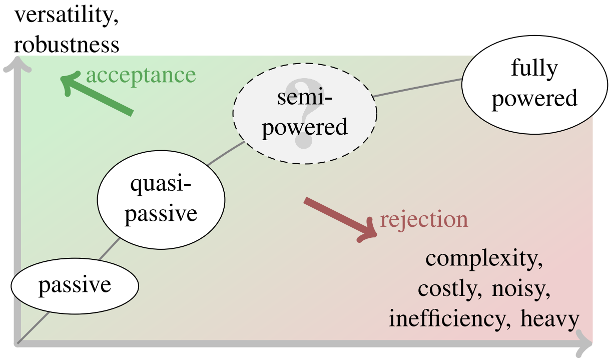

Fig. 1. Hypothetical qualitative representation of prosthetic knee acceptance by type. The figure has been adapted from the original work "Semi-active prostheses for low-power gait adaptation" from Adamczyk (2020) [11].

in conjunction with additional mechanisms that can store, modulate, or shape energy, such as variable transmissions, clutches, or springs. At most, elastic elements with negligible energy storage capability can be used for purposes of shock absorption or torque sensing.

Nevertheless, they are the least adopted prosthetic solution as passive and quasi-passive devices remain the conventional choice for transfemoral amputees [19]. One explanation is that – as of today – improved versatility and robustness entail a major trade-off in user needs. For instance, powered knees are generally much heavier, noisier, bulkier and have a shorter battery life than quasi-passive prosthetic solutions. Considering weight, an increase is negatively reflected in both biomechanical [20] and clinical results. Gait asymmetries [21], [22], excessive hip effort [23], socket instability and increased metabolic expenditure [24], [25], [26], [27], [28] are common consequences of heavy prosthetic devices.

Consequently, research is currently aiming to mitigate the deficiencies of fully powered prosthesis knees, partially sacrificing on device versatility. Although prosthetic devices falling in this category lack an agreed-upon scientific definition, they generally encompass both active and passive components. To fill this gap, our attempt is to group them into the definition of semi-powered, further divided in the two following clusters:

- *Partially powered* prostheses are typically designed to address a specific locomotion activity and/or phase. Therefore, they are generally unable to reproduce the majority of peak torques required to sustain active tasks.
- *Hybrid prostheses* feature an additional mechanism, either actively or passively controlled, to switch between operational modes, which could allow for an energetically more efficient or even passive behavior.

It is paramount to note that a given prosthetic device can belong to both clusters.

According to Sun et al. [29], a well-chosen actuation method can significantly enhance the walking gait of transfemoral amputees (TFA) and contribute to the prosthesis active capabilities. Speculating, semi-powered prostheses may provide the better trade-off in addressing most of the user needs, when compared to quasi-passive or powered solutions, as visualizable in Fig. 1. This figure is adopted from Adamczyk [11], but has been expanded to include the newly introduced category of semi-powered prostheses. It shows desired attributes on the vertical axis, and typical consequential negative attributes on the horizontal axis. However, it is rather complicated to classify user needs importance and translate such ranking to prosthetic design scenarios.

Therefore, the objective of this review is to understand which type of actuation solutions for knee prostheses provide the better trade-off between user needs fulfilment and device versatility. First, user needs are categorized and ranked in terms of importance. Then, an attempt on translating them into technological design requirements is presented. Next, a literature review of knee prostheses is performed and different actuation solutions are evaluated according to the user needs clustering and translation into design requirements. Lastly, existing and novel solutions of hybrid prostheses designs are discussed based on their actuation technologies.

Specifically, the work focuses on actuation technology in quasi-passive, semi-powered and powered prostheses. Instead, passive prostheses are excluded from this review study. Additionally, the different control methods are not directly discussed because the selection and sub-optimal implementation of controllers can significantly influence performance and introduce bias in evaluating the hardware's potential and capabilities. Moreover, numerous other studies have already extensively addressed control strategies [30], [31], [32], [33]. However, aspects that influence the controllability of the actuation solutions, such as backlash (hysteresis) and reduced bandwidth as a result of series elasticity, are considered.

II. Requirements

A. User Needs

An understanding of user needs for amputees is fundamental to define the technical requirements of a prosthesis device. This is the premise to any human-centered design process of innovative lower-limb prostheses, as proposed by studies in the last decade [34], [35], [36]. Designs that consider user needs could be substantially different from those based solely on engineering requirements. In support of this, Kooiman et al. [37] highlight a potential bias in literature, demonstrating that the majority of prosthesis design metrics only include kinetic and kinematic data.

Contrarily, a true human-centered design approach should embrace the complexity of the human and the human-machine system through a multidisciplinary approach. User research studies could reveal advantageous information through surveys and focus groups for defining and exploring user needs [35], [38], [39]. These methods often involve prosthetic users, experts, and all stakeholders in order to find unmet needs and open issues [40], [41]. Furthermore, it is suggested that user needs should be also included in the iterative processes of formative evaluation, driving the re-design of any device according to the international standards – as discussed about exoskeletons in [42].

Numerous investigations can be found on user needs for lower limb prostheses, as discussed by Manz et al. [43], which reviews 56 articles in which a need was explicitly reported by 8149 people with lower limb amputation. Four groups

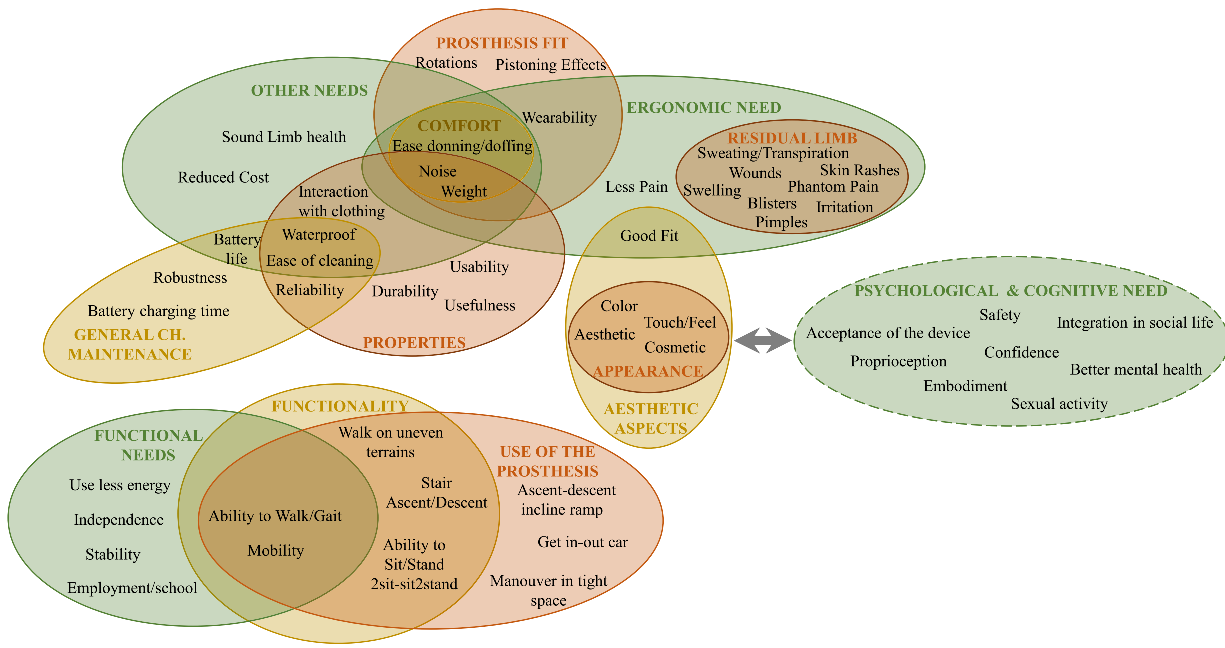

Fig. 2. Clusters and User needs extracted from the works of: Fanciullacci - yellow scale [36], Baars - red scale [45], Manz - green scale [43].

of needs were identified across these papers; namely, functional, psychological-cognitive, ergonomic and other. Certain desirable needs (e.g., the desire to experience less pain) could be shared by at least two groups. Overall, the subjects declared to desire an independent life with full re-integration in social activities. The prosthetic features that could be directly related to the aforementioned needs may be efficiency, versatility, stability, and facilitation of multiple activities (e.g., sitting and standing). These features would translate into perception of safety and increased confidence, alongside appearance and prosthesis embodiment. Different factors (e.g., mobility level [44], age, gender) also affected the user needs identification.

