Towards Accessible Mobility Support: User-Centered Design of a Passive, Multi-Functional, Low-Cost Knee Exoskeleton

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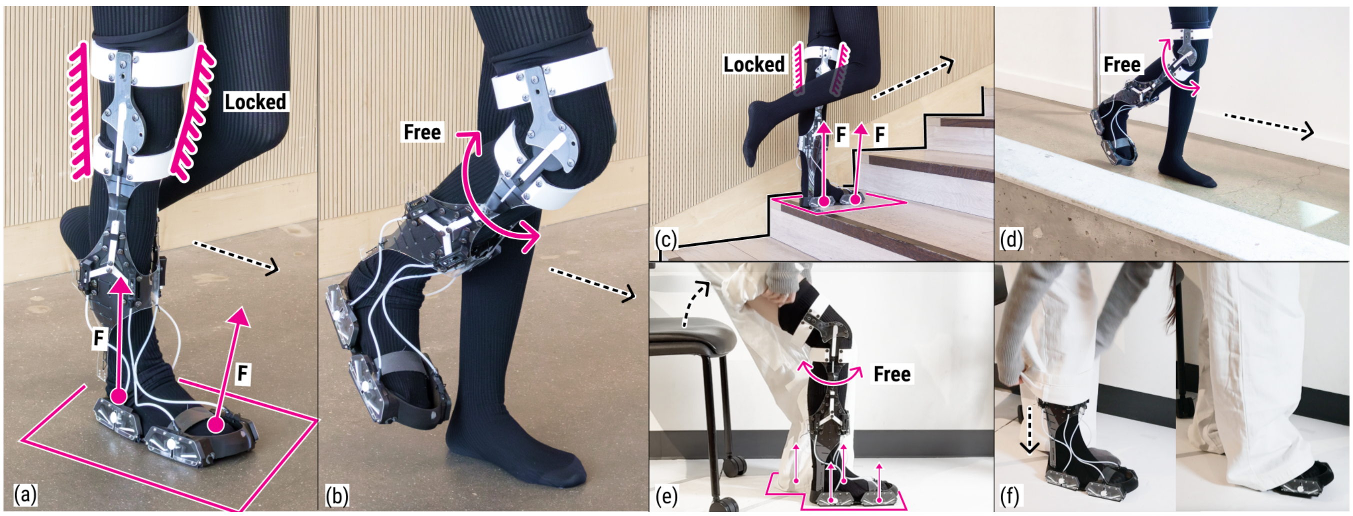

Figure 1: We present a design of an unpowered knee exoskeleton that (a) stabilizes the knee when the foot is in full contact with the ground and (b) allows free movement during leg swinging. Our design supports adaptation to the walking gait cycle on flat surfaces, (c) stairs, (d) hills. Additionally, our design allows (e) sit-stand transitions and (f) wearing under clothing.

Abstract

Walking aids are critical for people with mobility impairments, yet current options remain unsatisfactory. Static knee braces are lightweight and affordable, but their rigid joints force users into unnatural gait patterns, leading to fatigue, reduced safety, and high abandonment rates. Robotic exoskeletons, in contrast, offer dynamic assistance that adapts to gait phases but rely on sensors, motors, and batteries that make them heavy, complex, and prohibitively expensive.

In this paper, we propose a fully passive knee exoskeleton design that combines the accessibility of static braces with the adaptive functionality of robotic systems. Our design employs a mechanical trigger under the foot to lock and release the knee joint in sync with the gait cycle, enabling more natural walking without electronics or actuation. Using human-centered methods, we conducted interviews with clinicians and orthosis users to guide our design and evaluated an early prototype as a design probe with stakeholders.

CCS Concepts

• **Human-centered computing** → **Human computer interaction (HCI)**.

Keywords

Exoskeleton, Rehabilitation, Wearable Device

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1 INTRODUCTION

In HCI, we are concerned with how technologies can be designed to meaningfully support people in their everyday lives. A hallmark of our work is bridging technical innovation with a deep understanding of the target users. This spans a wide range of work: from novel haptic controllers that evoke specific sensations and experiences [38], to custom-fabricated furniture that integrates digital fabrication for hobbyists [48], to wearable sensing approaches that track physical activity and fitness [50].

While HCI is often associated with exploratory technologies, it also has a strong tradition of addressing concrete challenges for specific user groups. Rehabilitation and assistive technology are one such area. Here, researchers design new tools, processes, and devices for patients, caregivers, and clinicians: mindfulness-based tangible devices for stroke rehabilitation at home [47], standardized assessment tools that help therapists measure recovery progress [35], co-designed prosthetics tailored to individual patients [22], customizable mobility aids [11], new wearable devices such as knitted sleeves for hand compression [37], socks with integrated sensors for prosthetic feedback [40], or thermoformed orthoses with custom sensor integration [71].

With over 28% of adults in the U.S. living with some form of disability, and 12% reporting mobility limitations such as difficulty walking or climbing stairs [13], the development of new assistive and rehabilitative technologies is both timely and necessary. Beyond permanent aids like wheelchairs or canes, lower-limb orthoses and exoskeletons play a key role in supporting gait rehabilitation. These devices help patients recover from injuries such as strokes or spinal cord damage and manage age-related muscle loss [3, 20].

Orthoses for walking are typically static or robotic. The most widely available and affordable knee orthoses are static braces (Figure 5a). Their joints are preset to allow or block a fixed range of motion, which means they cannot adapt to the body's dynamic movement. As a result, users must modify their natural walking patterns—for example, swinging the entire leg outward in a "hip hike" to clear the braced limb—leading to unnatural, tiring, and sometimes unsafe movement. This lack of adaptability contributes to low adoption: studies report that 58–79% of static knee–ankle–foot orthoses are abandoned [29].

In contrast, robotic exoskeletons have been developed that provide adaptive assistance, changing their support in real time based on the user's gait. For instance, a robotic brace might unlock the knee during leg swing but lock it during midstance when the leg bears weight [77]. Other systems detect gait phases using underfoot pressure sensors or knee rotation signals and then apply adaptive support [1, 29]. These robotic devices better preserve natural walking patterns and increase safety, including on uneven terrain such as slopes. However, they rely on active sensing and actuation, making them bulky, heavy, expensive, and often impractical for everyday use.

In this paper, we introduce a new exoskeleton design that offers a middle ground: a passive brace that achieves some of the adaptability of robotic devices without requiring electronics or motors. Our design locks and releases knee flexion based on the gait cycle, triggered by foot load, but it does so entirely through passive mechanics. This means it retains the advantages of static braces—lightweight, discreet, wearable under clothing, no batteries to charge, and low cost—while providing adaptive support previously only seen in robotic systems. Our design process is grounded in user-centered HCI methods for understanding stakeholders and shaping technical requirements. To ensure that our design aligns with real needs, we conducted formative need-finding interviews with medical experts and orthosis users that shape our requirements. Building on fabrication techniques and unpowered, body-triggered interactive mechanisms in HCI, we developed a dynamic knee brace that uses a fully passive system to engage during stance and disengage during swing. We evaluate this prototype as a design probe with users and experts, gathering qualitative insights into its effectiveness and implications for everyday rehabilitation support.

1.1 Walkthrough

Walking consists of two alternating phases: a support phase, when the foot is in contact with the ground and bears body weight, and a swing phase, during which the foot lifts to propel the body forward. Our exoskeleton, shown in Figure 1, explicitly leverages this rhythm as its interaction mechanism by (a) locking the knee during the support phase to provide stability, and (b) automatically unlocking it during the swing phase to allow natural leg movement.

When the foot transitions from heel strike to full-foot contact, the device detects ground reaction forces and engages the lock at the knee joint, stabilizing the knee while the leg supports the body. As the gait progresses into the swing phase, the load is removed, and the joint unlocks, enabling free flexion so the leg can swing forward naturally. Because the switching mechanism is load-dependent rather than based on displacements, it supports movements beyond level walking, including (c) climbing stairs, (d) sitting down and standing up, and (e) walking on slopes. (f) The low-profile mechanism can also be worn under clothing, which is an important factor for patient comfort and acceptance.

1.2 Contributions

The main contribution of this work is a novel, user-centered design of a fully passive, low-cost knee exoskeleton that effectively combines the accessibility of static braces with the adaptive functionality of robotic systems. This paper makes the following specific contributions:

(1) **User-centered design requirements.** We conducted formative interviews with medical experts and orthosis users to identify needs and constraints for daily knee exoskeleton use.
(2) **A novel passive knee exoskeleton design** that adapts to the gait cycle using only mechanical triggers. We introduce an unpowered, load-triggered locking mechanism that engages during stance and releases during swing, enabling adaptive knee support without electronics or motors.
(3) **Technical validation.** We evaluate the mechanism's load-dependent triggers, timing accuracy, and strength, demonstrating how it synchronizes with the wearer's gait.
(4) **Qualitative evaluation.** Through stakeholder interviews, we assess wearability, comfort, and willingness to adopt the device in daily life. This validates the new direction for

accessible, everyday exoskeletons that bridge static braces and robotic systems.

(5) **Design considerations.** From our evaluations, we derive key considerations that provide actionable guidance for future passive knee exoskeletons. These highlight how such devices can be designed to be more functional, accessible, and comfortable for gait rehabilitation.

The scope of this paper is to address core challenges in knee assistance and to highlight key design considerations for future development. As such, the presented device should be understood as a research prototype to assess the design of this novel exoskeleton, rather than as a safety-tested, product-ready device. This paper lays the groundwork by inventing a novel brace design and performing the necessary initial evaluation—both technical (quantitative) and with stakeholders (qualitative). These contributions, in turn, open many opportunities for future research, including personalization, computational design tools, patient studies on trust or social acceptability, and many more, which are outside of the scope of this current paper.

2 RELATED WORK

In this section, we review prior work in participatory design for assistive and rehabilitative devices, unpowered body-triggered interactions in fabrication, and lower-limb exoskeletons.

2.1 Participatory Design for Assistive Mobility and Rehabilitation Technologies

Participatory design approaches [63] are widely applied in HCI research [56]. In healthcare and assistive technology, researchers commonly use interviews [12, 84], co-design sessions [11, 54, 64], and prototype probes [25, 73] to understand the unique contexts of specific user groups, learning from the knowledge and practice of professionals which are essential for shaping design outcomes, like assistive robots for the elderly [25, 64], information displays for people with sensory or motor impairments [54, 73], or clinical decision-support systems [84]. Here, we focus on the two most related topics to our work: mobility assistive technologies and rehabilitation devices.

Research on *mobility assistive technologies* aims to enhance daily mobility. User-centered design methods have been used to identify factors that affect long-term acceptance of these devices [9, 21], including social stigma [11, 24, 53, 57], limited usage scenarios [9, 11], limited customization [11], and changes in user needs over time [11, 21]. However, this body of work has largely concentrated on non-wearable aids, such as wheelchairs [12, 44], walkers, and canes [11], while wearable systems such as exoskeletons and orthotic braces are missing.

For research on *rehabilitation devices*, clinicians are often consulted to assess how systems support strength recovery. Prior HCI work has examined areas such as upper-limb rehabilitation exercises [35, 46, 47], mental health support [6], and muscle-training monitoring [84]. Because these devices require technical and clinical expertise, presenting evaluation data can help clinicians assess their effectiveness in supporting recovery [47, 71, 84]. However, lower-limb wearable devices for gait rehabilitation have not yet been explored.

Outside of HCI, previous work collecting stakeholder input on wearable exoskeletons has focused on powered robotic systems [16, 18, 70, 74] and static orthoses [65]. Their findings reveal interesting but different user expectations in appearance [65, 74], discreetness [16], and functionality [65]. For robotic exoskeletons, users and clinicians prioritize functional capability, with appearance playing a minor role [16, 74]. In contrast, for static braces, users place strong emphasis on aesthetics and stigma [65]. Opinions about discreetness vary across both users and clinicians, with no clear consensus on whether devices should be concealed or openly visible [16]. Therefore, they leave open questions about devices that sit between these categories, e.g., passive yet multifunctional exoskeletons. Moreover, most investigations rely on surveys or interviews rather than iterative engagement with working prototypes, offering fewer insights into how stakeholders respond to real design constraints. Although a few robotics projects have explored iterative development [52, 55], stakeholder feedback in guiding future designs is missing.

