Design and Development of an Active Knee Exoskeleton for Gait Assistance

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Abstract

Active knee exoskeletons require controllers that adapt assistance across different locomotion conditions while maintaining a simple embedded architecture. This study presents a kinematics-based adaptive controller that adjusts assistance timing and magnitude using onboard knee kinematics and predefined user body mass. The controller was evaluated during level walking and stair ascent at two walking speeds. Distinct assistance profiles were generated across all locomotion conditions. Exoskeleton-assisted locomotion was accompanied by reduced muscle activity and changes in knee kinematics, although these differences did not reach statistical significance. The findings demonstrate the feasibility of the proposed control strategy and identify opportunities for further refinement.

Keywords

Knee Exoskeleton, Adaptive Control, Kinematics-Based Control, Quasi-Direct Drive, Wearable Robotics

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1. Introduction

Walking and stair ascent are integral components of daily mobility and place substantial mechanical loads on the knee joint. Tibiofemoral contact forces reach approximately 2.6 times body weight during level walking and can exceed three times body weight during stair ascent [1]. For individuals with reduced muscle strength, joint degeneration, or those undergoing rehabilitation, these loads can translate into reduced mobility and increased physical effort [2]. Consequently, reducing the physical demands of locomotion has become a major objective in the development of wearable assistive technologies. Among these, powered knee exoskeletons provide assistive knee torque to reduce the muscular effort required during locomotion [3][4].

The effectiveness of this assistance depends on when and how it is delivered. The controller therefore plays a central role by determining the timing and magnitude of the applied assistance [5][7]. The required timing and magnitude of assistance vary across locomotion conditions and between users. Consequently, a controller tuned for level walking cannot be expected to perform equally well during stair ascent. Existing approaches often improve adaptability by incorporating additional sensing or more sophisticated control architectures, increasing overall system complexity [5],[8],[9]. Developing controllers that adapt assistance across different locomotion conditions while maintaining a simple embedded architecture therefore remains a central challenge in powered knee exoskeleton control.

2. State of the Art

Control strategies for lower-limb exoskeletons differ in how assistance is generated and adapted to the user and locomotion task [5].

Finite-state and gait-phase-based controllers divide locomotion into predefined phases and apply assistance according to the detected movement state [10]. Qian et al. [8], for example, used gait-phase detection within a hierarchical control architecture to coordinate walking assistance. Similarly, Liu and Wang [10] combined locomotion-mode recognition with adaptive torque generation across different terrains. These approaches are well suited for real-time implementation but rely on predefined gait logic, reference trajectories, or manually tuned control parameters.

Kinematics-based approaches generate assistance directly from joint motion using onboard sensor measurements. Huang et al. [11] developed a lightweight quasi-direct-drive knee exoskeleton with a stiffness-based controller that continuously estimates the biological knee moment from knee kinematics, providing smooth assistance across different walking speeds and speed transitions. This controller generates assistance through a continuous stiffness relationship developed for walking. Extending the same formulation to additional locomotion tasks involves task-specific assistance models together with locomotion-mode recognition and transition strategies.

Greater individualization can be achieved through physiological optimization or learning-based control. Human-in-the-loop optimization has demonstrated substantial reductions in metabolic cost by identifying user-specific assistance parameters through repeated physiological measurements [12]. Learning-based approaches instead infer biomechanical quantities from wearable sensor data. Molinaro et al. [6] estimated biological joint moments using a trained neural network to adapt exoskeleton assistance, while subsequent work extended this concept across multiple joints and locomotion activities [7]. Although these approaches provide greater adaptability, they rely on physiological optimization procedures or representative training data together with model development.

Rather than pursuing greater adaptability through additional sensing modalities or increasingly sophisticated control architectures, the present work investigates whether locomotion-dependent assistance can be achieved by more effectively exploiting the onboard kinematic information already available within a low-complexity embedded architecture. This design philosophy emphasizes practical implementation while seeking to maintain adaptability across different locomotion conditions. This leads to the following research question: Can locomotion-dependent assistance across level walking and stair ascent be achieved by more effectively exploiting the kinematic information already available on a minimal-sensor embedded platform while maintaining a low-complexity controller architecture?