Furthermore, Fanciullacci et al. [36] presented a multidisciplinary study for the user-centered design of novel transfemoral prostheses. They developed a custom survey for unilateral transfemoral amputees, assessing motor abilities and autonomy in activities of daily living [46], [47], quality of life [48] and prosthesis satisfaction [48]. The survey showed how users prioritize reliability, comfort, low weight and cost. Looking at the functionality offered by prostheses, stability is a priority for TFAs, followed by adaptability to gait speed and different terrain conditions (uneven terrains [49], slopes, stairs). Active assistance is valuable in preventing falls and improving movement robustness and, in general, during the most demanding activities of daily living (ADLs), such as stair and slope ascent, sit-to-stand tasks and walking at high ambulation speeds.

Additionally, Baars et al. [45] presented a literature search about prostheses satisfaction in lower limb amputees population. In their work they highlight how patient satisfaction is strongly influenced by appearance, fit and the capabilities of the prosthesis [50], [51]. However, user satisfaction is subjective and influenced by psychological factors. Then, Barberi et al. [52] proposed a review study based on the

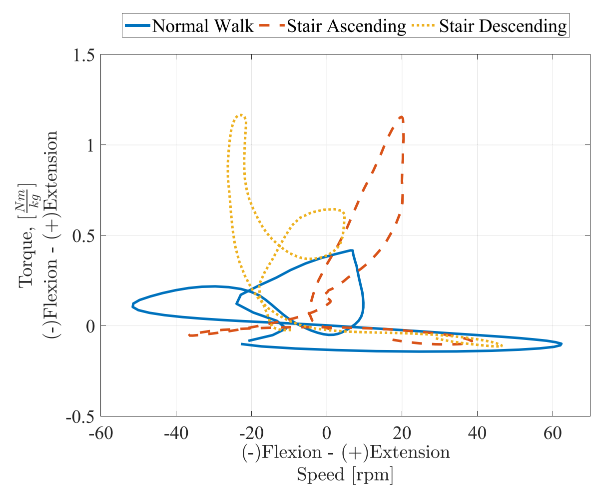

Fig. 3. Normalized torque speed requirements at walking, stair ascending, stair descending [59].

relationship between users' needs and lower limb prostheses. However, they propose a fully integrated neurally-controlled prosthesis as an optimal design to improve user experience, usability and functional restoration. Ideally, the works of Fanciullacci et al. [36], Manz et al. [43] and Baars et al. [45] can directly be combined to define a concatenated and quantifiable list of technological requirements for the purpose of prosthesis design. However, we found a lack of generalization of the many user needs and their clustering.

Fig. 2 presents clusters and user needs according to the references aforementioned, in different color scales. Although few needs are shared between the three and it is easy to cluster them (e.g., mobility), the majority are either connecting two out of three references or just one. Therefore, generalizing or prioritizing user needs becomes difficult as there is not a common vocabulary. Additionally, many user needs are not directly quantifiable for the purpose of defining technological design requirements. Therefore, only those that are quantifiable and influence design decisions are detailed and attempted to translate into technological requirements.

B. Technological Requirements: From User to Device

Among the range of user needs previously examined, identifying requirements related to versatility is crucial in determining the specifications for actuator sizing. Thus, biomechanical data from level ground walking [53], [54], [55], [56], [57], ascending and descending stairs [14], slopes [58] and task transitions [59] might help to define target specifications for actuators. Fig. 3 shows the normalized torque-speed profiles at the knee joint for different tasks such as walking, stair ascending and stair descending considering a 62.5 kg user (i.e., 50^{th} female percentile [60]).

Although walking is on average energetically passive, as it mainly evolves in the second and fourth quadrants [14], stair ascent requires a peak extension torque at slow speed greater than $1\,\mathrm{N}\cdot\mathrm{m/kg}$. During level-ground walking, the swing phase is typically characterized by ballistic motion, where the knee is primarily driven by hip movement, creating an inertial coupling with the prosthetic device. Consequently, active power at the joint is unnecessary [61], as quasi-passive prostheses can attain the desired swing-phase knee trajectory by leveraging the interplay of shank inertia, joint resistance, and thigh angular acceleration [5].

However, for certain activities (e.g., walking at slow speed and ascending stairs), the inertial coupling between the hip and knee is not sufficient to deliver the desired knee motion [62]. Therefore, active power can be exploited to either avoid compensatory mechanisms (i.e., pelvis tilt and hip hiking movements [63]) or avoid scuffing, falling or not clearing a step. Notably, during ballistic swing, inertial torques take precedence, steering the motion. Conversely, in non-ballistic swing, active knee torque assumes dominance, governing the motion dynamics [64]. Furthermore, different ways to climb a step can be considered. Generally, a prosthetic user either employs a step-to or a step-over approach. When equipping a quasi-passive knee, both of these approaches result in unnatural or limited knee motion during the swing phase, since the leg cannot be flexed actively. Nevertheless, individuals often need to perform compensatory manoeuvres to clear the step, such as hip circumduction, ankle vaulting, and/or pelvic tilting [65], when ambulating with energetically passive devices. As for the stairs stance phase, user, step and handrail combined form a closed kinematic chain. Therefore, upper body leaning, hip torque and handrail assistance might mitigate the lack of active power at the joint in conventional quasi-passive knees [62].

On top of that, to realize a natural ballistic movement, the prosthetic knee has to feature a low output impedance, by featuring low rotational inertia, damping and friction [66]. Considering the quasi-damping effect at peak knee speed during the swing phase, a biological minimum damping coefficient of roughly $0.02\,\mathrm{N}\cdot\mathrm{m}\cdot\mathrm{s}/(\mathrm{kg}\cdot\mathrm{rad})$ [67] is estimated at the knee joint. This value might be even lower in a prosthesis, due to a reduced prosthetic leg inertia. Inoue et al. [20], [68], [69] indicate that a prosthesis mass as low as 30 % of the physiological leg mass, while maintaining a similar center of mass position, significantly impacts knee and hip dynamics. Peak late-stance knee moment is increased by 43 %, whereas peak swing knee moment is decreased by 76 %. Additionally, peak stance hip power and work are reduced by 26 % and 22 %, respectively [69]. Moreover, changes in both prosthesis mass and center of mass location have a more significant impact on the knee moment than a change in center of mass alone [23]. This dual variation enhances functional safety and reduces the risk of injuries between the stump and socket [70]. In prosthetic designs where the actuation unit remains engaged during ballistic motion (i.e., excluding quasi-passive prostheses), the output impedance of the prosthesis is influenced by an additional factor: the actuator's reflected inertia at the joint, which is the product of the actuator's rotational inertia and the square of the transmission ratio. This term equally affects the system's dynamic behavior as the limbs' inertia. When designing an electrically actuated system that should be able to efficiently exert peak reference torques, one quickly finds that the reflected inertia is on the same order or even significantly larger than the limb's inertia, substantially disturbing its natural dynamics. In this context, minimizing reflected inertia is essential to allow for natural and passive swing motions. Whereas there is no defined threshold for an acceptable value for reflected inertia, we consider a value of 10 % of a biological shank or less based on the same value mentioned in [64]. Given that the rotational inertia of a biological shank and foot is approximately $3.8 \times 10^3\,\mathrm{kg}\cdot\mathrm{cm}^2$ at the knee joint, this suggests that the upper limit of the reflected inertia is roughly $4 \times 10^2\,\mathrm{kg}\cdot\mathrm{cm}^2$.