To understand the specific user needs of a passive and adaptive knee-support exoskeleton, we apply participatory and iterative design methods to collect user and clinician input to shape both its interaction mechanism and its physical form.

2.2 Lower Limb Exoskeletons

Wearable exoskeletons augment human movement [51]. While exoskeleton research spans various applications, from monitoring and sensing systems [72, 84] to custom rehabilitation devices [71, 80], we focus on lower-limb exoskeletons designed for walking assistance and rehabilitation. This body of work encompasses both active robotic designs and passive mechanical systems that provide robot-like functionality for gait support.

Powered exoskeletons use sensors, actuators, and control algorithms to stabilize or propel gait [43]. For knee stabilization, stance-control orthoses are widely developed to detect gait phase, commonly via ground contact sensors or inertial sensors, to lock the joint during stance for stability, and to unlock it during swing [29, 36, 61]. This robotic functionality is highly relevant to our design. Some multi-joint robotic exoskeletons also use this method of dynamically locking and unlocking the knee during gait [17], with comprehensive joint support across the hip, knee, and ankle [2, 8, 60].

Quasi-passive exoskeletons sit between fully powered and fully passive designs, using motor-free, passive mechanical elements with minimal active control components [83]. For instance, Shamaei et al. [61] developed knee exoskeletons with mechanical locking that implement dynamic stance control by insole-based heel and toe sensors.

Passive exoskeletons further introduce mechanical triggers [41]. For example, Jiménez et al. [33] presented a brace that engages stability on heel contact. These passive approaches are lighter and simpler, but they frequently rely on a *single contact point* (e.g., heel) and therefore lack the richer stance-phase detection used in robotic stance-control orthoses.

Beyond rehabilitation, exoskeletons in HCI are mainly designed to provide force sensation or augmentation, typically powered. For example, Roam Robotics’ robotic knee brace [5] and MO/GO™

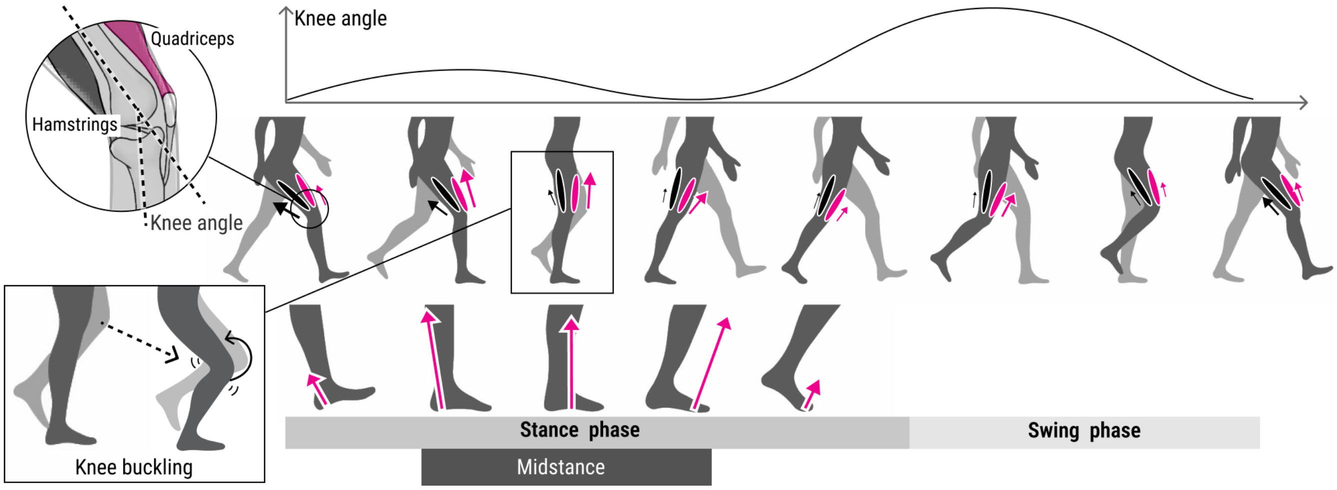


Figure 2: Walking gait cycle, ground reaction force, and knee flexion angles over the gait cycle (adapted from [67]). We highlight the primary muscles involved in knee control (quadriceps and hamstrings) and illustrate how quadriceps weakness can lead to knee buckling during the midstance phase.

motors [4] amplify strength for healthy users in knee motion, while VR haptics simulate varied terrains and gravity using kinesthetic feedback at lower forces [7, 26, 30, 34, 59, 68].

In sum, existing systems demonstrate the value of both powered and passive approaches. We extend this space with a fully mechanical, unpowered design that enables dynamic knee support for gait rehabilitation. Our passive design goes beyond a single contact point to distinguish, e.g., walking from sitting, by implementing a mechanical logic that uses both heel and toe triggers.

2.3 Passive Wearable Devices and Body-Triggered Interactions

This work is situated in HCI fabrication research, with rehabilitation as a specific application domain. Fabrication research includes both sensor-integrated and actuated devices, as well as a rich body of work on unpowered, passive physical artifacts for user interaction. These passive physical interfaces are compelling because of their simplicity and low maintenance. They often rely on users' own manipulation or body movements as both input and power, making such engagement an integral part of the intended interaction. Examples from HCI fabrication research include mechanical computation systems that utilize user input to trigger their shape- or material-changing behaviors [28, 31, 42], mechanisms that render haptic feedback [78, 82], or materials that allow users to switch between different material properties [27, 45, 79], to name just a few. Such human-powered, passive interfaces have also been demonstrated as wearables [32, 45], using the wearer's own motion to trigger and power the changes in the properties and shapes of the devices. We build on this body of work by leveraging the benefits of passive interfaces, which always remain available as people move through different contexts, and we use these principles to develop our passive exoskeleton.

Applying this direction to gait-based interaction, our work proposes *foot-load-triggered mechanisms* that use the mechanical pressure generated during gait as the input signal. Prior hand-scale passive structures are designed for large deformations and low forces. Instead, our system operates under high loads with small displacements, enabling pressure-driven gait detection without the need for electronics. By combining embodied interaction and passive mechanical design, this work advances a new form of body-triggered assistive sensing for wearable rehabilitation devices.

3 BACKGROUND

This section provides background on the walking process in healthy individuals and explains how knee impairments can disrupt it.

3.1 Walking Gait Physiology

Walking commonly follows a stereotypical and repeatable pattern [67], described as the gait cycle. As Figure 2 illustrates, a gait cycle is the period from when one foot touches the ground until the same foot contacts the ground again. Each leg alternates between two phases: **stance phase** when the foot is on the ground and supports body weight, and **swing phase** when the foot lifts off and moves forward for the next step. Within the stance phase, ***midstance*** is particularly critical, as the body's center of mass passes directly over the supporting foot, representing the point of maximum single-limb loading and requiring the greatest stability. During a gait cycle, vertical ground reaction force rises after foot contact, dips at midstance, peaks again during push-off, and then drops to zero after toe-off. The average total vertical ground reaction force equals one body weight.

The knee moves in predictable patterns throughout the gait cycle. At the start of stance, it is nearly straight to support body weight. After heel strike, it bends slightly to absorb shock, then straightens as the body passes over the leg, which is a critical stage requiring strong single-leg support. Before push-off, the knee bends to prepare for swing. During swing, it flexes further to lift the foot clear off the ground and then extends again before the next contact.

These knee movements are primarily controlled by two major muscle groups. The quadriceps (magenta in Figure 2) extend the knee to provide weight support, while the hamstrings (black in Figure 2) flex the knee during swing and control forward motion.

3.2 Atypical Gait Due to Knee Impairments

When muscles are weakened by aging [19], injury, or disease, the smooth rhythm of the gait cycle is disrupted, and people may adopt compensatory walking patterns to stay upright [69]. Quadriceps weakness reduces the ability to stabilize the knee during stance, which can lead to sudden buckling and falls (Figure 2). Hamstring weakness reduces control during swing and stance, often resulting in knee hyperextension (genu recurvatum). This hyperextension places abnormal stress on the joint, increases instability, and raises the risk of long-term damage [15].

These atypical gaits are not only inefficient but also unsafe. To address them, knee braces and exoskeletons are widely used in rehabilitation. By resisting collapse and preventing hyperextension, they support weakened muscles and help restore safer, more confident walking. However, as discussed in Section 1 Introduction, many existing devices are too rigid or uncomfortable, leading patients to avoid wearing them and ultimately reducing their effectiveness in rehabilitation. This motivated us to gather further insights directly from experts and patients.

4 FORMATIVE INTERVIEW

To inform the design of a knee exoskeleton that is both effective for gait rehabilitation and acceptable for everyday use, we conducted semi-structured interviews with medical experts and prospective end users. The goals were to understand: (1) patients' experiences with existing orthoses and exoskeletons; (2) clinical perspectives on functions and forms of assistance that meaningfully support gait rehabilitation; and (3) design requirements that influence long-term acceptance and use.

4.1 Participants

We conducted one-hour, one-on-one semi-structured interviews with 6 participants. To participate in our study, participants must be either: a licensed healthcare professional with rehabilitation or orthopedic experience, or a patient with knee mobility limitations who is currently using or needs to use an orthosis for assistance. Among the 6 participants we recruited, 4 are medical experts who treat patients in gait rehabilitation, and the other two are patients who need to use orthoses. Their background information is in Figure 3. Note that among clinicians, P4's background differs slightly, as they observe neurological patients rather than working directly with orthoses, but they provide a valuable perspective on broader patient populations who may need knee braces and can assess our target users within the wider context of neurological conditions. Regarding patients, PP1 (stroke survivor) uses an Ankle-Foot Orthosis (AFO) but experiences knee buckling and falls, having tried multiple knee braces and exoskeletons. PP6 (multiple sclerosis) transitioned from long-term wheelchair use to gait training recently, and their doctor thought they could benefit from trying a knee brace.

ID	Age	Gender	Condition	Duration	Orthoses	Usage
PP1	55	M	Stroke	43 years	(now) AFO (previously) C-brace, SCO	Left leg
PP6	69	M	Multiple sclerosis	35 years	AFO	Both legs

(a) Patients

ID	Professional Background	Primary Patient Populations	Orthoses
PC2	Physical therapist	Musculoskeletal conditions, athletes, and older adults after knee replacements	Knee braces, Stabilization braces
PC3	Physical therapist	Neurological conditions (stroke, SCI, TBI, degenerative diseases)	AFOs, KAFOs
PC4	Resident physician	General neurology in adult patients	
PC5	Physician	Neurological conditions (stroke, TBI, SCI)	AFOs, KAFOs

(b) Clinians

Figure 3: Participants' demographic information.

Participants were compensated with $25. The study followed local ethics procedures [1], and participants provided informed consent.

4.2 Procedure

We asked two sets of questions tailored to different stakeholder perspectives. For clinicians, we asked specific patient populations who could benefit from knee braces, current treatment protocols, opinions about current orthoses, and observed patient compliance. For patients, we focused on their medical conditions, walking challenges, experiences with existing orthotic devices, satisfaction levels, and usage compliance.

The interviews were conducted through Zoom and were audio-recorded and transcribed. We coded the transcripts through thematic analysis [10]. Two researchers in the research team independently coded all the transcripts and met to resolve conflicts. The list of codes was consolidated after reaching a consensus and updating the codes. Then, the research team grouped the codes using affinity diagramming [23] into themes: (1) patients' diagnoses and mobility challenges, (2) patients' rehabilitation trajectory from injury to hospital to at-home recovery, (3) the benefits and shortcomings of current orthoses, and (4) desired features for future knee braces. Through iterative discussion, the research team consolidated the themes and synthesized the design goals.