3. Methodology

This section presents the developed knee exoskeleton, the proposed control strategy, and the experimental methodology used to evaluate system performance.

3.1. Exoskeleton System

Figure 1 provides an overview of the developed knee exoskeleton. Figure 1a shows the assembled prototype, while Figure 1b illustrates the main mechanical components of the system. The exoskeleton was designed as a single-degree-of-freedom system providing active assistance in knee flexion and extension. This architecture provides a mechanically simple design focused on the primary knee motion required during walking and stair ascent. The mechanical structure consists of a fixed thigh attachment, an adjustable shank attachment, and passive alignment joints that accommodate the user's leg during movement.

Two inertial measurement units (IMUs) mounted on the thigh and shank, together with an actuator-integrated encoder, provide the onboard sensing used by the controller. The relative orientation of the thigh and shank is used to estimate knee motion, while the encoder measures actuator position and velocity. A quasi-direct-drive actuator generates the commanded assistive torque while providing low mechanical resistance when assistance is inactive [13][14].

3.2. Adaptive Control Strategy

The control strategy was designed to provide adaptive assistance using only signals that are continuously available from the onboard sensors together with predefined body mass. This approach reduces sensing and implementation complexity while allowing assistance to adapt to different locomotion conditions and users. Figure 2 illustrates the proposed closed-loop adaptive control architecture.

Assistance is adapted at three levels: locomotion, movement, and user. Locomotion-dependent logic determines assistance timing separately for level walking and stair ascent. Assistance is applied during knee extension because this phase contributes substantially to body support and forward progression. Knee angular velocity was selected for movement-dependent adaptation because it reflects changes in movement dynamics and is continuously available from the onboard sensors in real time. During walking, the transition from knee flexion to extension is identified by the change in the sign of knee angular velocity from negative to positive (zero crossing). Assistance is then applied during extension, allowing the control strategy to follow the natural knee motion rather than relying on a fixed joint-angle threshold.

During stair ascent, the controller includes an additional confirmation step before initiating assistance during knee extension. Depending on the detected movement progression, assistance is triggered once the required kinematic and temporal conditions are satisfied. This strategy accounts for variations in individual stair-ascent patterns while maintaining robust assistance timing.

Assistance magnitude is adapted through the proportional controller using both knee angular velocity and knee flexion. The magnitude of knee angular velocity, $|\omega|$, adjusts the proportional gain K_p within locomotion-specific limits, with larger angular velocities resulting in a higher gain. Knee flexion simultaneously influences the generated assistance through the position error, $pdes - p$, with deeper knee flexion producing greater assistance. During stair ascent, knee flexion depth additionally adjusts the desired position $pdes$, further increasing the assistance provided during extension. At the user level, the available gain range is scaled according to body mass entered before operation. Body mass was selected because it is a readily available user-specific parameter that provides a simple and practical basis for scaling assistance.

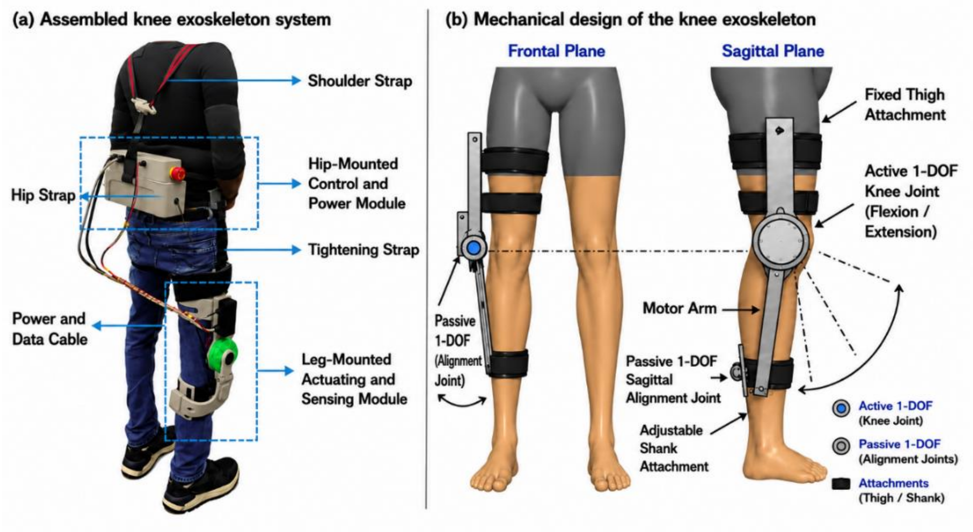

Figure 1 Developed active knee exoskeleton: (a) assembled prototype and (b) mechanical system architecture.