Table I collects the torque-speed requirements for each task together with their occurrence per day based on healthy subjects.

Level ground walking is often considered as the primary design objective for prosthetic devices. Whereas tasks governed by positive power such as sit-to-stand and stair ascent are requested activities by users [36], they are less frequent than level ground walking. Hence, Table I defines a technological baseline to compare the prostheses capabilities.

User needs other than those pertaining to versatility or ADLs result in additional technological requirements. In particular, an anthropomorphic-oriented design increases the psychological acceptance of the device [76]. Therefore, the shape of the prosthesis, its minimum height and size based on percentile length [60], and the distance between the knee center of rotation and the top of the pyramid establish the foundational ergonomic constraints of the device, as summarized in Table II. In fact, Lubis and Putri [77] presented three fundamental principles to incorporate anthropomorphic data into design methodologies. These considerations involve the design of devices to accommodate individuals at both extremes: those in the larger percentiles (90 % to 99 %) and those in the smaller percentiles (5 % to 10 %). This approach enables the fulfillment of individual patient needs while structuring the design around average sizes.

The strength, durability, and safety of prosthetic components are typically assessed using the standard ISO 10328, which

TABLE I
KNEE BIOMECHANICAL REQUIREMENTS FOR DIFFERENT TASKS: LEVEL WALKING, STAIR ASCENDING, STAIR DESCENDING AND SIT-TO-STAND [14], [71], [72], [73], [74], [75]

Data		***Level Walking***		***Stairs***				***Sit to stand***
				Ascending		*Descending*		
		Stance	*Swing*	*Stance*	*Swing*	*Stance*	*Swing*	
θ_{max}	[deg]	37.4	64.1	69.0	94.7	90.7	93.2	84.3
w_{max}	[rad/s]	4.78	5.59	5	6.37	4	5.6	1.9
$M_{ext,max}$ $M_{flx,max}$	$[N \cdot m/kg]$	0.49 0.26	0.19 0.22	1.1 0.22	0.04 0.18	1.35 0.0	0.05 0.09	0.94 0.22
P_{max} P_{min}	[W/kg]	0.52 -1.25	0.12 -0.9	2.58 -0.03	0.69 -0.36	0.24 -4.46	0.01 -0.49	1.76 -0.07
Damping Impedance	$[N \cdot m \cdot s/(kg \cdot rad)]$	-	0.005	-	-	0.55	-	-
Occurrence/day	[steps]	5500		47-66				33-71

TABLE II
RELEVANT DIMENSIONS FOR ANTHROPOMETRIC LEG IN FUNCTION OF PERCENTILE [80], [81]. *FROM KNEE CENTRE OF ROTATION TO ANKLE CENTRE OF ROTATION [60]

		1^{st}%	**50^{th}%**	**99^{th}%**
Shank length* [mm]	male	373	422	457
	female	338	384	417
Knee diameter [mm]	male	86	99	114
	female	76	89	104
Shank weight [kg]	male	2.12	3.65	5.17
	female	1.96	2.9	4.5

outlines requirements and testing methods for lower limb prostheses, with an emphasis on structural integrity and mechanical performance [78]. This standard provides a designated test loading level (P-code). ISO 60529, instead, relates to the degree of protection provided by enclosure in terms of dust and waterproofness (IP - code) [79].

Based on the translation of user needs (defined in Section II-A) into design requirements, a summary of the proposed mapping is presented in Table III.

III. SOLUTIONS

A. Search Strategy

The literature review of prosthetic legs is conducted considering IEEE Xplore, PubMed.gov, WEB OF SCIENCE, ASME and SCIENCE DIRECT databases, considering a time window of 30 years. To find hybrid prostheses, the first search term is either "semi-active", "semi-powered" OR "partially powered" OR "hybrid", AND logically connected with either "knee" OR "transfemoral" OR "lower-limb". Then, the last terms that are AND connected to the search string are "prosth*" AND "design", to broaden the search to all possible devices presented in the literature. Conversely, papers including terms like "orthotic", "exoskeleton", "transtibial", "below-knee", "paraplegics", or similar ones are discarded. After eliminating duplicates, our search across four databases yielded a total of 36 relevant conference and journal publications. Due to either the absence of prototype realization or a predominant focus on control strategies, only 15 of these publications were deemed suitable for inclusion in our review.

B. Type of Actuation Systems

Accounting for the biological requirements to be fulfilled by robotic knee prostheses, both active and passive mechatronic systems can be assessed.

1) Electro-Hydraulic Versus Electro-Mechanical Actuation: When evaluating actuation capabilities, the key distinction lies between electro-hydraulic and electro-mechanical actuators, as these are the most commonly used solutions. Electro-hydraulic systems are generally characterized by a high force output, at the cost of increased complexity and frequent maintenance due to potential fluid leaks. Their higher force density in comparison with mechanical systems makes them suitable in applications in which the overall size is an important constraint.

2) Linear Versus Rotary Actuation: In general, actuators can be furthermore categorized in linear and rotary. The former often require an additional gearing or linkage mechanism to convert linear into rotary motion at the knee joint, whereas the latter allows for direct gearing at the joint. Hydraulic technology for knee prostheses is usually linear, as it typically exploits hydraulic cylinders. This choice is generally related to a better fit within the biological shape of a lower leg. Nevertheless, applications of rotary hydraulic actuators exist (e.g., Ottobock 3R80).

On the other hand, electro-mechanical actuators are typically rotary, requiring gearing to obtain the desired torque-speed characteristics, as the available space at the knee joint is limited. As highlighted in Table III, the axial length from the pyramidal connection to the prosthesis rotation center should be ideally minimized (e.g., 26 mm for C-Leg). This justifies the preference for linear actuators, or for offsetting the rotary actuator from the knee joint, such as by employing a belt drive [82]. Furthermore, to achieve effective ballistic movement, having a distal center of mass may be beneficial, as it increases the prosthesis's rotational inertia. However, there is a counterargument that a more distal center of mass could make the leg feel heavier in certain situations.

3) Rigid Versus Elastic Actuation: Motion transmission between actuator and joint can either be rigid or elastic. A rigid coupling increases the control bandwidth and diminishes both actuation costs and size as it requires fewer bodies than elastic

TABLE III
OVERVIEW OF USER NEEDS AND TECHNOLOGICAL REQUIREMENTS THAT ADDRESS THESE NEEDS. USER NEEDS THAT WERE NOT FOUND TO HAVE A DIRECT OR QUANTIFIABLE RELATION WITH TECHNOLOGICAL REQUIREMENTS ARE LISTED IN THE BOTTOM ROW