4.3 Results

In this section, we first describe clinicians' and patients' perspectives on the role of orthoses throughout rehabilitation, and then present three major design goals for knee exoskeletons that emerged

[1] Carnegie Mellon University IRB #STUDY2025_00000229

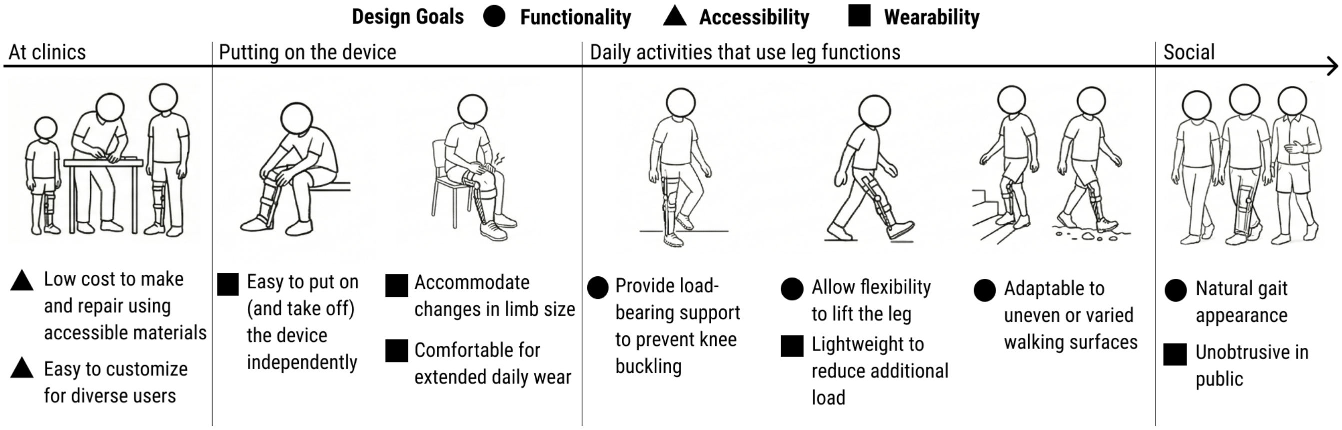


Figure 4: Design goals for knee orthoses across the gait rehabilitation journey.

from the interview results: ***Functionality***, ***Accessibility***, and ***Wearability***. These goals reflect users' needs across their rehabilitation journey (Figure 4) and inform our exoskeleton design. We elaborate on each goal below.

4.3.1 *Expected Roles of Orthoses in Rehabilitation.*

According to clinicians (PC2–PC5), knee orthoses are recommended to restore mobility and independence in daily activities, typically for rehabilitation after injury, stroke (PP1), or prolonged wheelchair use (PP6). These patients often experience knee hyperextension, muscle weakness, or instability (PC2–PC5, PP6). These orthoses should provide physical assistance for gait rehabilitation and daily mobility. By offering reliable support and safety, orthoses can further provide psychological reassurance by reducing fear of falling, thus "*allowing them to engage in more rehabilitation*" (PC4).

4.3.2 *G1: Functionality.*

A core function is that knee orthoses must reliably prevent knee buckling when support is needed, without hindering movement when it is not.

Load-bearing support to prevent knee buckling. The exoskeleton must provide timely and sufficient support, particularly during leg loading when collapse may occur. Without adequate support, patients experience fatigue, tripping, and imbalance (PC5, PP6), which increases anxiety and fear of falling (PC4).

Flexibility for bending and lifting motions. In addition to support, knee orthoses are expected not to hinder normal movements like leg lifting, sitting, or stair climbing (PP1, PC3, PC5). Conventional KAFOs (Knee-Ankle-Foot Orthoses) lock the knee in full extension for stability, effectively preventing "*sudden buckling*" (PP1, PC2–PC5), but introduce new mobility challenges. Clinicians report that patients are "*fighting against the KAFO on bending the knees*" during walking and sit-stand transitions (PC3), and they experience difficulty landing the braced leg stably on uneven terrain (PP1) or stairs (PC3), as demonstrated in Figure 5a. Because the knee cannot flex, users must hike the hip for ground clearance, which increases metabolic effort [14] and reliance on additional walking aids such as canes or caregiver support (PP1, PP6). In contrast, supporting more natural gait mechanics can reduce compensatory effort and minimize unwanted social attention.

Dynamic adaptation for arbitrary walking surfaces. Beyond phase-based support, orthoses must also adapt to diverse real-world use contexts and daily activities (e.g., sit-stand transitions, stair climbing), which vary across users' homes, community facilities, and surrounding terrains (PC5). These contexts demand more reliable phase detection and thus different timings of providing stability and flexibility.

While advanced robotic braces (e.g., stance-control orthoses, C-Brace) can enable more natural gait patterns (Figure 5b) with gait phase detection, clinicians reported that they "*can be clunky and lock unexpectedly*" in real-world use (PC5), making them difficult to learn and get used to (PP1).

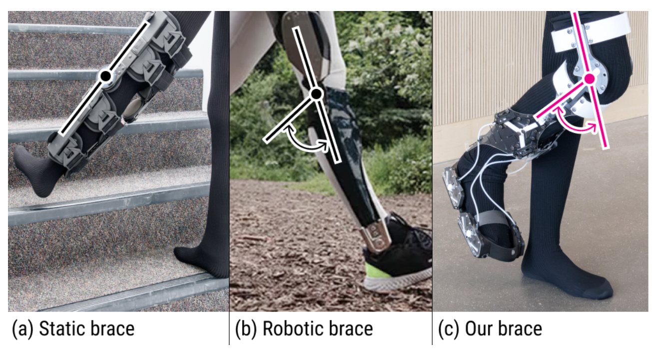


Figure 5: Comparison of the range of motions of (a) a static, (b) a robotic [1], and (c) our brace.

4.3.3 *G2: Wearability.*

To support long-term daily use, the exoskeleton must be comfortable, unobtrusive, lightweight, low-profile, and socially acceptable to wear. Limitations in any of these factors can reduce user adoption.

Ease of donning and doffing is the first step of use. Independent donning and doffing is critical for adoption and a prerequisite for clinicians to consider a device feasible for home use. As PC3 emphasized, "*If they can't get it on themselves, they're never gonna wear it*" (PC3). Moreover, when donning is difficult, patients become stressed, and braces are more likely to be damaged (PC3, PC5, PP6).

Adjustable fit for swelling limb. Physical changes such as swelling or muscle atrophy can alter limb size and affect fit (PC5). Therefore, a safe and comfortable fit must accommodate variations in limb size and shape. Poor fit and swelling-related pressure can cause "*pain*," "*rubbing*," "*different stress throughout the brace*," or even "getting an infection" over extended daily wear (PC3, PC5).

Lightweight to reduce additional effort. As an on-body device, an exoskeleton increases physical effort, especially when it is heavy. As PP1 noted, a major issue with a robotic brace he tried was: "*It was so heavy*" (PP1). Reducing weight was a common request to prevent added walking effort.

Unobtrusiveness to reduce unwanted social attention. Another common complaint about conventional KAFOs and robotic devices relates to their bulk and noise, respectively. When engaging socially, users preferred braces that fit under clothing and remain quiet. PP1 described frustration with full-length KAFOs: "*Each time I have to wear the thing, the pants don't come on, it's just a real hassle.*" Noise was also unacceptable: "*Any sound I will not use, if it's making too much sound*" (PP1). One stance-control device was abandoned specifically due to audible clicking (PP1).

4.3.4 G3: Accessibility. For broader accessibility, the device should be affordable, easy to obtain, easy to maintain, and usable for diverse users through easy adjustment.

Affordability to lower the total rehabilitation cost. Rehabilitation involves significant costs, including devices, treatments, and therapy. Making rehabilitation devices more affordable would help reduce patients' overall financial burden (PC3). Robotic exoskeletons are typically considered "*expensive*" (PP1, PC2) due to factors such as low production volume, specialized materials, and labor-intensive manufacturing. As one clinician observed, "*cost is always kind of an issue*," making these devices "impractical" for many patients (PC2).

Long-term rehabilitation needs often change, requiring new or modified devices. Users' physical conditions may improve (PP6) or decline due to aging or progressive weakness (PP1), leading to different orthotic needs "*six months, a year, or two years out*" (PC5). For example, slower reaction to perturbations with age (PC2) may require adjustments in support strategy. When repair or adjustment is not feasible or affordable, patients may simply "*not wear it for a period of time*" rather than obtain a new brace (PC5). Therefore, the device should support sustained use over time through low-cost repair, modification, and re-adjustment.

Ease of adjustment for clinicians. During fitting, patients typically try a standard-size brace first to evaluate basic function. Then, they may prescribe a custom-made device, which can improve fit and reduce weight, but require specialized orthotist labor and additional fabrication time (PP1, PC5). Thus, the device should be designed to allow clinicians to adjust it easily across diverse users without the need for extensive customization.

5 DESIGN

To address the aforementioned design requirements, we present a knee exoskeleton design. To address the essential **functions (G1)**, i.e., to both provide support and flexibility during walking, we design a load-triggered mechanism that locks and releases the knee motion in coordination with ground pressure. We design our exoskeleton as a fully passive mechanism, which makes our device lightweight (0.95 kg) and low-cost ($38), therefore **accessible (G2)** to broader populations. Lastly, we keep **wearability (G3)** in mind and construct this research prototype to conform to the leg in a sock-like form factor.

5.1 Overview

Our passive exoskeleton design consists of three main parts, as we show in Figure 6. To support the knee, we design (1) a **lockable joint** that allows free rotation when unlocked and can lock the knee in a straight position. The locking is triggered by (2) two pairs of **load-triggered switches** under the foot. The switches are designed as non-linear springs to activate when ground reaction forces exceed half body weight. Since we only want to lock the joint during midstance, we connect the input of the switches to (3) a **mechanical logic gate** (AND). As we show in Figure 6b, the overall functioning sequence flows as follows: ground pressure triggers the load-triggered switches, which provide input to the logic gate, which in turn locks the joint. The forces are transmitted through a Bowden-style cable actuation, as it is flexible and conformable (implemented as a metal wire running through a tube to propagate displacements). The joint unlocks automatically when the pressure is released from the foot switches, i.e., at heel-off. To distribute locking forces and prevent shear forces during support, we place identical sets of these components on both sides of the leg.

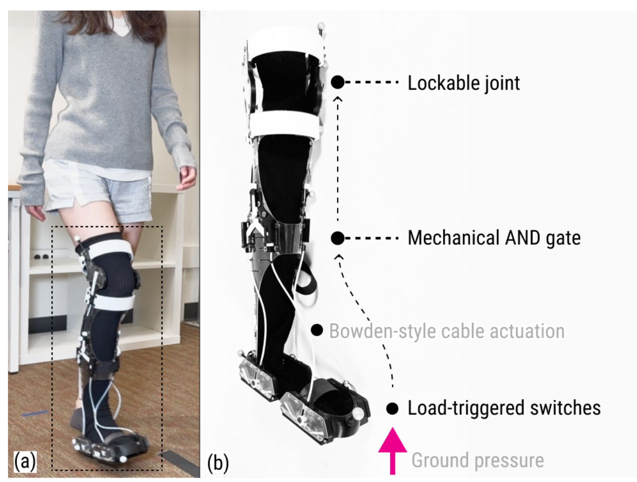


Figure 6: (a) Our knee exoskeleton prototype includes (b) three core components: a lockable knee joint, a mechanical AND logic gate, and two pairs of load-triggered switches under the foot.

Accurate gait cycle detection is essential to support body weight when needed and to allow free knee motion otherwise. Existing exoskeletons typically detect either the joint angle and rotation velocity [1] or ground reaction forces [2, 29]. We adopted the latter, since joint motion detections can be prone to misactivation on stairs and slopes, which is noted in the literature [49, 81] as well as in our formative expert interview (see Section 4). In contrast, ground pressure offers a straightforward indicator of the weight-bearing phase and enables a fully passive design that lowers complexity and cost, thereby improving accessibility.

5.2 Lockable Joint

Our dynamic locking mechanism is built as a hinge with a sliding pin (Figure 7). The joint consists of two discs: an inner disc connected to the lower leg and an outer disc connected to the upper leg. Together, they form a pivoting joint that allows about 85° of knee motion (Figure 7a). A pin slides within a slot on the inner disc and can be pushed upward into a matching slot on the outer disc. When engaged, the pin locks the joint and prevents rotation (Figure 7b). We placed the joints on the inner and outer sides of the knee so they align with the knee's natural axis of rotation, allowing normal bending and straightening while walking or sitting. To keep the brace stable under load, we added rigid rings that wrap around the front and back of the leg. These distribute pressure across a larger area and keep the joint securely aligned with the knee.