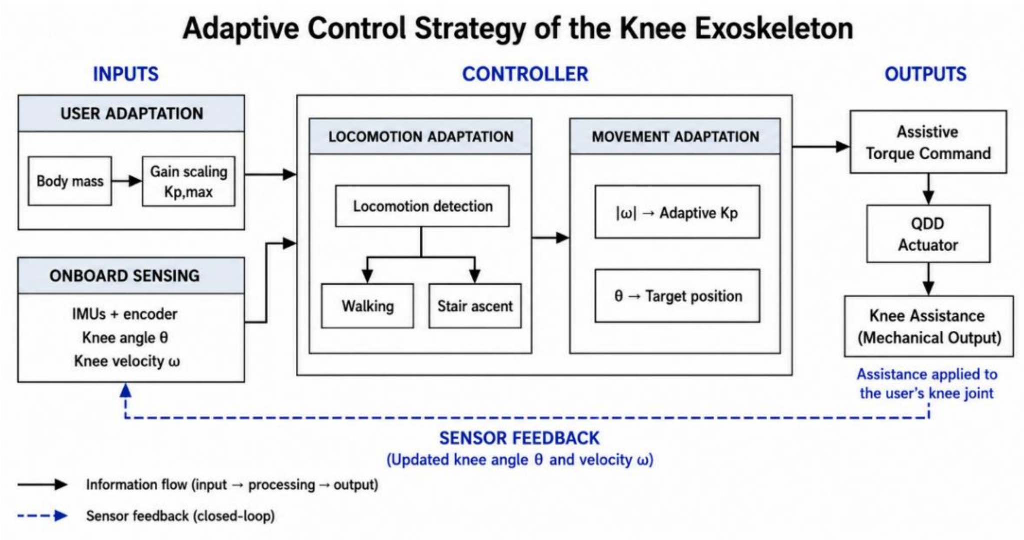

Figure 2 Closed-loop adaptive control architecture for locomotion-dependent knee assistance.

3.3. Experimental Evaluation

Five healthy adults participated in the study after providing informed consent. Each participant performed level walking and stair ascent under slow and fast conditions. A metronome was used to provide target cadences of 66 and 102 steps/min for level walking and 45 and 75 steps/min for stair ascent, and participants were instructed to synchronize their movements with the prescribed rhythm. These target cadences were selected based on the operational capabilities of the developed exoskeleton while providing standardized conditions across participants and trials. Each participant completed all locomotion tasks both without the exoskeleton (No-Exo) and with the exoskeleton (Exo). Participants were familiarized with the device and experimental tasks before data collection, and the task order was randomized.

Lower-body kinematics were recorded using an Xsens inertial motion-capture system [15]. Muscle activity was recorded using wireless surface electromyography (EMG; Cometa Systems, Milan, Italy) at 2000 Hz. The quadriceps femoris vastus medialis (VM), vastus lateralis (VL), biceps femoris (BF), and lateral gastrocnemius (LG) were monitored. Surface electrodes were placed according to the SENIAM recommendations [16]. Following the locomotion trials, participants completed the System Usability Scale (SUS) [17], NASA Task Load Index (NASA-TLX) [18], and an additional questionnaire addressing comfort and fit. Paired Exo–No-Exo comparisons of the primary kinematic outcomes were evaluated using Wilcoxon signed-rank tests, with Cohen's $d_{(z)}$ reported as the paired effect size. Holm correction [19] was applied within each locomotion condition.

4. Results

The experimental results are organized into three parts. Controller behavior is presented first to characterize the assistance generated by the proposed control strategy. This is followed by the biomechanical response, including lower-limb kinematics and muscle activity, and concludes with the subjective evaluation of usability, workload, comfort, fit, and perceived assistance.