User needs	**Technological requirements**
Relevant and quantifiable user needs for actuation technology design	
Ability to Walk/Gait	Braking torque $\sim 0.5\,\mathrm{N\cdot m/kg}$, low impedance $\sim 0.02\,\mathrm{N\cdot m\cdot s/(kg\cdot rad)}$, (table I), and low reflected inertia $< 4\times 10^2\,\mathrm{kg\cdot cm^2}$
Stability	High stance braking torque max $1.35\,\mathrm{N\cdot m/kg}$, locking mechanism, (table I)
Stair Ascent	Stance support torque $> 1\,\mathrm{N\cdot m/kg}$ (@2 rad/s), Swing support speed >5 rad/s, (table I)
Stair Descent	Braking torque $> 1.35\,\mathrm{N\cdot m/kg}$, (table I)
Stand2sit-sit2stand	Stance support torque $> 1\,\mathrm{N\cdot m/kg}$, RoM $>$ 95 deg (table I)
Other mobility (Get in-out car, Manoeuvre in tight space, Walk on uneven terrains)	Low impedance $\sim 0.02\,\mathrm{N\cdot m\cdot s/(kg\cdot rad)}$(table I)
Durability (Safety, Robustness, Reliability)	ISO standard 10328 [78] (structural testing of lower-limb prostheses), (e.g. C-Leg P5)
Size (Good Fit, Wearability, Appearance)	Anthropometric shape (table II), minimize pyramid-to-knee-joint length (e.g. C-Leg 26 mm)
Low perceived mass	Low absolute mass, low impedance $\sim 0.02\,\mathrm{N\cdot m\cdot s/(kg\cdot rad)}$, CoM location
Battery life (use less energy)	High efficiency of the actuator, capability of passive usability, low absolute mass
Reduced Cost	Less and affordable components (off-the-shelf/easy to manufacture)
Waterproof	ISO standard 60529 [79] (degree of protection provided by enclosure), (e.g. C-Leg IP67)
(Low) Noise	Motor disengagement or use of a small motor with efficient transmission
Irrelevant or unquantifiable user needs for actuation technology design	
Battery charging time, Interaction with clothing Independence, Ease donning-doffing, Acceptance of the device, Integration in social life, Ease of cleaning, Sexual activity, Embodiment, Proprioception, Employment/school, Color, Touch/Feel, Aesthetic, Usefulness, Cosmetic, Confidence, Better mental health, Sound Limb health, Comfort, Usability, Less Pain, Residual Limb Health	

transmissions. However, high control bandwidth implies high-frequency vibrations (e.g., teeth meshing or motor cogging torque vibration transferred to the joint). Additionally, a back-driving impact (such as the one resulting from heel strike) can generate a significant shock, leading to possible discomfort for the user and damage to the transmission. To address such issues, the integration of elastic elements, in particular in series with the actuator – referred to as a series elastic actuator (SEA) [83] – has gained momentum in the field of robotic transfemoral prostheses. Typically, the target is to design its stiffness to be sufficiently low to enable torque sensing and mitigate impacts, but high enough to not impede the desired actuation bandwidth. Note that the use of SEAs does not inherently classify a device as semi-powered or hybrid. When the elastic element is used solely for purposes such as shock mitigation or torque sensing, and does not enable significant power modulation, it does not alter the operational modes of the system (e.g., Power Knee [84]). Conversely, mechanisms such as clutches or variable transmissions enable operational flexibility, such as transitions between active and passive regimes or modulating torque-speed relationships, thereby qualifying the device as semi-powered.

A special type of SEA is a variable stiffness actuator (VSA), which contains at least one additional degree of actuation freedom that allows for significant modification of the series elastic stiffness. A temporal reduction of stiffness allows for power and speed modulations, potentially enabling the fulfillment of the different requirements between stance and swing phases. Comparative studies have demonstrated that a properly designed SEA can contribute to a significant reduction of mechanical work from the actuators during level-ground walking (reductions ranging from 14 % to 39 %) and running (reductions ranging from 37 % to 75 %) when compared to actuators rigidly connected to the knee joint [15]. However, the drawback is that such reductions entail substantial weight increases due to additional components. Furthermore, they can only be optimized for specific gait patterns, which typically results in reduced control bandwidth. Promising benefits of series elasticity include output smoothing, reduction of (reactive) shock propagation, and torque sensing, which could improve device robustness and usability. The main disadvantages of SEA are reduced control bandwidth and additional bodies, which make the mechanism heavier, bulkier and more expensive.

4) Damping and Braking Systems: Additionally, for power dissipation tasks in knee prostheses, dampers are the most adopted elements. Hydraulic dampers featuring motor-controlled valves and magnetorheological (MR) dampers are the most promising technology [85]. They share the same principle: they behave as a controllable impedance by varying their damping. Following hydraulic principles, the damping force is largely proportional to the squared flow rate and thus knee velocity, implying that hydraulic dampers can respond naturally to walking behavior [86]. In MR dampers, the application of an external magnetic field modifies the physical properties of its fluid, transitioning from liquid to semi-solid states [87], effectively changing the viscosity and thus damping behavior. Instead, the damping behavior of a hydraulic damper with control valves is controlled through mechanical closure of valve orifices with compact electric motors. In general, magnetorehological dampers offer faster response and a more straightforward method of adjusting the damping behavior, but come at a higher cost and do not allow for locking the joint (i.e., infinite damping). Mechanical control valves, on the other hand, provide high damping forces but may have slower response times and require more maintenance.

Although most of the devices available in the market today rely on adaptive technologies with controllable damping elements (e.g., Ottobock's C-Leg®), literature is exploring the possibility to use electric motors (EM) with suitable transmissions as positive power generators and active brakes or dampers [88]. This can potentially translate into improved system efficiency and extended battery life [89]. However, the power dissipation requirements during locomotion would render an electric motor that is large in size, arguably decreasing the device usability.

5) Transmission Ratio Considerations: The transmission ratio has to be defined depending on the design rationale of the actuator. Its definition comprises a trade-off between torque generation, efficiency, and dynamic behavior. Although high ratios allow for high torque and could improve efficiency by reducing Joule losses in the motor windings, they also reduce maximum speed and increase the reflected inertia which lower the leg's natural dynamics capabilities. Specifically, the inertia of the motor reflected at the knee joint is proportional to the square of the transmission ratio. Consequently, a higher transmission ratio results in poorer inertial backdrivability properties of the actuator [90]. Generally, prosthesis fluidity and user comfort can benefit from a highly back-drivable actuator. Moreover, as elucidated in Section II, the swing and stance phases entail distinct combination of torque-speed requirements. This discrepancy could be resolved by either concentrating on one of the two phases as the primary rationale for actuator design, implementing a variable transmission system, or devising a mechanism capable of disengaging the actuator at the knee joint (e.g., a magnetic clutch).

6) Specific Mechanical Transmission Systems: Lead-screws and ball-screws are two commonly used mechanisms to convert rotational into linear motion and vice versa [88]. Lead-screws generally offer cost-effectiveness, simplicity and self-lock capabilities. However, their low efficiency and limited accuracy make them less suitable for cyclical applications such as human locomotion. Conversely, ball-screws provide high efficiency, are accurate and characterized by reduced backlash, but their cost, acoustic noise emissions and maintenance requirements may represent limiting factors [91].

The most common rotary transmissions are represented by planetary gearboxes, belt transmissions and strain wave gears. Planetary gearboxes allow for reductions of up to roughly 1:7 per stage and benefit from compact construction and high torque density, allowing for efficient power transfer with uniform load distribution.

Belt transmissions offer advantages in placement by offsetting the motor from the joint and by requiring reduced mounting tolerances. The disadvantage is that they are not space-efficient, since they require a tightening wheel, and they are limited to low transmission ratios per stage. For these reasons, they are often combined with different types of transmission, like a planetary gearbox. In addition, belts can stretch and wear out over time, requiring periodic replacement.

Another type of rotary transmissions are strain wave gears, such as Harmonic Drives© [92], and Cycloid Drives© [93], [94]. They are both characterized by high transmission ratios, allowing for reductions in a single stage that would otherwise require three or more stages in a planetary gearbox or a belt drive, while retaining a small form factor, zero backlash, and a lightweight design. In fact, both strain wave gears and cycloid drives do not perform well for lower gear ratios, which typically make them less viable candidates for realizing transmission ratios that could also be achieved with two or fewer stages of planetary gears. Strain wave gear tooth engagement can provide smooth and backlash-free operation, contributing to significant positional accuracy compared to gearboxes that are affected by backlash when reversing motion. However, they are less efficient than a planetary gearbox, as they may experience efficiency loss due to sliding friction, leading to heat generation and decreased efficiency.