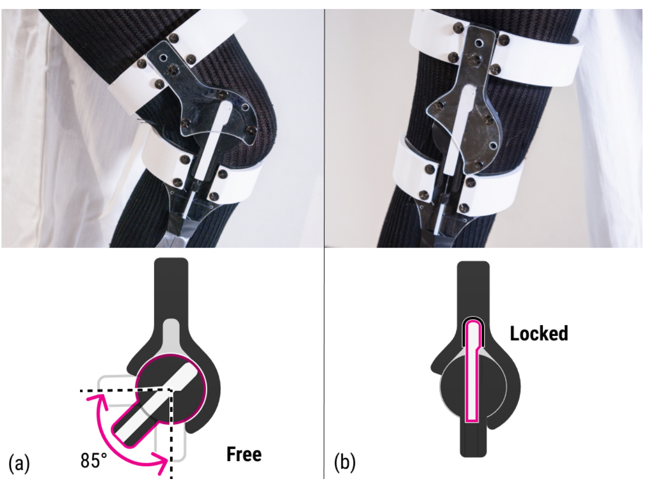


Figure 7: Our lockable joint has two nested discs and a sliding pin. (a) When the pin sits in the inner disc slot, the knee can bend freely for up to 85°. (b) When the pin is pushed into the outer disc slot, it prevents relative motion between the nested discs and thus the upper and lower parts of the knee attached to them.

Knees often carry weight while slightly bent, e.g. during the early midstance phase. To ensure the lock still engages in these situations, we carefully designed the shape of the outer disc, as we show in Figure 8, such that the pin can engage when the knee is within 10° of full extension. On the contrary, if the knee is bent more than this, the mechanism will not lock, which prevents accidental locking.

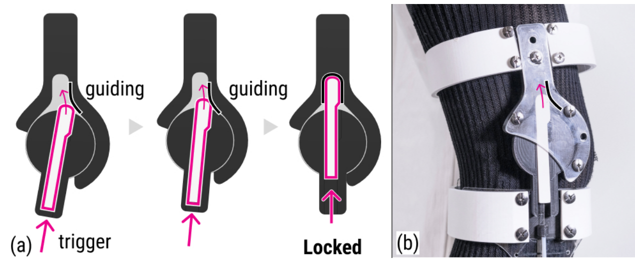


Figure 8: Our guiding geometry (a) enables locking within 10° of full knee extension, (b) ensuring stability even when the leg is slightly bent.

5.3 Mechanical Logic Gate

Knee exoskeletons should only lock during midstance, as this is where the largest load is on the knee. The challenge is to detect midstance mechanically. We achieve this by implementing two switches under the foot; one is triggered when there is pressure on the heel, and the other is triggered when there is pressure on the toe. Gait phases are typically defined by heel and toe contact, which we capture and can distinguish with these switches. However, midstance requires loading on both heel and toe. We solve this by adding mechanical logic, specifically an AND gate. We adopt a popular design of this gate (shown in Figure 9) and make it comfortable. The functionality remains: two input bars at the bottom connect to the Heel and Toe switches, respectively. When either switch is triggered, it displaces its corresponding input bar (Figure 9b,d), but the output bar remains in the "down" position. Only when both switches are triggered does the mechanism mechanically push the output bar to the "up" position (Figure 9c). This output displacement then activates the knee locking mechanism. The output bar returns to the "down" position as soon as one of the switches returns to its unloaded position when the load is removed. To ensure a quick return, we add a coil spring to each Heel and Toe switch.

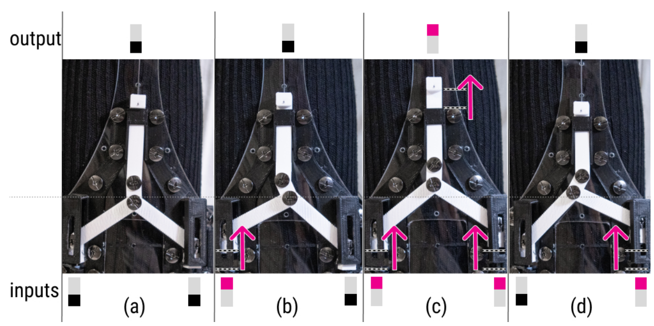


Figure 9: Our mechanical "AND" gate that triggers output only when both Heel and Toe switches (inputs) are activated.

5.4 Load-Triggered Switches

To detect when the knee should lock, we use two pairs of switches at the heel and toe (Figure 10). Importantly, these are not simple displacement switches. A naïve design, where pressing the footplate directly pushes the locking pin, would require rather thick underfoot switches to generate enough displacement. This would interfere with walking and further hinder users' already challenged gait. Worse, such switches would also activate when sitting down, locking the knee at the wrong time.

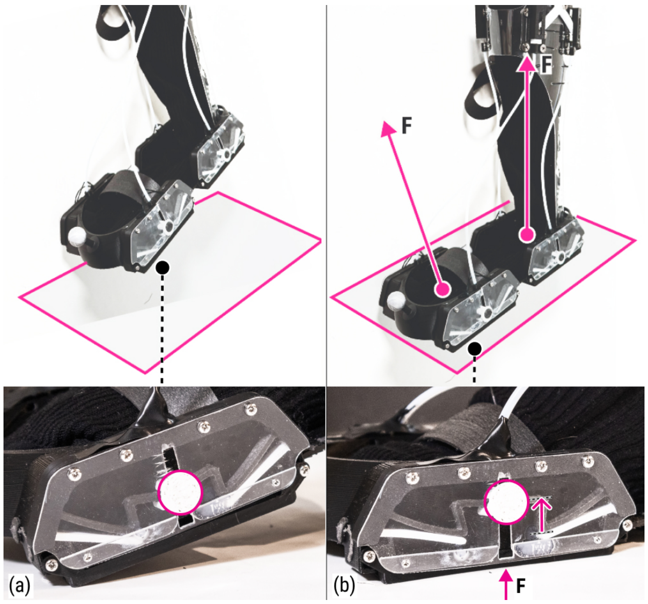


Figure 10: We use two pairs of switches to capture the load under heel and toe. (a) When no load, the switch is down; (b) When loaded, the switch is pushed up.

We address these issues with load-triggered switches that respond to force rather than displacement (Figure 11a,b). At their core is a bistable spring mechanism (Figure 11c), which produces a large output motion from a very small input displacement only once a force threshold is reached. This threshold can be tuned to the wearer's body weight through the spring's geometry. Each switch is calibrated to activate only when a leg supports more than half of the body's weight, which is the condition of single-leg stance during walking. This prevents false triggers during partial loading, such as sitting where users' weight is distributed across both legs.

To minimize gait interference, we need to minimize the displacement of the underfoot plates. To do so, we integrate a lever mechanism into our non-linear switch, as we show in Figure 11d. This lever converts a small input displacement (3 mm) into the larger motion (21 mm) needed to trigger the spring and ultimately lock the knee joint.

Finally, unlike typical bistable springs that stay locked in their new position, our design limits the stroke so the spring returns automatically when the load is removed. A small return spring ensures the switch resets reliably. This creates two well-defined states, i.e., "loaded" and "unloaded", that correspond to the stance and swing phases. Together, these designs (load-dependent thresholds, motion amplification, and automatic reset) allow our switches to dynamically and unobtrusively trigger the knee lock only at the right time. This design is key to making a fully passive, gait-adaptive exoskeleton feasible.

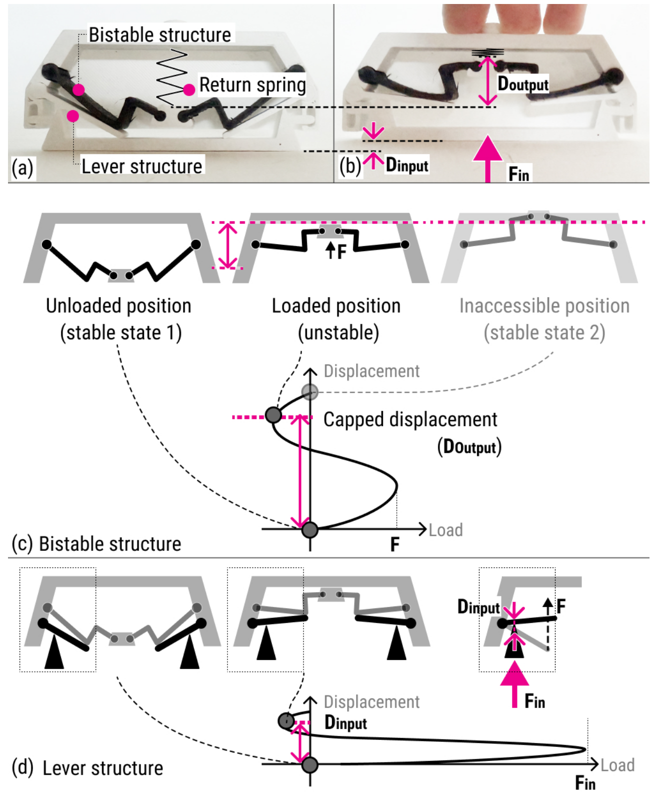


Figure 11: Our bistable switch design can transition between two states: (a) an unloaded position ("down"), which remains stable without any external force, and (b) a loaded position ("up"), which stays when the applied load exceeds a threshold. (c) We constrain the loaded position to prevent full snap-through, enabling automatic reset. (d) A lever mechanism reduces the required input displacement while still generating sufficient motion to engage the knee lock.

Figure 12 shows the entire mechanism during a gait cycle. In summary, our mechanical "AND" logic gate ensures that the knee locks only during the midstance phase, while remaining unlocked during all other gait phases. The timing is designed to match natural gait dynamics: during heel strike, the knee still requires flexion for shock absorption; during midstance, the leg is fully loaded and straightening under body weight, and a weak knee would buckle, so locking provides critical support; during heel-off and swing phases, the leg is transferring weight and needs to bend to clear the ground and take the next step.

5.5 Design for Wearability

Although our exoskeleton is a research prototype, we designed it with wearability in mind. The goal was not to create a final product but rather to demonstrate that our mechanical design can meet the form factor and comfort requirements that future product versions will need. With additional engineering and material refinement, we believe this design direction can enable lightweight, unobtrusive braces that are easy to wear in daily life.

