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User and Environmental Context Adaptive Knee Exoskeleton Assistance using Electromyography

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Abstract— **Proportional myoelectric controller (PMC) has been one of the most common assistance strategies for robotic exoskeletons due to its ability to modulate assistance level directly based on the user's muscle activation. However, existing PMC strategies (static or user-adaptive) scale torque linearly with muscle activation level and fail to address complex and non-linear mapping between muscle activation and joint torque. Furthermore, previously presented adaptive PMC strategies do not allow for environmental changes (such as changes in ground slopes) and modulate the system's assistance level over many steps. In this work, we designed a novel user- and environment-adaptive PMC for a knee exoskeleton that modulates the peak assistance level based on the slope level during locomotion. We recruited nine able-bodied adults to test and compare the effects of three different PMC strategies (static, user-adaptive, and user- and environment-adaptive) on the user's metabolic cost and the knee extensor muscle activation level during load-carriage walking (6.8 kg) in three inclination settings (0°, 4.5°, and 8.5°). The results showed that only the user- and environment-adaptive PMC was effective in significantly reducing user's metabolic cost (5.8% reduction) and the knee extensor muscle activation (19% reduction) during 8.5° incline walking compared to the unpowered condition while other PMCs did not have as large of an effect. This control framework highlights the viability of implementing an assistance paradigm that can dynamically adjust to the user's biological demand, allowing for a more personalized assistance paradigm.**

I. Introduction

Robotic lower-limb exoskeletons hold great promise in improving human mobility in various applications, such as human augmentation for healthy individuals and people with motor impairments [1]. Conventional control approaches for these robotic devices involve providing assistance that mimics human biomechanical joint demands (e.g., providing positive knee joint power during the early stance phase of the gait cycle) [2]. Using this control scheme, several research studies have explored and demonstrated the viability of off-loading the user's biological effort via robotic exoskeleton assistance, such as reducing the user's metabolic cost of walking [3].

Ramp ascent, especially in a steep slope incline walking scenario, is a great example of a locomotor task for which a robotic device can provide a significant benefit. Ramp ascent walking often exhibits a greater magnitude in the user's biological knee joint moment (to bring the body's center of mass upward), leading to an increased knee extensor muscle activation [4]. In this highly demanding walking scenario, a knee exoskeleton can be a potential solution, which can have a direct societal impact in the real world, such as reducing the strenuous workload of soldiers, firefighters, and industry workers face in daily activities [5-7]. Several previous studies have investigated effective control strategies for knee exoskeletons in walking scenarios with significant biomechanical demand, such as loaded walking and ramp ascent [8].

Conventional control strategies for knee exoskeletons utilize state information about the user's walking (e.g., gait event based on the user's joint kinematics) from the user to generate a relevant assistance profile [9]. While this control approach may provide a decent and generic assistance profile, it is static and often fails to accommodate the user's intended changes in the underlying biological joint demand. In contrast, a proportional myoelectric controller (PMC) leverages the user's physiological signal, muscle activation, to determine the assistance envelope, including the onset/offset timing and magnitude, in real-time [10]. This is a promising control strategy because it provides exoskeleton assistance more naturally and seamlessly to the user's movement. Previous research groups have developed different PMC approaches to provide the most effective exoskeleton assistance to the user during locomotion.

The traditional proportional myoelectric controller (static PMC) determines the magnitude of exoskeleton assistance by mapping muscle activation directly to commanded torque. It achieves this by multiplying a static gain by the muscle activation level [11]. However, this static approach has limitations since it cannot capture changes in the user's muscle activation level over time as they adapt to the assistance. For instance, during locomotion, the user's muscle activation level often decreases as they adapt to the exoskeleton. Consequently, the overall assistance magnitude is reduced since the static PMC uses a constant scaling factor [12], requiring manual adjustment. To address this particular issue,

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Koller et. al. previously developed a user-adaptive PMC by allowing the controller to dynamically adapt to the changes in the user's muscle activation level [12]. The user-adaptive PMC comprehends the user's muscle activation level from a previous number of strides to rescale the adaptive gain, achieving a consistent targeted assistance magnitude over a long period of time, regardless of the changes in muscle activation. This works well for steady-state speeds on a treadmill, but it does not perform robustly in more dynamic environments, such as changing speeds and grades, where natural EMG variation and co-adaptation with the device are present.