Finally, another family of transmission systems is traction-based drives, where force transmission is achieved through Coulomb friction by pre-stressing components, rather than through the shape of elements like gear teeth. Examples of such systems include cable-based capstan drives and roller-based Vectis© or Archimedes© drives [94]. Their rolling contact and lack of meshing of gear teeth can realise low noise, high efficiency and zero backlash, and the latter can – just like strain wage gears – realize higher transmission ratios in a single stage. However, the inherent pre-stressing of components typicially result in poorer torque density, especially for transfering peak loads. Instead, the drives would slip for higher loads, which in turn could be exploited as overtorque protection. Yet, the requirements for high precision and lack of industrial integration result in costly solutions and lack of knowledge about long-term reliability.

7) Electro-Hydrostatic Actuators: Other than passive dampers and electromechanical actuators, another types of actuators investigated are Electro-Hydrostatic Actuators (EHA). They consist of a hydraulic cylinder driven by an electric motor geared to an hydraulic pump. This solution might allow for a smoother operation and integration than using conventional gearing. However, they are prone to fluid leakage which further lowers efficiency with respect to electro-mechanical solutions [95]. A substantial differentiation with respect to rigid mechanical (e.g., toothed) transmissions is the non-rigid transmission inherent in EHAs, mainly due to fluid leakage paths inside the actuator. Generally, a non-rigid transmission might be perceived as less efficient than a rigid one. Nevertheless, it contributes to a natural impedance that yields smooth motion and greater adaptability to the user's perception.

C. Prosthetic Legs Evaluation Methodology

The literature review search identified 13 different knee prostheses, listed in Table IV. Commercial devices are also included in the table, as benchmark for comparison.

TABLE IV
LOWER LIMB PROSTHETIC KNEES

	Type	Name, Institute, Country	Year	Reference
1	K	C-Leg 4, Ottobock	2022	[96]
2	K	Power Knee II, Ossur	2012	[83]
3	K	ECT, Vanderbilt, USA	2022	[64]
4	K	CSEA, MIT, USA	2014	[97]
5	K	SA, Vanderbilt, USA	2018	[98]
6	K	SCSA, Vanderbilt, USA	2020	[99]
7	K	RIC, Chicago, USA	2015	[100]
8	K-KA	BerkeleyKnee, Berkeley, USA	2009	[7], [101]
9	K	AMKR, Minas Gerais, Brasil	2018	[102], [103]
10	K	MREC, Inha, Korea	2016	[87]
11	KA	CYBERLEGs, VUB, Belgium	2015	[104]
12	K	Dual Motor, IIT, Italy	2022	[105]
13	K	EHA actuator, IIT, Italy	2022	[95]
14	K	HybridUtah, Utah, USA	2018	[106]
15	KA	Bionic Leg, Utah, USA	2022	[107]

Among all, three consist of knee and ankle together, while the others consider just the knee. To make a fair comparison, only knee prostheses are considered.

1) Classification Criteria: To obtain an overview and to compare the prostheses identified with the technological requirements defined in Section II, Table V summarizes the actuation technologies, their characteristics, and their performance. 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;
- *Motion Transmission:* this column describes whether the actuator is directly coupled to the knee joint (Stiff) or with a series elastic actuator (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 five subcolumns concerning the hardware configuration of the actuation, the presence of a braking element – such as a hydraulic damper, 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 seven 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, cost and battery life.
- *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 given in Section I between hybrid and partially powered prosthetic knees;
- *Status:* this field illustrates at which stage the prosthesis is between commercially available (CA) or tested on a bench (TB);
- *Experiments:* this column presents, based on published data, whether the prosthetic knee has already been tested with an able-body adapter (AB) or amputees (A) or none of the above (N).

The presence of a question mark means that the information could not be obtained either from the respective publications or calculated by the author. Moreover, colours are used to elucidate performance approximations where green, yellow, orange and red correspond to good, acceptable, poor and bad, respectively. In reference to the technological criteria listed in Table III, addressing versatility involves considerations such as braking torque, active torque for stance and swing support, and the joint's high transparency. Since all devices excel in meeting requirements for braking and stance support (i.e., depicted as passive quadrants in Fig. 3), these standard features of quasi-passive prostheses have been excluded from the table for clarity. Instead, only the presence or absence of a braking element is presented.

2) Benchmarking Requirements: Additionally, because no prostheses can fulfil the torque requirements of the 99^{th} percentile of men, to get a consistent comparison and to define an upper limit for non-partially powered definitions, we consider the 50^{th} percentile of females as a benchmark to define prosthesis properties. Legs capable of exerting more than $68.75\,\mathrm{N \cdot m}$ deliver the required torque for tasks like sit-to-stand and stair stance extension. Consequently, they are not marked as partially powered prostheses. Devices are evaluated based on their ability to provide this torque, with two plus signs (green cell) for full support and one plus sign (yellow or orange) for partial support. A minus sign (red cell) indicates an inability to support a specific task. Furthermore, the ability to walk passively is positively assessed with a plus sign, while the need for motor assistance during the swing phase of level walking is evaluated negatively with a minus sign.

Lastly, a minus sign along with a red cell is used when the evaluated prosthesis is not able to address a specific task. Conversely, in evaluating passive tasks such as walking swing phase, the prosthesis performance is measured in terms of ballistic movement capability, as outlined in Section II-B. A plus sign and a green cell indicate that the prosthesis performs well. The same reasoning is applied to mass and height subcategories. The RoM and pyramid-to-knee joint distance are compared based on two commercially available prostheses considered as references. Their measure of the pyramid-to-knee joint distance has been considered to be the two extremes. For the RoM, Table I sets the maximum flexion angle as the minimum requirement, while the RoM of the Power Knee serves as the benchmark for optimal performance. Background colours are chosen according to the above definition. Then, noise, cost and battery life are marked with up and down arrows. Finally, reflected inertia during swing is considered to be acceptable when it is $4 \times 10^2\,\mathrm{kg \cdot cm^2}$ or lower, in line with explanations in Section II-B. A value at least four times lower or higher is considered good (green) and bad (red), respectively.

D. Technology Trends and Performance Evaluation

A detailed comparison of technological advancements of the prostheses in relation to the user needs is provided below.

TABLE V
OVERVIEW OF PROSTHESIS ACTUATION CHARACTERISTICS AND PERFORMANCE CRITERIA. COLOURS ARE USED TO ELUCIDATE PERFORMANCE APPROXIMATIONS BASED ON CRITERIA IN TABLE I, WHERE GREEN, YELLOW, ORANGE AND RED CORRESPOND TO GOOD, ACCEPTABLE, POOR AND BAD, RESPECTIVELY. UNKNOWN PROPERTIES ARE MARKED WITH "?" AND ARE COLOURED GREY. ABBREVIATIONS: K, KNEE; AK, ANKLE-KNEE; STIFF, STIFFNESS; SEA, SERIAL ELASTIC ACTUATOR; VSA, VARIABLE STIFFNESS ACTUATOR; T-S, TORQUE-SENSITIVE; SA, STANDALONE; T, TETHERED; D, DISENGAGED; CA, COMMERCIALLY AVAILABLE; TB, TESTBENCH; A, AMPUTEE; AB, ABLE-BODY ADAPTER; N, NO