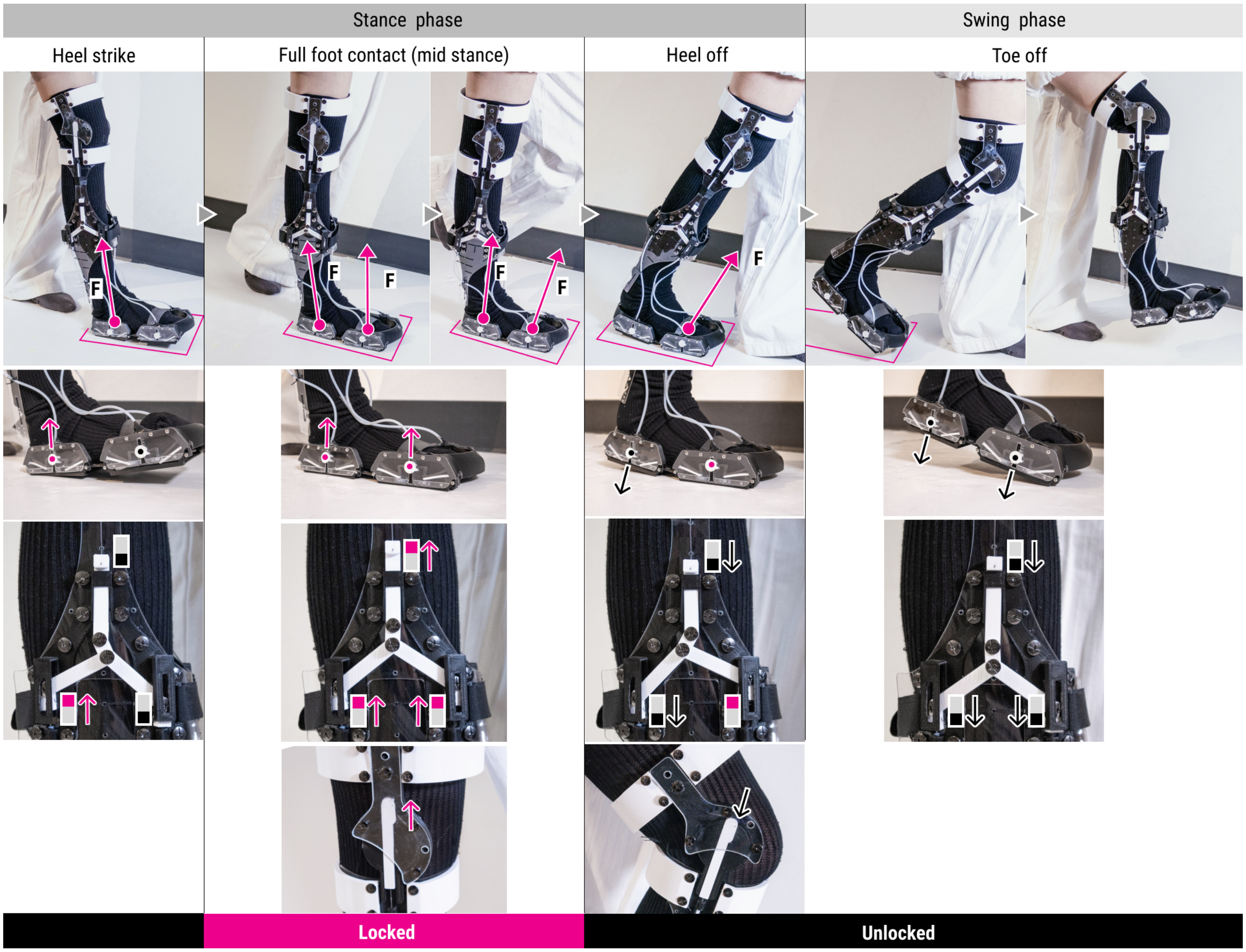


Figure 12: Exoskeleton behavior over a full walking gait cycle. Our exoskeleton locks the knee during the weight-bearing midstance phase and unlocks it during other gait phases, based on the ground forces detected by two pairs of foot switches.

Conformal design & flexibility. Our prototype conforms to the curved shape of the leg while remaining thin, adding only 12 mm of thickness. A 1 mm PET sheet serves as a flexible substrate that holds all components in place. The foot plate is curved to match the sole and allows slight deformation as the foot shape changes during walking (Figure 13a). Because the mechanical logic gate requires additional width, we added thin slots in the connecting bars (i.e., kerf bending in Figure 13b), which permit lateral bending while still transmitting force vertically. Together, these design choices show that this passive exoskeleton can be made low-profile, lightweight, and comfortable while preserving mechanical functionality during gait.

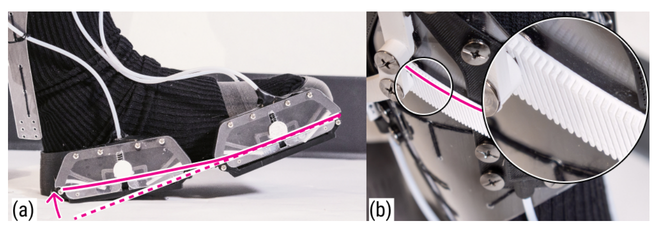


Figure 13: Conformal design. (a) Foot plate that deforms with the foot, and (b) kerf-cut rigid bars that conform around the leg.

Putting on and taking off. Formative interviews highlighted that ease of use is critical for wearability. Inspired by recent approaches that integrate exoskeletons into clothing [58], we attach our components directly to fabric by sewing the PET substrate and foot switches onto a sock (Figure 14). With this, we intend to keep all components well-aligned within one wearable with the goal of being put on quickly without straps or complex adjustments. In this prototype, all rigid components were 3D printed in PLA filament except for the bistable switches, which we 3D printed from carbon-filled Nylon. While this prototype is not intended for long-term use or generalization across users, these choices allowed us to demonstrate feasibility and promise. Future product developers should explore advanced materials to further improve comfort, durability, and robustness.

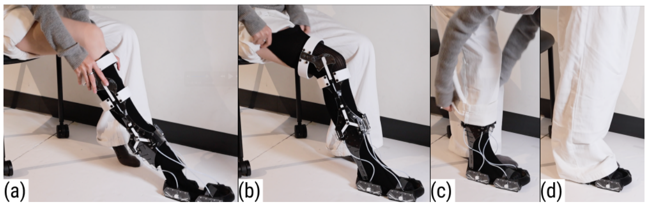


Figure 14: (a-b) The process of putting on the device. (c-d) Our integration of compression socks makes our design wearable under clothing.

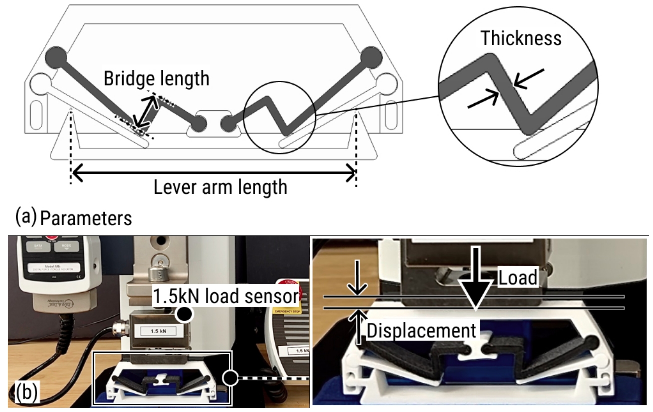


Figure 15: Evaluation setup for foot switch characterization. (a) Geometry parameters of the bistable spring: thickness, bridge length, and lever arm length. (b) Force tester with a 1.5 kN load sensor measuring triggering threshold.

6 TECHNICAL EVALUATION

We conducted three technical evaluations to validate our prototype design. We first evaluated the **load-triggered mechanism**, characterizing the switch threshold and examining how it is influenced by geometric parameters. This allows us to customize the switching threshold based on an individual's weight. Second, we evaluated the **locking mechanism**, testing both the pin displacement required to ensure locking and the maximum locking strength achievable with the current prototype material. Lastly, we tested the overall **function response in the gait cycle**, specifically the timing of locking and unlocking functions in the gait cycle across different conditions, including level-surface walking, uneven-surface walking, and sit-stand transitions. We note that this study does not assess usability or rehabilitation outcomes, which are beyond the scope of this paper.

6.1 Load-Triggered Switch Threshold Characterization

To guide the customization of our load-triggered switch for wearers with different body weights, we characterized how geometric parameters influence the triggering threshold. Three key parameters were tested: the thickness and bridge length of the bistable structure and the lever arm length of the lever structure (as Figure 15a illustrates). These parameters were selected because the switch size is constrained by the foot's length and height, which fixes the overall available space. To maintain a consistent maximum output displacement within this space, the bistable spring angle and total length were kept constant across tested samples as well.

We 3D-printed and tested 19 samples by varying each parameter and keeping the other parameters the same. We used a Mark-10 ESM303 motorized tensile tester with a 1.5kN load sensor (Figure 15b) to characterize each configuration. The compression plate was driven at a constant speed of 1 in/min until the switch triggered. The peak force in each cycle was recorded as the triggering threshold.

As Figure 16a-b shows, lever arm length affects both triggering force and input displacement. Triggering force increases proportionally with lever arm length ($R^2 = 0.90$), while input displacement decreases inversely with lever-arm length ($R^2 = 0.96$). Increasing input displacement from 2.4 mm to 3.4 mm approximately doubles the triggering threshold. Bistable structure thickness influences the triggering threshold following a cubic relation ($R^2 = 0.96$) (Figure 16c). Thresholds rise steeply beyond 3.2 mm, and at 3.6 mm, the spring became overly stiff. Bridge length shows a negligible effect on the triggering threshold ($R^2 = 0.04$) (Figure 16d). Overall, thresholds can be broadly adjusted through thickness, with lever-arm length fine-tuning force upward and displacement downward.

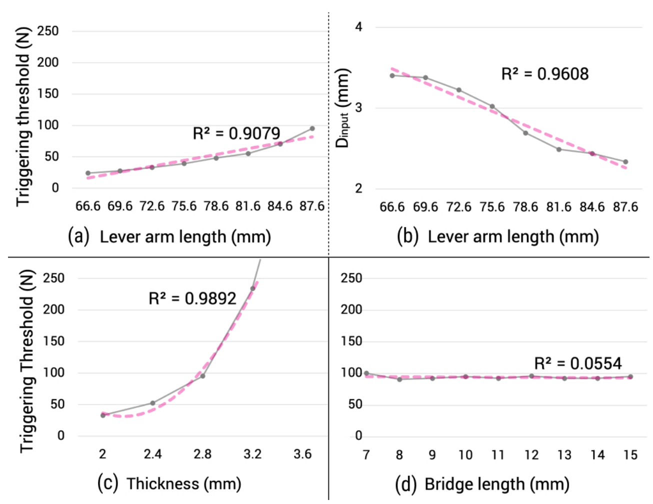
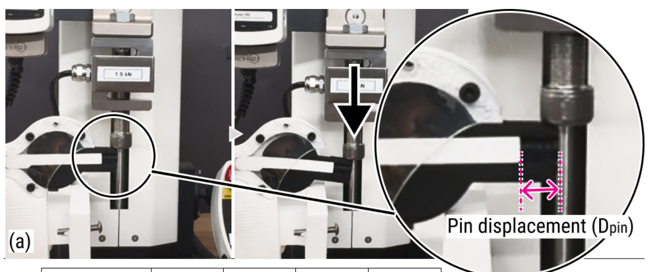


Figure 16: Effect of geometric parameters on the triggering threshold of the load-triggered switch.

6.2 Locking Mechanism Strength and Engagement Threshold Test

To determine the minimum pin insertion required to lock the knee and the corresponding locking strength in our prototype, we used the same force gauge as in the previous test and varied the pin insertion displacement (Figure 17a). As shown in Figure 17b, when the pin displacement was less than 9 mm, the mechanism could not hold under load and returned to the unlocked position as the gauge bent the joint. For insertions of 10 mm or more, the pin remained securely in place when the gauge pushed down to rotate the joint. We recorded the maximum force just before the knee locking mechanism cracked. These results indicate that a minimum insertion of 10 mm reliably locks the brace, providing a maximum locking strength of 265 N, while a 13 mm insertion can sustain up to 510 N. Research shows that a standard static knee brace supports about 40N of force [39] during walking, showing that our current locking mechanism is sufficient to support typical knee loads during gait.



(b)

D_{pin} (mm)	≤9	10	11	13
State	Unlock	Lock	Lock	Lock
Strength (N)	/	265	445	510

Figure 17: Locking mechanism evaluation (a) experiment setup and (b) results.

6.3 Function Response Test

To evaluate the device's functional response, a wearer wore the device to provide realistic loading and gait dynamics, serving as a test platform for system measurements. The goal is to capture the timing and accuracy of the locking and unlocking functions during walking under different conditions. We first examined walking on level ground, then on hills and stairs, and finally analyzed device performance during sit-stand transitions.

6.3.1 Setup and Procedure. Data were recorded using an OptiTrack motion capture system at 120 Hz. Nine retroreflective markers were attached to the outer side of the brace (Figure 18). To capture gait cycle phases, three markers were placed on the front, middle, and back of the shoe. The midpoint between pairs of markers was used to calculate heel distance to the ground (D_{Heel_ground}) for heel contact detection and toe distance to the ground (D_{Toe_ground}) for toe contact detection.

To record timing when load-triggered switches transitioned between loaded and unloaded states, two markers were attached to the bistable switch, allowing us to measure their displacement relative to the sole ($D_{Heel-switch\_sole}$, $D_{Toe-switch\_sole}$). The sole plate was defined by the three shoe markers with thresholds set at the midpoint of the switch stroke range.

Knee joint motion was measured by attaching four markers: two on the upper disc of the lockable joint and two on the lower section. These pairs defined two lines representing leg segments, from which the knee angle was calculated. To determine lock engagement, one of the lower-section markers was attached to the joint pin and the other to the outside of the slot for the output bar of the logic gate, which is fixed; the distance between them indicated pin displacement (D_{pin}). Based on the test above, displacements exceeding 10 mm were classified as lock engagement.