4.1. Controller Behaviour

Figure 3 presents a representative example of the relationship between knee angle and commanded torque, while Figure 4 summarizes the normalized assistance profiles across all locomotion conditions. The variability in these profiles is represented by shaded regions corresponding to ±1 standard deviation across repeated assistance events in Figure 3 and across participants in Figure 4. Together, the figures illustrate the assistance patterns generated by the proposed controller during walking and stair ascent.

During walking, the commanded torque increases progressively throughout the assistance period (Figure 3). The normalized assistance profiles (Figure 4) show the same gradual increase, with greater commanded torque during fast walking than during slow walking.

During stair ascent, the controller produces a distinct torque profile (Figure 3). Commanded torque develops earlier during the assistance period and generally reaches higher values than during walking. The normalized assistance profiles (Figure 4) show the same locomotion-specific behaviour, with greater commanded torque during slow stair ascent than during fast stair ascent.

The normalized assistance profiles also exhibit variability throughout the assistance period (Figure 4). Despite this variability, the profiles remain distinct across the four locomotion conditions.

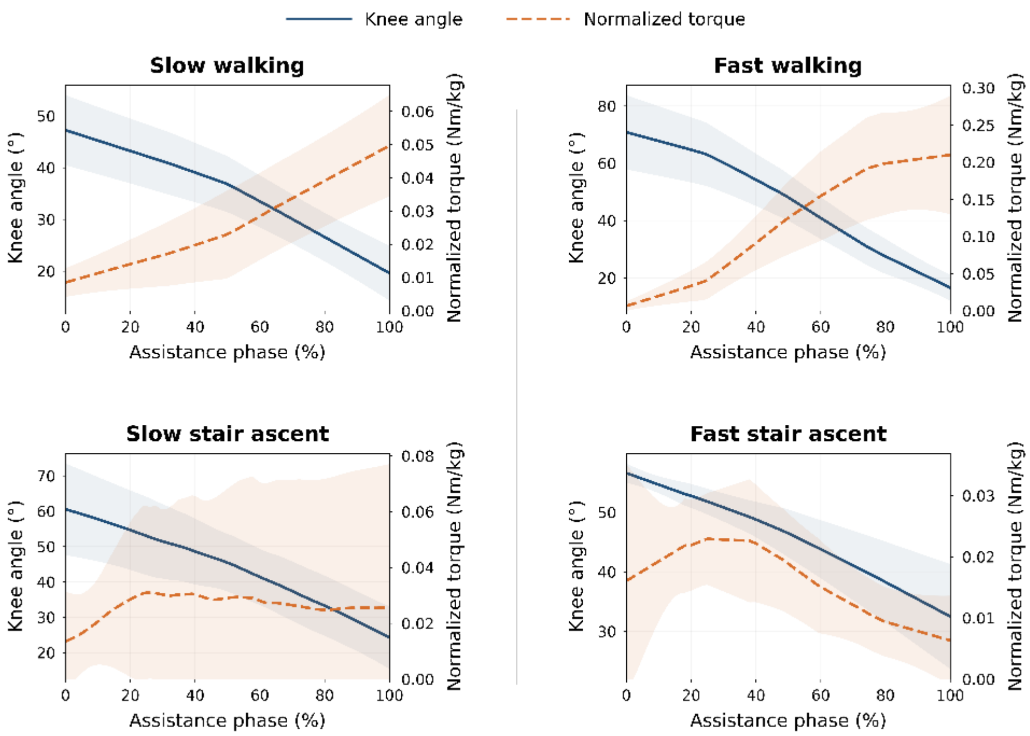

Figure 3 Knee joint angle and body-mass-normalized assistive torque across locomotion conditions.