				Actuator characteristics					*Prosthetic characteristics*							*Active Tasks*				*Semi-powered*			
Legs	Type	Motion Transmission	Power Supply	Motor and Transmission	Braking element	Power Rating [W]	Active Moment Rating [N · m]	Reflected Inertia at swing [kg · cm^2]	RoM [deg]	Mass [kg]	Height [mm]	pyramid-to-knee joint [mm]	Noiseness at walking	Cost	Battery consumption	Walking: Swing	Stairs: Swing Flexion	Stairs: Stance Extension	Sit to Stand	Hybrid	Partially Powered	Status	Experiments
C-Leg	K	Stiff	SA	(none)	✔	0	0	$< 10^{‡}$	0-130	1.24	289	26	$\vee$	$\vee$	$\vee$	+	−	−	−			CA	A
Power Knee	K	SEA	SA	DC motor, Harmonic Drive Gear		?	96	$> 1 \times 10^{3‡}$	0-120	3.1	275	45	$\wedge\wedge$	$\wedge\wedge$	$\wedge\wedge$	−	++	++	++			CA	A
ECT	K	Stiff	SA	BLDC Maxon EC-4pole 22, Lead screw & planetary gearbox, ECT		90	34	$25^{*†}$	-5:125	1.95	360	$30^{†}$	$\wedge$	$\wedge$	$\wedge$	+	++	+	+	✔	✔	TB	A
CSEA	K	SEA	SA	BLDC Maxon EC-4pole 30, parallel clutch, ball screw		200	120	$7.0 \times 10^{2†}$	0-120	2.7	285	$40^{†}$	$\wedge\wedge$	$\wedge$	$\wedge\wedge$	−	++	++	++	✔		TB	A
SA	K	Stiff	SA	BLDC Maxon EC-8pole 45, leadscrew		70	7	$46^{*†}$	0-105	1.07	?	?	$\wedge$	$\vee$	$\wedge$	+	++	−	−		✔	TB	A
SCSA	K	Stiff	SA	BLDC Maxon EC-4pole 22, Spur gear & Lead screw	✔	90	7.5	$31^{*†}$	0-130	2.18	278	$25^{†}$	$\vee$	$\wedge$	$\vee$	+	++	−	−	✔	✔	TB	A
RIC	K	Stiff	T	BLDC Maxon EC-4pole 22, Ball screw, motorized clutch	✔	90	85	D	?	1.7	?	?	$\vee$	$\wedge$	$\wedge$	+	++	++	++	✔		TB	AB
BerkeleyKnee	AK	Stiff	SA	BLDC Maxon EC-powermax 30, hydraulic pump		40	?	$> 1 \times 10^{2‡}$	?	4	280	$40^{†}$	$\wedge$	$\wedge$	$\wedge$	+	++	?	?		✔	TB	A
AMKR	K	Stiff	SA	BLDC Maxon EC-7pole 60, harmonic drive, MR clutch	✔	100	40.4	D	?	2	117	$45^{†}$	$\wedge$	$\wedge\wedge$	$\wedge$	+	++	+	+	✔	✔	TB	A
MREC	K	Stiff	T	BLDC Maxon EC-11pole 90, planetary gearhead	✔	260	43.4	$9.4 \times 10^{3†}$	0-155	5	313	$45^{†}$	$\wedge$	$\wedge$	$\wedge$	−	++	+	+	✔	✔	TB	N
Cyberlegs	AK	VSA	T	Motor, ball screw	✔	?	75	$> 1 \times 10^{2‡}$	0-90	?	?	?	$\wedge\wedge$	$\wedge$	$\wedge$	+	++	++	++	✔		TB	N
Dual Motor	K	Stiff	T	BLDC Maxon EC-8pole 45 & Koll-Morgen BMS 1712, 2 strain gears		320	75	$1.8 \times 10^{3†}$	0-125	3	257	$55^{†}$	$\wedge\wedge$	$\wedge\wedge$	$\wedge\wedge$	−	++	++	++	✔		TB	AB
EHA actuator	K	Stiff	SA	Custom motor, gerotor pump		220	35	$> 1 \times 10^{2‡}$	0-110	2.4	315	$50^{†}$	$\wedge$	$\wedge$	$\wedge$	+	++	+	+		✔	TB	N
Hybrid Utah	K	Stiff	SA	BLDC Maxon EC-4pole 22, roller & timing belt, AVT	✔	120	125	D	0-155	1.7	290	$55^{†}$	$\wedge\wedge$	$\wedge$	$\wedge$	+	++	++	++	✔		TB	A
Bionic Leg	AK	T-S	SA	BLDC Maxon EC-4pole 22, Helical gears & ball screw, variable transmission		120	125	$3.6 \times 10^{2*†}$	0-120	1.2	255	23	$\wedge\wedge$	$\wedge$	$\wedge$	+	++	++	++	✔		TB	A

* computed at the highest transmission ratio at swing † estimated based on simplified calculations from available graphics/data ‡ conservative estimated minimum/maximum based on used technology and lack of data

1) Actuation Technology and Power Ratings: The majority of the devices are characterized by a stiff actuation technology, while just three feature elastic elements. Out of these, only one is an SEA, while the other comprise a VSA or a torque-sensitive joint (T-S), in which a spring can 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. This trend reflects the increasing focus on developing devices with high Technology Readiness Levels (TRLs) instead of platforms for investigating control strategies or actuation solutions.

The majority of the motor choices are Maxon BLDC motors that span power ratings between 70 W to 260 W. Only one design considers a custom motor. Five prostheses include braking elements in their layout. Depending on the transmission choice, power ratings vary between 70 W to 360 W while knee active moment ratings are widely spread between 7 N · m to 125 N · m. Four devices are able to provide active peak moments of less than 35 N · m, two between 35 N · m and 75 N · m, and six legs are capable of providing more than 75 N · m. Regarding transmission choices, four devices employ ball screws, while three utilize lead screws. Other types of transmissions, such as harmonic drives, planetary gearboxes, and hydraulic pumps, are each used twice among all the remaining devices. Eight legs have a mechanism capable of changing the actuation configuration with either clutches, springs or actively variable transmission (AVT).

2) Hybrid Versus Partially Powered Designs: As per the definitions outlined in the manuscript for "hybrid" and "partially powered" systems, this distinction sheds light on the designer's intent. In particular, both approaches can achieve desired reduction of reflected inertia and friction in order to realise a smooth and efficient swing phase, either by employing a disengagement mechanism or an under-powered motor, respectively.

Six 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 *CSEA* leverages a parallel clutch to alternate between a mode that employes an SEA and a passive spring, significantly reducing electrical energy consumption during walking compared to traditional SEAs. The *Bionic Leg*, although nearing the upper bound defined in Section II with a maximum swing reflected inertia of 356 kg · cm^2, can still walk passively. On the other hand, the *Dual Motor* design requires motor engagement for walking, impacting battery life.

Three legs are partially powered, with the only aim to actively power the swing phase during stair climbing. Devices such as SA and SCSA maintain a reflected inertia around 10 % of the upper bound, ensuring the possibility of passive walking.