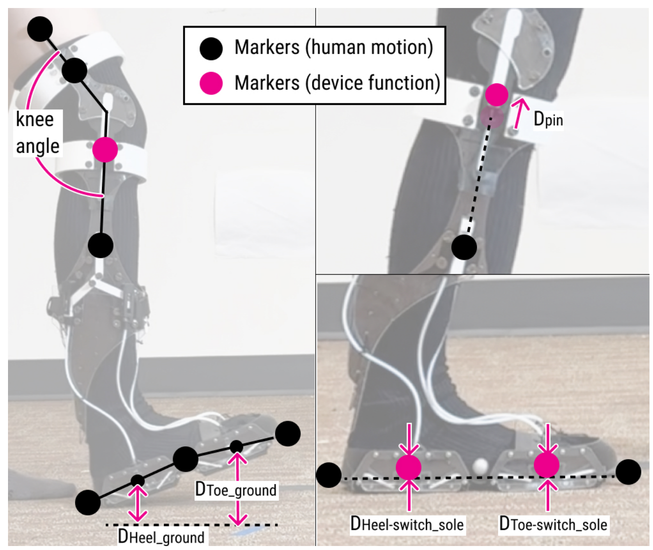


Figure 18: Marker placement and the key variables that are inferred from the marker positions.

As a baseline condition, the wearer also walked without the device. In this case, two markers were placed above the knee and two below the knee to capture natural knee angle profiles for comparison. In each condition, the wearer walked along a straight line while data were recorded. After each session, we applied position prediction algorithms and distance minimization techniques to maintain marker identity despite OptiTrack's ID switching. All measurements were projected onto a defined plane for analysis.

6.3.2 Response Timing During Level Walking. For level walking, we recorded 24 complete gait cycles. The average cycle duration was 3.030 (std = 1.432) seconds. To evaluate the response timing of the switches and the knee lock mechanism, we compared the ground contact periods with the corresponding activation and deactivation times for both Heel and Toe switches (as Figure 19a-b shows).

As a result, Heel switch activation (95.5ms) and toe switch activation (51.4ms) demonstrate rapid response times, which are 3.2% and 1.7% of the average gait cycle duration, respectively. Both switches deactivated before the heel and toe fully left the ground, but after they had begun to lift, consistent with our design–once the foot

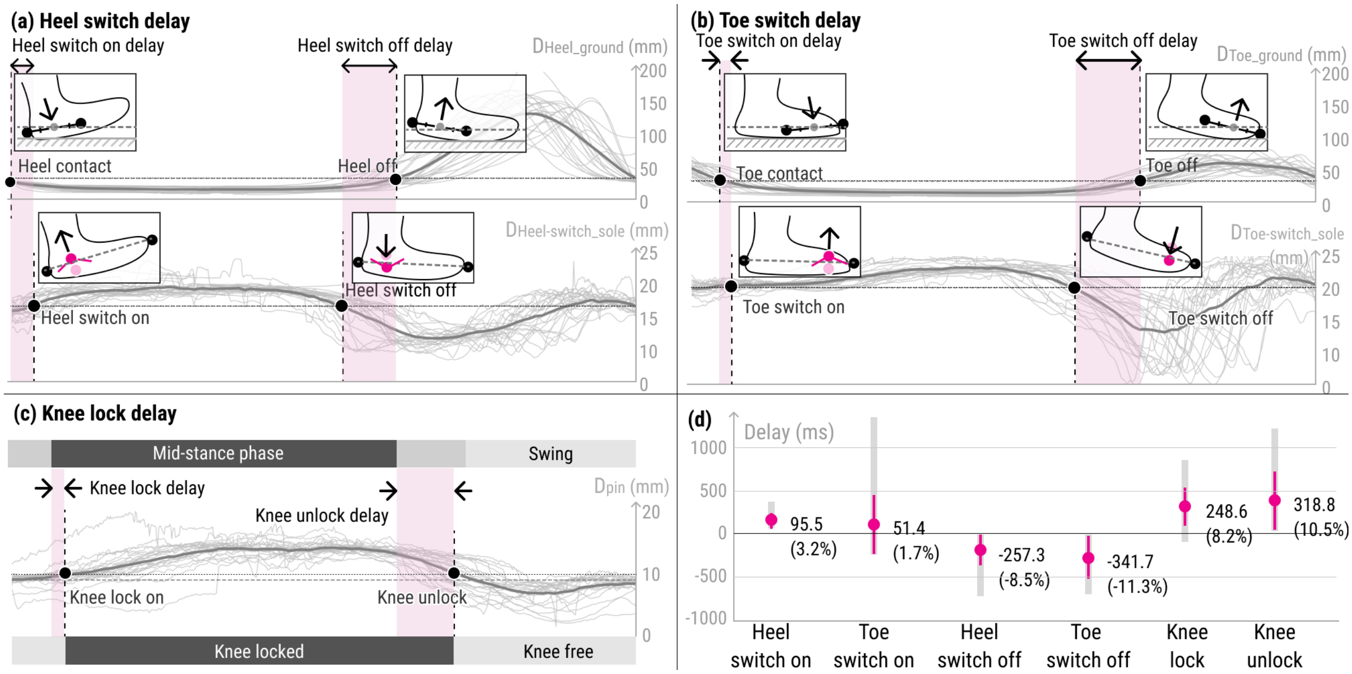


Figure 19: Temporal response analysis across 24 gait cycles: (a) Heel switch activation/deactivation timing, (b) Toe switch timing, (c) Knee lock engagement/disengagement timing. Each curve shows data from all 24 cycles, with the cycle-averaged response in black. Heel and toe switch periods define the stance and swing phases in (c). (d) Response delays and variability: dots = mean, magenta bars = standard deviation, gray bars = full range.

begins to offload, the switches are released and return to their down position.

Using the period of simultaneous heel and toe switch activation to define the midstance phase, we found that the knee lock mechanism exhibited an activation delay of 248.6 ms (8.2% of cycle duration) and a deactivation delay of 318.8 ms (10.5% of cycle duration). These delays indicate that the knee lock engagement and disengagement lag behind the natural midstance phase boundaries.

6.3.3 Range of Motion Analysis During Level Walking. Knee bending angles throughout each gait cycle are presented in Figure 20. The observed range of motion aligned well with the midstance phase timing, indicating that the device's locking and unlocking mechanisms are timed appropriately relative to natural gait. Compared to baseline walking without the device, the overall pattern of knee extension and flexion is preserved. However, while walking and wearing the device, the wearer exhibits a smaller range of knee motion on average (6.45° - 55.81°) compared to walking without a brace (0°- 70°).

6.3.4 Walking on Uneven Surfaces. We applied the same methods to walking on a 15° slope and a 2-tier staircase with 10 gait cycles for each condition. The results, as shown in Figure 21, demonstrate that our device enables walking on uneven surfaces.

As shown in Figure 21a,c, walking downhill and descending stairs produce similar knee extension and flexion patterns, though stair descent exhibits a longer midstance phase and extended knee locking period. In both conditions, the braced leg lands straight

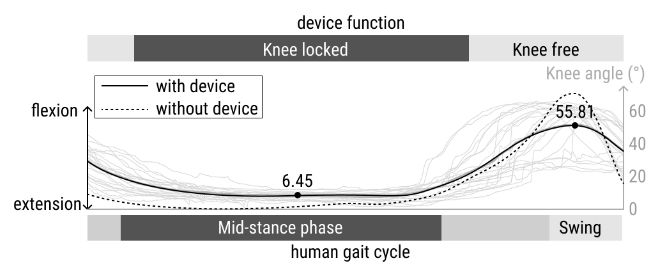


Figure 20: Knee bending angle during level walking with and without the device.

with the lock engaged, providing initial stability. The wearer can then bend the knee by shifting weight forward to the toes, which disengages the lock and allows the other leg to reach the lower step safely. Without this ability to unlock and bend the braced leg while on the higher step, the user would be unable to achieve sufficient clearance for the opposite leg to land on the lower surface.

Uphill and upstairs walking differ slightly: upstairs requires more knee bend at heel strike but less during swing phase (Figure 21b,d). In both conditions, the braced leg steps forward with the knee bent. If the knee cannot bend during this forward step, the wearer would be unable to place the leg on the higher slope or step (see Figure 5a). As weight shifts onto the braced leg, the knee extends, and the lock

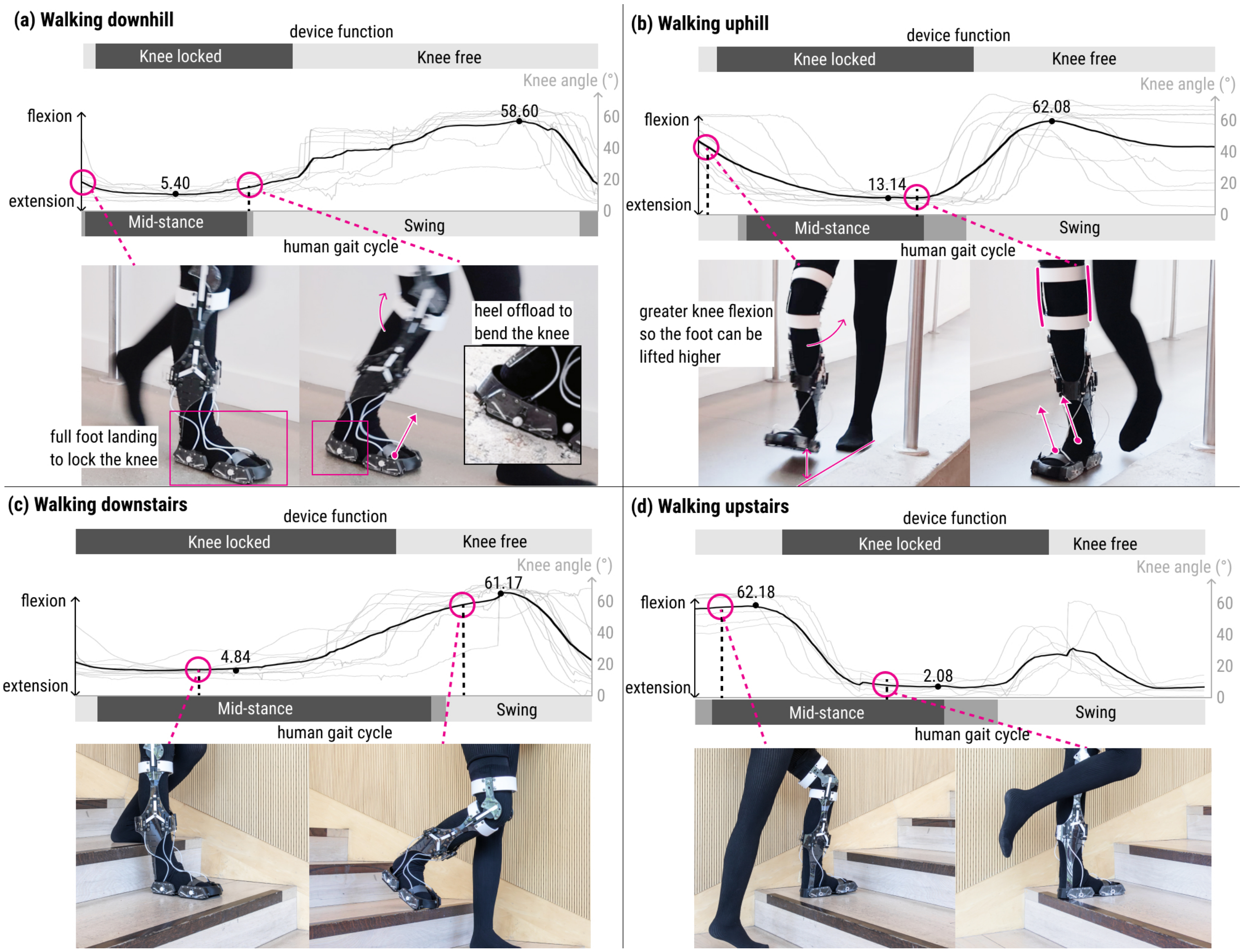


Figure 21: The knee bending angle, gait phase, and device function phases for walking (a) downhill, (b) uphill, (c) downstairs, and (d) upstairs.

engages near full extension, providing stability while supporting the next step.