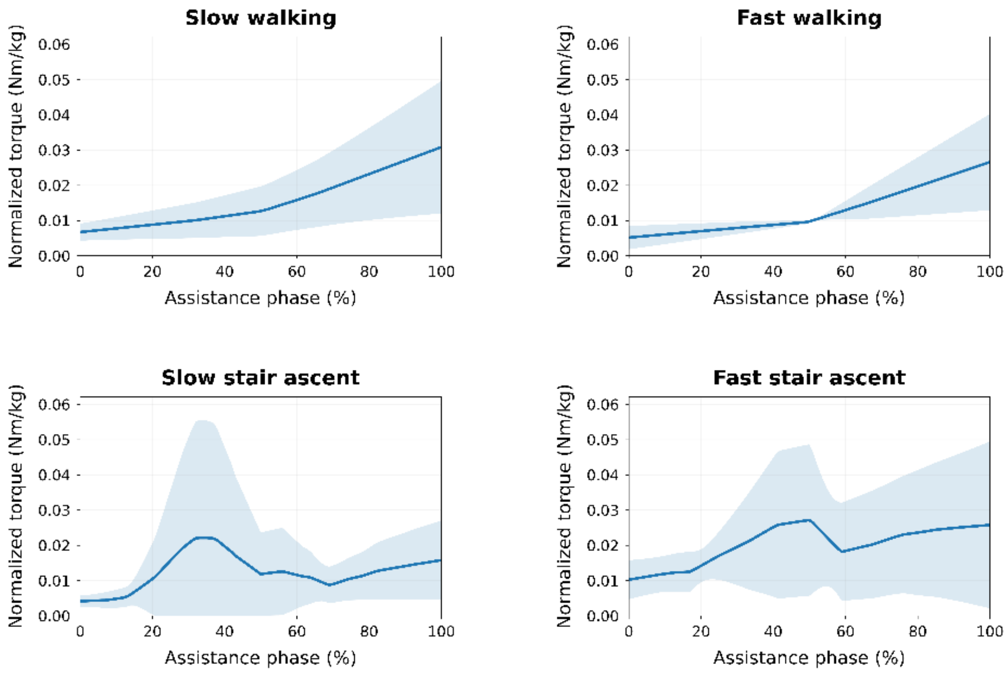

Figure 4 Body-mass-normalized assistive torque profiles across locomotion conditions.

4.2. Biomechanical Response

Biomechanical responses are presented using lower-limb kinematics (Figure 5) and muscle activity (Figure 6). Across all locomotion conditions, the largest biomechanical responses are observed during fast walking, whereas slow walking and stair ascent show smaller or more variable changes. None of the paired comparisons remain statistically significant after Holm correction [19].

Walking and stair ascent show different kinematic responses to the applied assistance (Figure 5). During walking, knee ROM (Figure 5a) changes little during slow walking but decreases by 6.82° during fast walking. Peak flexion velocity (Figure 5b) decreases by 39.6°/s and 86.7°/s during slow and fast walking, respectively. Fast walking also shows a 75.0°/s reduction in extension velocity magnitude (Figure 5c) and a 0.09 s increase in cycle duration (Figure 5d). During stair ascent, peak flexion velocity (Figure 5b) decreases by 37.9°/s during slow ascent and 53.3°/s during fast ascent. Knee ROM (Figure 5a) decreases by 6.35° during slow stair ascent but remains nearly unchanged during fast stair ascent. Cycle duration (Figure 5d) increases by 0.08 s and 0.20 s during slow and fast stair ascent, respectively.
Normalized muscle activity shows condition-dependent reductions across all locomotion tasks (Figure 6). During slow walking (Figure 6a), muscle activity changes little in the vastus medialis and vastus lateralis but decreases by 13.3% in the biceps femoris and 22.9% in the lateral gastrocnemius. During fast walking (Figure 6b), activity decreases across all four muscles, with reductions ranging from 30.8% to 42.2%. During stair ascent, muscle activity also decreases across all four muscles, with reductions ranging from 6.9% to 26.3% during slow stair ascent (Figure 6c) and from 5.5% to 17.8% during fast stair ascent (Figure 6d).