Three prostheses belong to both the definitions of hybrid and partially powered designs. *AMKR* provides active torque

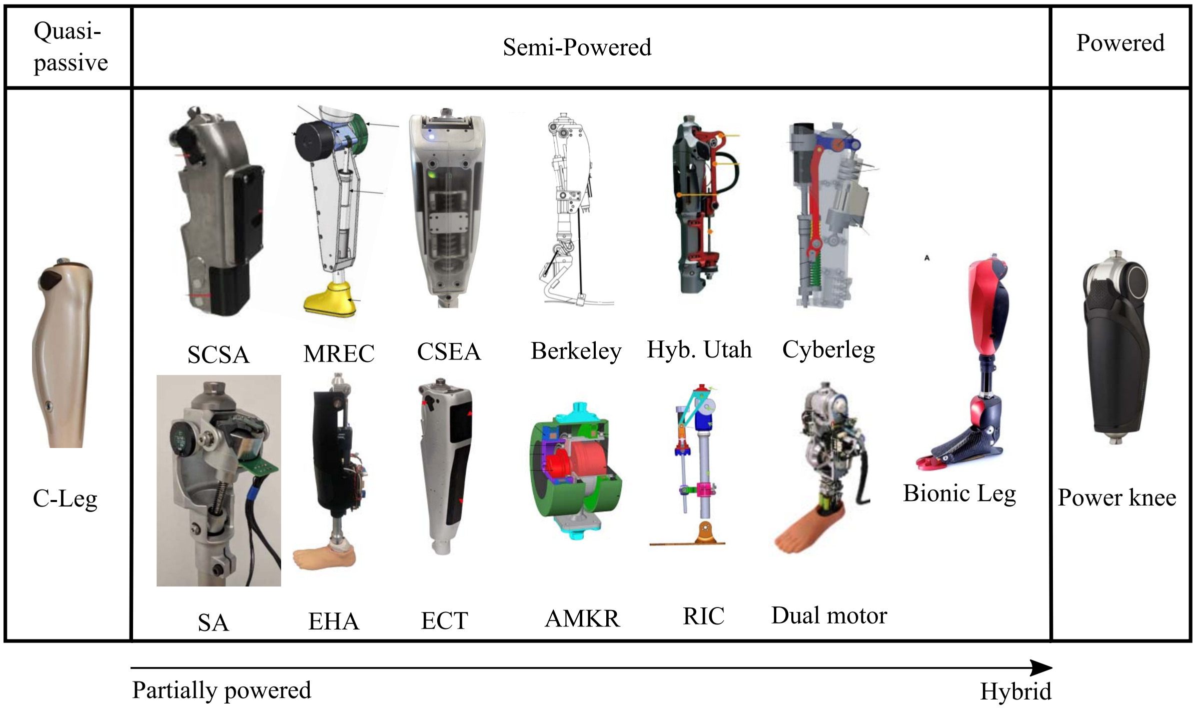

Fig. 4. Lower limb prosthetic knees and their clusters [7], [64], [84], [87], [95], [96], [97], [98], [99], [100], [101], [102], [103], [104], [105], [106], [107].

to support also energetically demanding tasks while it allows motor disengagement to walk passively. *ECT* focuses on swing support while the motor is always engaged, without sacrificing the ability to offer ballistic swing-phase behaviour. Notably, while low output impedance (high joint transparency) is a crucial requirement, only the *ECT* knee [108] characterized leg impedance with a value of $0.5\,\mathrm{N\cdot m\cdot s/rad}$. The *MREC* leg focusing on partially support for both stairs and sit-to-stand. However, it cannot walk passively as a result of its high reflected inertia.

3) Additional User Needs: Additional user requirements for ensuring user satisfaction include possessing an anthropomorphic shape with a perceived weight that is low. For the comparison, the three levels of female percentile are used as boundaries in Table II. Hence, five systems can fulfil the 1^{st} female percentile, four between the 1^{st} and the 50^{th}, two between 50^{th} and the 99^{th}, and only one supports more than the 99^{th} female percentile weight. Comparing the shank height of the 1^{st} female percentile to the devices built height, all the prostheses are smaller except for the *ECT* device (360 mm). Furthermore, the vertical distance between pyramidal connection and knee joint center relates to the possibility to fit the device to a large population. This length is directly influenced by the design choice and the type of actuation as rotary actuators geared to the knee joint shift such length by at least half of the motor radius. Even if it is a relevant parameter, only the *Bionic Leg* claims that its built height is lower than the C-Leg one (26 mm). For all the other systems the pyramid-to-knee joint distance is graphically estimated. Three devices are longer than the Power Knee (45 mm), while the others fall between the two commercial devices.

Noise, cost and battery life are important attributes according to the assessment in Section II-A, yet noticeably very few quantitative information is available in literature regarding these measures. As such, we have attempted to assess them qualitatively based on the used technologies. Low noise is of particular importance in repetitive tasks such as walking (Table I). Only the *EHA* prosthesis provides some reference considering experiments in an anechoic chamber with a peak sound pressure level of around 60 dBA [109]. For all the other devices, the authors decided to assign qualitative scores for noise based on the passivity of walking, i.e., on whether the actuator is disengaged, back-driven or required to be actively used as a brake due to the high inertia and friction of the transmission type and size. Cost was assessed similarly. By considering the type and number of (active) components, and the need for frequent maintenance, a qualitative estimate was made for device cost. Finally, battery life (i.e., low battery consumption) was evaluated using the same parameters as noise, considering the efficiency of the transmission and whether the device has a low absolute mass (Table III).

None of the prostheses are commercially available. However, most of the groups are clinically evaluating the devices. Subjects tests have been performed for 10 prostheses, eight of them including transfemoral amputees population. Finally, Fig. 4 displays the prostheses being investigated and indicates the specific clusters to which they belong.

IV. Discussion & Conclusion

Lower limb amputation has a profound effect on individuals, causing a decline in mobility and introducing challenges that negatively influence activities of daily living. In the presented work, prosthetic characteristics were discussed based on a human-centered perspective. Although user needs are various and generally subjective, many of them are not directly quantifiable and difficult to translate into technological design requirements. Additionally, challenges in generalizing and clustering them emerge, especially regarding device versatility. Consequentially, the review highlighted that prosthetic leg designs mainly prioritize biomechanical features (i.e., Table I), resulting in a potential engineering bias. In fact, the translation of user needs into technological requirements is typically expressed based on knee biomechanical requirements (Table I), anthropometric dimension and weight (Table II), as summarized in Table III.

In Section III, prosthetic knees that belong to the semi-powered definition given in Section I were identified and compared to commercially available devices. Prosthetic legs and their actuation technologies were clustered and compared in Table V with an insight into the advantages and disadvantages related to each actuation typology. Actuation systems with ball screws have emerged as the predominant choice of motion transmission, underlining their promising advantages in prosthesis design. Considering partially powered solutions, the combination of compact motors and suitable transmission systems seems to provide the better trade-off between device usability, prosthetic embodiment and prosthesis robustness and versatility. Conversely, variable transmission systems seem the promising design for hybrid solutions, either by reducing the actuation reflected inertia (i.e., increase the user perception of natural movement) or completely disengage the actuation unit from the knee joint.

Moreover, it emerged that the actuation sizing requires a thorough analysis of joint impedance for fully or partially powered solutions, as the biomechanical requirements are wide and complementary during load-support and swing phases.

Noticeably, none of the prostheses have employed traction-based drives, despite their promise regarding low noise, efficiency and controllability. Possibly these drives have not yet gained momentum in prosthesis designs due to the lack of proven reliability, higher price or poor torque/volumetric density. Nevertheless, their lower peak torque could still be promising in combination with overtorque protection (slip) for hybrid or partially powered design solutions.

The outcomes of the presented review suggest that semi-powered prostheses may have the potential for widespread user acceptance thanks to their ability to combine positive aspects of passive and active solutions, as proposed in Fig. 1. Other relevant outcomes of the presented study include a limited exploration of noise emissions, output impedance, and ISO compliance in prosthesis design. These aspects may hinder the usability and acceptance of the device. Whereas ISO compliance is crucial for increasing the device technology readiness level (TRL), it may not be relevant in research endeavors. Finally, despite the review highlighted noise reduction and cost-effectiveness as relevant prosthetic features, they are often overlooked.

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Andrea Berettoni (Member, IEEE) was born in Gubbio, Italy, in 1996. He received the B.S. degree in mechanical engineering from the University of Perugia, Italy, in 2019, and the M.S. degree from the Polytechnic University of Turin, Italy, in 2021. He is currently pursuing the Ph.D. degree in mechanical engineering with Polytechnic University of Turin and the Rehab Technologies INAIL-IIT Laboratory, Italian Institute of Technology.