6.3.5 Sit-Stand Transition. The wearer performed sit-stand-sit for 5 cycles twice. The results are shown in Figure 22. To stand up, the wearer simply rises from the chair; during this motion, the switches are triggered, and the pin tends to push upward. As the knee extends to below 10°, as designed, the pin passes through and locks the knee. Although the switch trigger threshold is higher than during quiet standing, the underfoot pressure increases further during the transition to standing. We hypothesize that this additional pressure enables the locking. For sitting down, the wearer can either distribute weight evenly across both legs to offload the switches or lift the toe or heel to release one switch. This allows the brace to unlock so the wearer can sit.

7 EVALUATION INTERVIEW

To assess how well our design meets the goals identified in formative interviews and to inform future development, we conducted one-hour semi-structured evaluation interviews with clinicians and orthosis users.

7.1 Participants

We conducted evaluation interviews with 6 participants, including 4 returning participants from the formative interviews (PP1, PC3, PC5, and PP6). Information about the newly recruited clinicians is shown in Figure 23.

7.2 Procedure

During the interview, we first presented our design to the participants in three parts: (G1) videos and knee motion data during walking, (G2) device thickness and a video showing how to put it

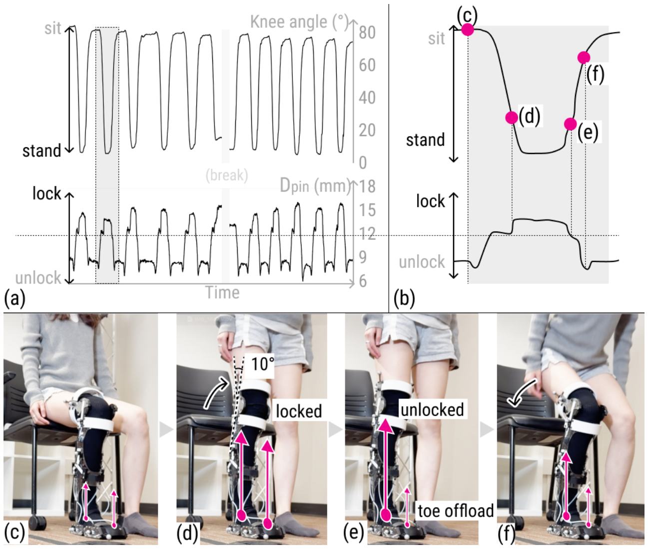


Figure 22: (a) Knee angle and device function during the sit-stand-sit cycles, with (b) a detailed view of a single cycle showing the progression from (c) sitting to (d,e) standing to (f) sitting positions.

ID	Professional Background	Primary Patient Populations	Orthoses
PC7	Physician	Adults in inpatient rehab after stroke (CVA), TBI, or SCI	AFO (solid, hinged, PLS), KAFO, HKAFO
PC8	Physical therapist	Patients with neurological or vestibular conditions	AFO, C-brace, FES

Figure 23: Newly recruited participants' demographic information (clinicians).

on and take it off, and (G3) the cost of the device's materials ($38). For each part, participants were asked to give their assessment and identify key factors to consider. We then asked about the overall strengths and weaknesses of the design, as well as important considerations for future orthoses. Interviews were recorded, transcribed, and analyzed using the same thematic analysis approach as in the formative study.

7.3 Results

Presenting a concrete design helped us collect detailed feedback and better understand users' needs. This section covers insights from the interviews on how stakeholders assess the design and suggest improvements. We listed the key findings in Figure 24a. Overall, participants highlighted the device's promising potential to support rehabilitation. All participants appreciated the functions of our design, particularly its flexibility and intuitive control. The low cost was also highlighted, making the device more accessible to a broader population. While the overall size is generally small, participants suggested improvements to make the device easier to put on and to better accommodate users' own shoes.

7.3.1 Functionality. Participants agreed that the prototype successfully meets the goal of *Functionality* by providing reliable, easy-to-control knee support while preserving natural movement across gait phases and everyday tasks.

Automatic lock-unlock switching enables more natural knee movement. Participants emphasized our implementation of the "*neat feature*" of automatic locking and unlocking, noting that it preserves natural knee motion rather than forcing a fixed, fully extended posture (PP1, PC3, PC5, PC7, PC8). PC3 elaborated that retaining a true swing phase enables users to lift the leg more easily to move forward during walking, supporting our core design rationale. Additionally, PP1 liked that locking only occurs when the knee approaches near full extension, rather than while it remains highly flexed, thereby preventing the accidental locking they experienced with their robotic devices.

Timing of locking and unlocking matches the gait cycle. When reviewing our response analysis data (e.g., Figure 20), most participants agreed the timing aligned well with the gait cycle (PP1, PC3, PC5, PC7, PC8). As PC3 explained, "*That timing is perfect. At initial contact, you want a little bit of knee bend to absorb force,*" supporting our results that locking takes place shortly after initial contact rather than at the moment of landing. However, PC3 also observed that in some cycles, the lock released slightly later than heel-off. Such occasional over-locking could resemble the limitations of a fully static KAFO and may lead users to adopt compensatory gait strategies such as circumduction, highlighting the need to further reduce lock-release latency.

Underfoot load as a gait-phase indicator is easy to learn. Participants found the user-controlled load-triggered mechanism intuitive and easy for patients to understand and operate (PP1, PC5, PC8). For example, PP1 quickly understood the interaction logic: "*if you just lift your leg, it'll unlock,*" and PC8 also noted that users can easily grasp that "*the more weight you put on that leg, the more stable it is.*" PC8 also suggested providing optional feedback to indicate when the knee is securely locked for weight-bearing versus unlocked for bending, especially during initial familiarization with the device. In addition, although our implementation of the customizable triggering thresholds was originally intended to accommodate interpersonal weight differences, participants highlighted the benefit of adapting to evolving user needs, such as changes in strength, mobility, or use of additional assistive devices (e.g., adopting a cane). To achieve this, they noted that the threshold-adjustment process should remain simple for patients or clinicians.

Multiple interaction methods of navigating uneven surfaces and managing sit-stand transitions. All participants are excited about the device's potential support on slopes, steps, and varied surfaces. As PC5 noted, "*Most braces that we have right now can't really account for those differences in grade; this brace could allow people more freedom*". To better support different real-world walking conditions, clinicians also suggested evaluating whether compliant surfaces (e.g., carpet or grass) reduce the underfoot load necessary for triggering the lock (PC5) and incorporating solutions to control side-to-side ankle motion on uneven ground, such as additional ankle stabilization design (PC3, PC6, PC7).

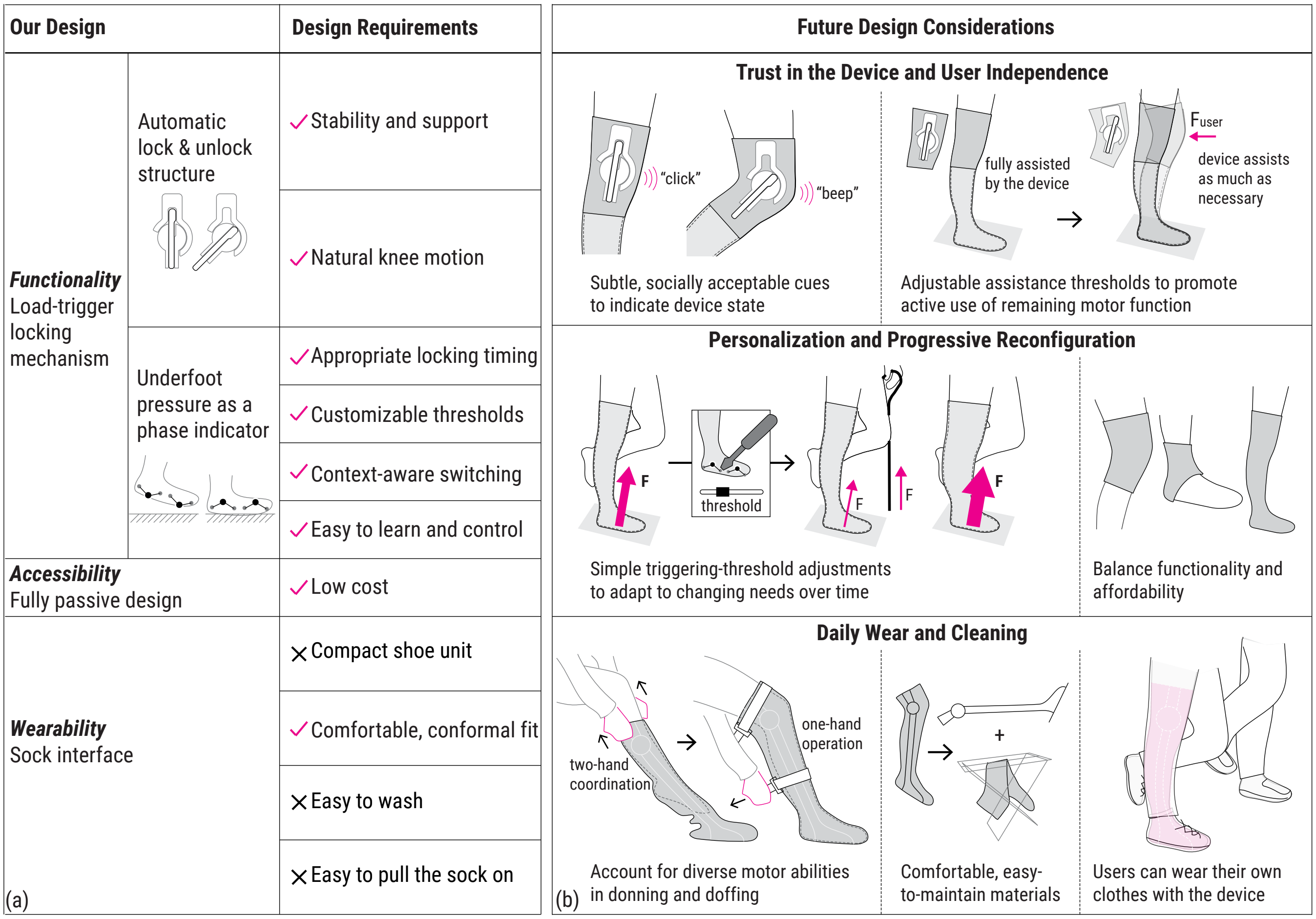


Figure 24: (a) Summary of stakeholder comments on our prototype design. Checkmarks ✓ indicate areas that our current device satisfies, and "×" mark areas for more design opportunities in the future, primarily in *Wearability*. As shown, the current design satisfies most desired requirements—particularly in *Functionality* and *Accessibility*. (b) We illustrate future design changes and considerations to satisfy these requirements and support long-term rehabilitation and everyday use.

For sit-stand transition movements, participants liked that they could control locking and unlocking by shifting weight to different parts of the foot (such as the toe or heel) or between both legs, giving users multiple ways to perform sit-stand transitions depending on their abilities (PP1, PC3, PC5, PC8). PC8 noted that this flexibility can reduce overuse of the stronger leg, "*prevent[ing] the other leg from getting so much wear and tear.*" Additionally, after reviewing the video on sit-stand transitions, PC3 suggested increasing the allowable knee-flexion angle in future designs. While our current maximum was chosen based on typical gait-cycle data, users with lower-extremity weakness may require deeper flexion to generate adequate forward momentum during sit-stand transitions.

7.3.2 Accessibility. Participants felt the device strongly supports *Accessibility* through its significantly low cost in materials, while emphasizing that practical accessibility also depends on clinician awareness, fitting workflow, and clear pathways for adjustment and maintenance.

Participants described the device as highly affordable compared to existing bracing solutions (PP1, PC3, PC5, PC7, PC8), especially considering its dynamic functionality: "*for what it's able to do, that's honestly really impressive. I was expecting it to be quite high. I think that would be pretty accessible for almost everyone that we see*" (PC5).