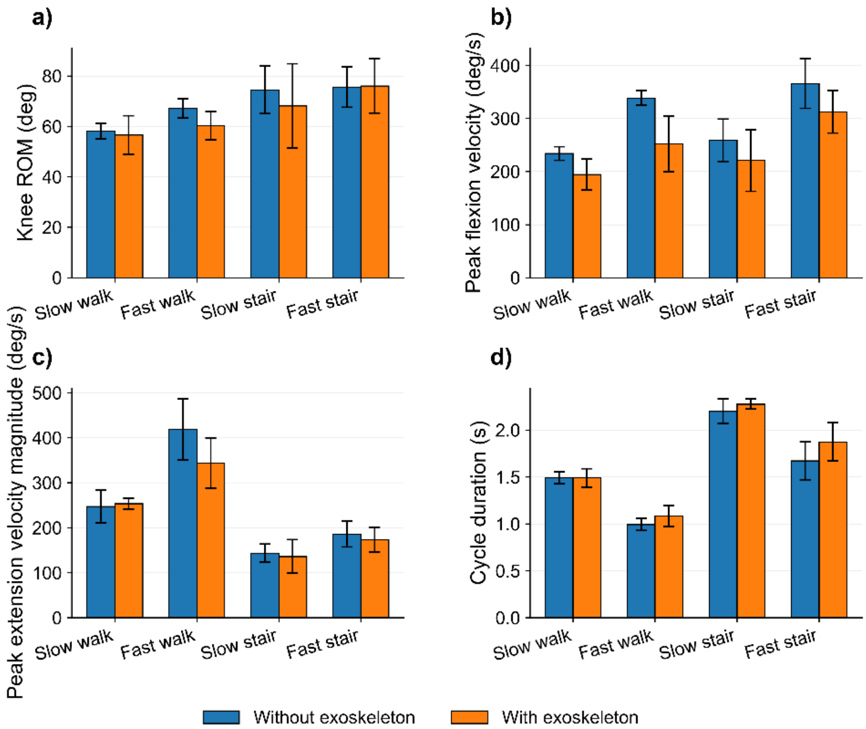

Figure 5 Lower-limb kinematic outcomes during level walking and stair ascent with and without exoskeleton assistance.

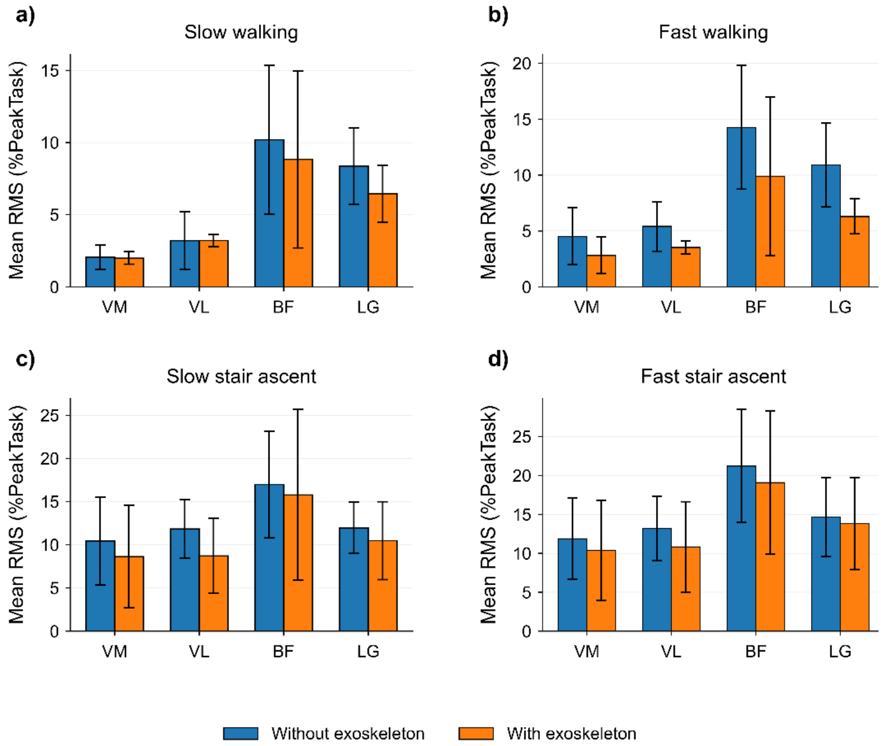

Figure 6 Normalized muscle activity during level walking and stair ascent with and without exoskeleton assistance.