In 2024, he was a Visiting Researcher with the Neurobionics Lab, University of Michigan, under the supervision of Dr. Elliott J. Rouse. He has teaching experience in applied mechanics and machines, applied mechanics to biomedical systems courses. His research interests include design and control of robotic and wearable devices, with a particular focus on bionic limbs.

Josephus J. M. Driessen (Member, IEEE) was born in Delft, The Netherlands, in 1991. He received the B.S. and M.S. degrees in mechanical engineering from the Delft University of Technology, Delft, in 2012 and 2015, respectively, and the Ph.D. degree in bioengineering and robotics from the University of Genoa, Genoa, Italy, in 2019.

From 2016 to 2019, he was a Doctoral Fellow with the Department of Advanced Robotics, Italian Institute of Technology (IIT), and from 2019 to 2023, he was a Postdoctoral Researcher with the Rehab Technologies INAIL-IIT Lab, IIT. Since 2024, he is a Senior Mechanical Engineer with the R&D Department of ANYbotics, Zürich, Switzerland. His research interests include mechanism, electromechanical and controller co-design, especially in the fields of dynamic legged and rehabilitation robotics.

Marco Puliti (Student Member, IEEE) was born in Assisi, Italy, in 1996. He received the B.S. degree in mechanical engineering from Università degli studi di Perugia in 2018 and the M.S. degree in mechatronic engineering from Politecnico di Torino in 2020.

Since 2020, he has been a Doctoral Researcher in mechanical engineering with the Rehab Technologies INAIL-IIT Lab, Italian Institute of Technology and the Mechatronics Laboratory, Politecnico di Torino. In 2022, he was a Visiting Researcher with the Center for Rehabilitation Engineering and Assistive Technology, Vanderbilt University, under the supervision of Dr. Michael Goldfarb. His research interests include control and design of mechatronic systems and fluid dynamics, especially in the fields of rehabilitation and assistive robotics.

Giacinto Barresi (Member, IEEE) received the M.Sc. degree in experimental psychology and cognitive-behavioural neuroscience from the University of Padua, Italy, in 2010, and the Ph.D. degree in robotics, cognition and interaction technologies from the University of Genoa, Italy, in 2015. He is a Professor of robotics with the Bristol Robotics Laboratory, University of the West of England, Bristol, U.K. Previously, he was a Researcher in human-centred biomedical robotics with the Rehab Technology Lab and the Advanced Robotics Department of Istituto Italiano di Tecnologia (IIT), Genoa, Italy. He is active as a Specialist (Teacher and Consultant) in cognitive ergonomics and neuroergonomics since 2006. From 2012 to 2019, he was a member of the Advanced Robotics (ADVR) Department of IIT. He was invited as a Guest Lecturer with Kyushu University, Fukuoka, Japan, in 2023, and as a Scholar-in-Residence with the Indian Institute of Technology, Gandhinagar, India, in 2024. He currently is a Visiting Fellow with the Knowledge Media Institute, Open University, Milton Keynes, U.K. He contributed to the European project μRALP, on surgical robotics, and (as co-coordinator) to TEEP-SLA, a project funded by Fondazione Roma to devise assistive technologies for people with Amyotrophic Lateral Sclerosis. In Rehab Technologies Lab from 2019 to 2024, he contributed to projects on prosthetics and wearable robotics, especially in collaboration with Centro Protesi INAIL. In 2021 he secured a grant from the Italian Multiple Sclerosis Foundation (FISM) for coordinating ENACT, a project on rehabilitative and assistive systems (with special attention to exergames). He also contributed to the RAISE ecosystem for innovation, supported by the National Recovery and Resilience Program, Italy. He also is an Associate Editor of *Sage International Journal of Advanced Robotic Systems*, the IEEE TRANSACTIONS IN HUMAN–MACHINE SYSTEMS, and *Frontiers in Neuroergonomics*, and a member of the Editorial Board of *Wiley International Journal of Medical Robotics and Computer-Assisted Surgery*.

Carlo De Benedictis was born in Italy, in 1991. He received the B.S. degree in biomedical engineering from University of Naples "Federico II", Italy, in 2013, the M.S. degree in biomedical engineering and the Ph.D. degree in mechanical engineering from the Polytechnic University of Turin, Italy, in 2015 and 2020, respectively.

From 2019 to 2021, he was a Research Fellow with the Department of Mechanical and Aerospace Engineering, Polytechnic University of Turin, where he has been a Researcher (Assistant Professor with time contract) since 2021. He has teaching experience in applied mechanics and machines, applied mechanics to biomedical systems and fluid automation courses. He has been tutor and co-tutor for more than 20 master thesis students in biomedical and mechanical engineering. He is author of more than 30 Scopus-indexed articles, and 2 inventions. His research interests include orthotics and prosthetics design, human motion and posture analysis, passive exoskeletons, and haptic systems.

Dr. De Benedictis is currently member of the Technical Committee for Biomechanical Engineering of the International Federation for the Promotion of Mechanisms and Machine Science, the European Society of Biomechanics, and the Italian Society of Movement Analysis in Clinic.

Carlo Ferraresi was born in Pavia, Italy, in 1954. He received the M.S. degree in mechanical engineering from the Polytechnic of Turin, Italy, in 1980. Since 1983, he has been an University Researcher with the Department of Mechanics, currently the Department of Mechanical and Aerospace Engineering, Polytechnic of Turin, where an Associate Professor since 1992 and a Full Professor of Applied Mechanics since March 2000. He has been a Regular Teacher of basic engineering mechanics, mechanics applied to machines, robot mechanics, control of mechanical systems, and mechanics of biomedical systems. He is the author of several textbooks on applied mechanics, fluid automation and the control of mechanical systems. He is the author of over 290 scientific articles in national and international journals and conference proceedings, and is also the author of 18 industrial patents. His main research topics are: automation, robotics, mechatronics, fluid transmissions, and biomedical engineering. His main academic and scientific commitments the Deputy Chair of the University Diploma Course in Mechanical Engineering from 1999 to 2001, a Chair of the Ph.D. course in Applied Mechanics from 1999 to 2012, the Vice Dean of the 1st Faculty of Engineering from 2003 to 2012, the Technical Committee for Biomechanical Engineering of IFToMM (International Federation for the promotion of Mechanism and Machine Science) from 2016 to 2023, and the International Scientific Committee of Robotics in Alpe-Adria-Danube Region from 2019 to 2022, and the Vice Director of the Department of Mechanical and Aerospace Engineering DIMEAS from 2020 to 2023.

Matteo Laffranchi received the master's degree in mechatronic engineering from the Polytechnic University of Turin in 2006 and the Ph.D. degree in robotics from the University of Sheffield, U.K., in 2011.

He is a Coordinator of robotics with the Rehab Technologies Lab, Italian Institute of Technology (IIT). His main works focus on the development of novel mechatronic systems for robotic applications. Following a brief experience in the automation industry at OSAI A.S., from 2008 to 2011. He has been a Research Fellow and the Postdoctoral Researcher with the Department of Advanced Robotics (ADVR), Italian Institute of Technology from 2011 to 2014. At ADVR, he spent 6 years developing compliant actuation systems for robotics and robots (CompAct actuators, CompAct arm) for safe physical human–robot interaction. Since 2014, he works at Rehab Technologies, IIT, with specific focus on the development of novel healthcare robots, particularly robotic prostheses and exoskeletons (Hannes, Pro-Leg, Twin, and Float). He is currently manages the activities related to robotics research and product development within the lab starting from 2016. Along with research and development, he is also involved in technology transfer activities and in entrepreneurial projects to bring the developed technology to the medical and industrial robotics industries since 2012.