Participants also emphasized that clinician knowledge and confidence in prescribing and managing the device influence accessibility (PC7, PC3). In particular, supporting clinicians' ability to fit, adjust, and repair the device was seen as important for ensuring that patients are aware of the option (PC3) and feel comfortable navigating future adjustments or maintenance needs (PC8).

Finally, PC5 highlighted the potential accessibility of the device for a diverse range of users, noting that its intuitive, low-cognitive-load operation may reduce the cognitive demands typically associated with gait-assistive devices, and thus suggesting evaluation of the device with a broad range of patients, from those who are cognitively intact and can follow device strategies to those with neurological impairments who may have difficulty perceiving gait-related feedback.

7.3.3 Wearability. Participants reported that the device partially meets the goal of *Wearability* in comfort and lightweight form factor, but identified donning difficulty, sock hygiene, and shoe compatibility as key areas that need refinement for seamless daily use.

The sock interface was considered comfortable (PC5, PC8) and reduced skin contact with rigid parts (PC5, PC7). However, donning the device was often described as the biggest challenge, especially for users with weak arms, limited hand coordination (PC7, PC8), or reduced ankle strength (PC3). They suggested that modular components be used and fastened together with Velcro strips and that the sock be removable for laundering (PP1, PC7). PC5 also emphasized that knee-lock rings should be adjustable to accommodate swelling limbs.

While the overall thickness is comparable to other braces worn under pants, the shoe component was considered bulky and socially limiting (PP1, PC3, PP6, PC7, PC8). As PC7 explained, "*Patients want to wear something that, when they're grocery shopping or going to church, no one can tell that they're wearing it.*".

The lightweight of our device was highly praised (PC5, PC8). As PC5 explained: "*It is pretty light overall, because that's another big consideration. For some patients with significant weakness, if we give them things that are weighing them down... There could still be a big issue for fall.*"

7.3.4 Promising Impact in Rehabilitation. Beyond the design goals, participants described the device as having meaningful potential to support rehabilitation by encouraging active muscle use, improving mobility confidence, and filling a currently underserved niche in knee-buckling assistance.

Encouraging active muscle use. Participants highlighted that the device provides support while still encouraging patients to engage their own muscles (PP1, PC8). In contrast, fully automatic devices can reduce muscle activation due to over-reliance. PC8 explained, "*With devices that automatically move the leg, patients tend to work less, and their muscles turn off a little more. Our device allows them to use the function they have with just a little more assistance where they need it.*"

Supporting more involvement in daily activities. The device was seen as helpful for re-engaging in everyday mobility, especially walking (PC5, PC7, PC8). As PC8 noted, "*That's going to give patients a lot more confidence to stand on that leg, especially if it tends to buckle.*" PC5 also noted that patients may live in a multi-level home. Supporting them going upstairs can give them more freedom at home and even "*allow them to go out into the community*".

A "game changer" for a rehabilitation niche. Clinicians highlighted that the device fills a gap for patients who lack suitable options. PC5 excitedly commented, "*especially for the knee buckling, this could be a game changer.*" PC7 noted it would be most effective after the early recovery stage, "*at least a year out from injury, when patients aren't experiencing rapidly changing return of function.*" They emphasized its role for those with "*more strength than your typical KAFO wearer, but less than your typical solid dorsiflexed AFO wearer—there's a niche in between the two.*" PP1 stressed its novelty: "*Nobody has an articulating knee joint that is controlled by foot pressure—that doesn't exist.*" PC8 added that it "*provides a bit more knee stability for someone with global leg weakness, like a more mobile and active version of a KAFO.*"

8 DISCUSSION: Design Considerations for Future Unpowered Exoskeletons

HCI research [11, 22, 25] highlights the need to develop assistive devices through an integrated process of fabricating, engineering, and evaluating systems with multiple stakeholders. This paper contributes to this research area by surfacing specific design considerations for passive, body-adaptive knee exoskeletons. These considerations (as illustrated in Figure 24b) extend existing work on exoskeleton usability, personalization, and daily integration, and highlight how future designs can become more functional, accessible, and wearable in real-world rehabilitation contexts.

8.1 Trust in the Device and User Independence

Both our formative interviews and prior research [70] indicate that robotic exoskeletons often require a substantial learning period. In contrast, participants described our passive mechanism as simple to learn (PP1, PC5, PC8), though their comments also highlight opportunities to further improve the learning experience to build users' confidence in the device's reliability. However, effective *functionality* in our rehabilitation context should also support active engagement of their own motor functions during movement, as discussed below.

Provide subtle, socially acceptable feedback. Prior research on robotic ankle exoskeletons suggested that haptic cues can be used to signal forthcoming device actions [76]. Our interviews with clinicians and patients provide a more nuanced understanding. For users to trust the device, they need to be aware of when it is providing support—for example, whether the knee is currently locked (PC8). Feedback that is too overt, such as loud clicks or visible mechanical movements, risks making patients self-conscious in public (PP1). These sensitivities highlight that subtle yet perceivable cues, as widely researched in the haptics field in HCI, such as quiet tactile feedback (e.g. [27, 82]) or low-profile form factors (e.g. [66]), can inform users of the exoskeleton status without drawing unwanted attention.

Define assistance thresholds. Prior work on robotic exoskeletons highlights the need for adjustable assistance, providing only as much support as needed [16, 74]. Strong mechanical assistance can improve safety, build confidence, and prevent knee buckling or falls. Yet, if the brace takes over too much of the work, patients may reduce their own muscle engagement, risking further weakness over time. Our design addresses this by offering what participants (PP1, PC8) described as 'just-enough' support: users initiate knee extension using their own muscle strength while the device then secures the joint and carries the load. Looking ahead, introducing tunable locking strength (as noted by PC8) could further tailor support to each individual's motor abilities and further encourage muscle use during stabilization. Recent fabrication research has explored mechanisms for adjustable stiffness [7, 46, 62]. Applying these advances to lower-limb orthoses may enable more precise and personalized assistance. Additionally, future research into sensing technologies [75, 84] that assess or monitor a user's preserved motor function could ensure that support remains sufficient for safety, yet never overpowering.

8.2 Personalization and Progressive Reconfiguration

Customizability is widely emphasized across mobility aids [11], robotic exoskeletons [16], and upper-limb rehabilitation devices [46]. In extending this requirement to a passive knee exoskeleton for long-term use and a diverse patient population, our findings surface two design considerations.

Balance functionality with affordability. More features are not always better. Complex features can make a device bulky, costly, and less relevant for many patients. Core functions should be delivered at low cost while still allowing for optional add-ons to meet individual needs and preferences. For example, clinicians (PC3, PC7) highlighted that some patients experience side-to-side ankle instability; thus, an optional, reconfigurable ankle-lock module could address this need, while remaining fully removable for users who require only knee support (PP1).

Enable adjustments over time. The therapists in our study emphasized that patients' mobility and strength change over the course of rehabilitation. Therefore, our device should accommodate this by allowing adjustments such as modifying the load required to trigger a lock or altering the permitted range of motion. Since patients may lack the ability or knowledge to make such changes independently, these adjustments should be simple and intuitive, including making parts swappable, color-code settings, or similar.

8.3 The Brace as Functional Clothing: Daily Wear and Cleaning

Prior research reveals contrasting expectations concerning device aesthetics among mobility technology users. Robotic exoskeleton users typically prioritize performance over appearance [16, 18, 74], while users of static mobility aids strongly value aesthetics [11, 65]. Our findings sit between these two perspectives: for a passive yet multifunctional brace, our participants valued the technical merits but also emphasized the everyday practicalities of wearing it. Viewing the device as *functional clothing* draws attention to these mundane but essential aspects, such as ease of cleaning, maintaining hygiene, and fitting naturally with users' existing shoes and pants, without adding burden to daily routines.

Account for motor abilities in donning and doffing. Ease of putting on and taking off is a major predictor of daily use (PC3). While reducing the number of steps is valuable, our interviews highlighted that the type of steps matters as much as the quantity (PC7). For example, pulling on a sock may be difficult for some stroke patients, whereas fastening several Velcro straps may be easier. Designers should consider the motor abilities required for each step and balance minimizing effort with matching user abilities.

Select comfortable and maintainable materials. Materials that contact the skin should be soft, breathable, and washable to avoid irritation from daily use. At the same time, load-bearing components must withstand repeated stress without breaking (PC5). Designers should consider modular or replaceable components so that softer comfort layers can be swapped when worn out without compromising the structural parts.

Allow users to wear their own shoes. If we consider the device as clothing, it should fit seamlessly into users' everyday routines and personal styles. The design should allow users to wear their own shoes and pants for comfort, familiarity, and self-expression rather than requiring specialized or visibly different apparel.

9 LIMITATIONS & FUTURE WORK

The scope of this paper was to investigate the need for knee exoskeletons, propose a functional design concept, and evaluate it through expert interviews. Our focus was on addressing core challenges of knee assistance and surfacing design considerations for future development rather than delivering a product-ready device. Below, we outline limitations of our current prototype and evaluation, as well as future directions that can build on this work.

9.1 Generalizing the Design

Our prototype was tailored to a single wearer to establish and validate the underlying mechanism. For broader application, exoskeletons will need to accommodate a wide range of users with different body sizes, weights, and conditions. For example, the load-triggered switches must be calibrated depending on the wearer's weight, whether they exhibit partial weight bearing, the recommended level of assistance, etc. Future work should focus on systematic testing across users and on adjustment mechanisms that make personalization straightforward. In addition, design tools that facilitate customization will be needed to support wider adoption. In this paper, we established that passive mechanisms can deliver adaptive knee support and that the next step is extending this principle to robustly serve a broader population.

9.2 Durability and Robustness

Our prototype used PLA and nylon filament. These materials are appropriate for rapid iteration but not for sustained daily use. Current braces use materials like metal (heavy) or carbon fiber (expensive). To make our exoskeleton design closer to real-world impact, exploration of durable, fatigue-resistant materials is needed. Future iterations should investigate low-cost, tough materials and potentially investigate hybrid solutions that combine strong structural components if needed. Beyond longevity, safety is highly important. Optimizing the balance between strength, comfort, safety, and cost will be central to advancing toward practical, everyday exoskeletons.

9.3 Form Factor and Integration

Although wearability was an important design consideration, refining the form factor into a polished device was beyond the scope of this paper. Future work should aim for seamless integration between structural parts, skin-contact materials, and the mechanical components of our exoskeleton. Promising directions include custom knitted socks with integrated channels for components or semi-rigid composite braces that allow modular mounting of mechanisms. Another improvement is relocating bistable switches higher on the leg, allowing wearers to use their own shoes. Here, underfoot plates could transfer displacement upward through lightweight linkages. Design tools for automated customization, such as generating brace geometries from body scans, could further support fitting and personalization.

9.4 Toward Gait Studies and Clinical Evaluation

While this paper focused on establishing the design and demonstrating its working principle, the next step is to evaluate biomechanical outcomes through gait studies. Such studies will examine parameters such as stability, metabolic energy cost, and user confidence. It will allow direct comparison to static braces and robotic exoskeletons. Additionally, clinical studies to assess long-term rehabilitation impact, e.g., whether the brace encourages muscle activation or inadvertently discourages it, should be carefully planned and conducted as well.

10 CONCLUSION

This work contributes to the design of accessible exoskeletons by showing how HCI methods and fabrication approaches can inform new classes of assistive devices. Our formative interviews with clinicians and orthosis users surfaced design requirements that extend beyond mechanical function, such as wearability, ease of use, and confidence in everyday mobility. Guided by these insights, we designed a fully passive knee exoskeleton that uses ground reaction forces to trigger a load-sensitive locking mechanism. This allows the knee to lock securely during stance and release naturally during swing—capabilities typically reserved for robotic systems, but here achieved without electronics or motors.

This work represents an important step forward in the design of accessible exoskeletons. We not only invented a new class of knee brace, i.e., a fully passive device that mimics some of the adaptive behavior of robotic systems, but also realized this concept in a functional prototype and qualitatively assessed it with clinicians and orthosis users. These achievements demonstrate the feasibility and promise of passive mechanisms for everyday mobility support and provide actionable design considerations for future development. At the same time, we emphasize that this is a research prototype rather than a commercial product. As such, its role is to explore possibilities, surface insights, and chart a path for future advances in assistive technology.

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