4.3. Subjective Response

Subjective feedback is available from four participants. The mean comfort score is 45 ± 26 on a 100-point scale, corresponding to a moderate to low comfort rating. The majority of participants report difficulties during donning and doffing and identify the upper thigh as the primary location of discomfort. Furthermore, the exoskeleton achieves a mean System Usability Scale (SUS) score of 49 ± 16 on a 100-point scale, indicating below-average perceived usability. Perceived workload is low to moderate, with a NASA-TLX score of 38 ± 17. Physical Demand receives the lowest workload ratings, whereas Mental Demand, Temporal Demand, and Frustration receive the highest ratings.

5. Discussion

The present study investigates whether onboard kinematics and predefined user information can be used to adapt the timing and magnitude of assistance during level walking and stair ascent within a low-complexity embedded control architecture. The controller generates distinct torque profiles for the four locomotion conditions (Figure 3 and Figure 4), demonstrating that assistance is continuously adapted throughout the assistance phase rather than following a predefined trajectory. This continuous modulation of the commanded torque is further illustrated by the normalized assistance profiles (Figure 4).

Slow stair ascent consistently produces higher peak torque than fast stair ascent (Figure 3 and Figure 4). As the proposed controller adapts assistance according to both knee angular velocity and knee flexion, the observed differences likely reflect variations in movement execution between the two conditions rather than locomotion speed alone.

In contrast, both walking conditions exhibit a more gradual increase in commanded torque than stair ascent (Figure 3). This behaviour may indicate delays within the sensing and control pipeline. The effect could be more pronounced during walking because the movement pattern may provide a shorter interval between knee flexion and extension, leaving less time for the controller to generate assistance. Further investigation is required to determine the underlying cause of this behaviour.

The corresponding reductions in muscle activity further suggest that the generated assistance influences muscle activation during locomotion. Although none of the differences reach statistical significance, lower muscle activation is observed across the investigated muscles during exoskeleton-assisted locomotion. The reduced activation of the knee extensor muscles (*m. quadriceps vastus medialis* and *vastus lateralis*) is biomechanically plausible because the exoskeleton provides an external knee extension moment that partially compensates for the internal extension moment generated by the quadriceps [3][4]. The simultaneous reduction in activation of the knee flexors (*m. biceps femoris* and *m. gastrocnemius lateralis*) may indicate reduced agonist-antagonist co-contraction.

The kinematic analysis provides further context for the observed changes in muscle activity. Reduced knee range of motion, lower flexion and extension angular velocities, and longer movement duration are observed during exoskeleton-assisted locomotion. Although these differences do not reach statistical significance, they may contribute to the observed reduction in muscle activity by reducing joint moments and the associated muscular demand. Conversely, the altered kinematics may also represent a consequence of the reduced muscular demand resulting from the applied assistance. As both mechanisms are plausible, the present study does not allow conclusions regarding the causal relationship between the observed kinematic and electromyographic adaptations.

The subjective assessment provides additional context for interpreting the biomechanical findings. Participants report only moderate comfort with considerable inter-individual variability, indicating that the current prototype does not adequately accommodate differences in anthropometry. Most participants identify the upper thigh as the primary source of discomfort, suggesting that the mechanical interface may also influence gait behaviour, consistent with reported challenges in achieving comfortable thigh-cuff interfaces for knee exoskeletons [20][21]. Consequently, the observed kinematic adaptations are likely to reflect both controller-induced assistance and limitations of the current mechanical prototype. The subjective questionnaires are consistent with this interpretation. Participants report moderate comfort and usability despite relatively low physical demand, indicating that further refinement of the mechanical interface is required before the biomechanical effects of the proposed control strategy can be fully isolated.

The present findings highlight the close interaction between controller behaviour, biomechanical response, and mechanical implementation when evaluating wearable assistive systems.

6. Conclusion

The experimental evaluation demonstrates that the proposed controller generates locomotion-dependent assistance using onboard knee kinematics and predefined user information within a low-complexity embedded architecture. Corresponding changes in muscle activation and lower-limb kinematics indicate that the generated assistance influences user movement while also identifying opportunities to further improve assistance timing and overall system performance. Future work should therefore focus on refining the sensing and control pipeline to reduce processing delays and investigate whether additional biomechanical feedback can further improve assistance timing and enable more individualized assistance. In parallel, improvements to the mechanical design should enhance anatomical conformity and user comfort, while broader experimental validation across larger participant cohorts and additional locomotion tasks will further establish the applicability of the proposed control strategy.

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