This current state-of-the-art PMC is shown to be effective in reducing the user's biomechanical demand during a constant environmental context (single slope incline or walking speed). However, there is still a gap in the controller as the system cannot dynamically adjust to different locomotor contexts. This 'static' nature of the current existing control paradigms greatly hinders the translation of the exoskeleton technology to realistic settings such as overground outdoor locomotion where the slope incline may be varying continuously. In this work, we designed a novel user and environmental context adaptive PMC that can adjust the target torque magnitude based on the slope incline level. Furthermore, our control system leverages the user-adaptive PMC scheme where the controller modulates the assistance level while accounting for the user's long-term adaptation (i.e., a decrease in the knee extensor muscle activation level). This flexible and versatile control allows for a seamless and naturalistic assistance profile while maintaining the desired assistance magnitude relevant to the user's peak biological knee joint moment across various inclination levels.

Our central hypothesis for this study is that adjusting the target assistance magnitude that matches more closely to the underlying joint moment when using a PMC will further reduce the metabolic cost of walking compared to a static PMC. Our underlying rationale is that the adaptive control scheme will allow for maintaining optimal assistance levels across various walking conditions (i.e., incline settings). To test this hypothesis, we utilized a novel pneumatically powered soft bilateral robotic knee exoskeleton and tested the effectiveness of knee assistance using three distinct PMCs: static (S-PMC), user-adaptive (U-PMC), and user- and environment-adaptive (UE-PMC). Our research findings provide a meaningful contribution to developing a flexible and robust assistance strategy that directly leverages human biological signals in the control framework, allowing for a more naturalistic movement when using an exoskeleton system.

II. Methods

A. Pneumatic-based knee exoskeleton and system integration

Each knee exoskeleton (Fig. 1) was constructed of a single lateral carbon composite tube, polypropylene proximal anterior thigh and distal anterior tibial shells (4mm thick), and a mechanical knee joint head that provided a leverage mechanism for the power system of four Pneumatic Actuated Muscles (PAMs). The carbon composite round tube uprights (OD-16mm, ID-14mm), served as the structural support frame

Fig. 1. Experimental setup for validating the PMC controllers.

for the exoskeleton. The proximal and distal upright sections attach to a stainless-steel orthotic knee joint head into a circular housing (i.e., clamp) with machine screws.

Power actuated knee extension was achieved with four PAMs, two at the thigh and two at the shank. Each pair of PAMs (i.e., thigh/shank) was attached to a set of proximal and distal articulated levers (thigh and shank respectively) and then to a single cable that rides over a pulley on an anterior outrigger projection of the mechanical knee joint head. Power actuation occurred when compressed air expands the PAMs bladder shortening the length of the cable forcing the thigh and shank sections of the exoskeleton into an extended position. Pneumatic power was provided by a small electric air compressor via a computer-controlled solenoid valves. Both the pneumatics and control system were tethered to the knee exoskeleton system.

To resist distal migration of the exoskeleton on users, another lower upright was attached to the shank upright and extended into the lateral portion of the shoe. Subjects wore conventional running shoes with the exoskeleton. The device is attached to the subject's limb via the anterior thigh and tibial shells with four Velcro closures. The system had some custom fit adjustment features. The thigh and tibial shells provided some medial to lateral flexibility and could be adjusted proximally and distally to accommodate length adjustments.

B. Controller Design

We used a combination of two primary, uni-articular knee extensor muscles (vastus lateralis and vastus medialis) to proportionally control the knee extension assistance magnitude. In real-time, the root-mean-squares (RMS) of raw EMG signals sampled at 1 kHz were calculated with a 100 ms moving window to create the linear envelope of each muscle similar to the previous U-PMC work for a knee exoskeleton by *Lee et al*. [13]. The commanded assistance torque was determined using (1) and (2).

$$G_i = \frac{1}{N}\left(\sum_{j=i-N}^{i-1} g_j\right) \quad (1)$$

$$\tau = G_i \times (RMS\ EMG - threshold) \quad (2)$$

For the S-PMC, G_i, was a static gain multiplied by the RMS EMG signals after subtracting each EMG signal by a threshold

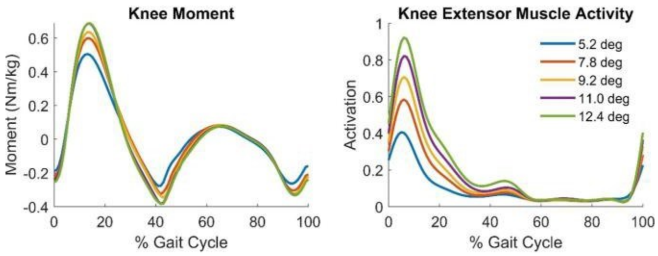

Fig. 2. The profiles of knee joint biological moment and knee extensor muscle activation of able-bodied adults during walking on across various incline levels (5.2°, 7.8°, 9.2°, 11.0°, and 12.4°).

to prevent commanding active assistance torque when the muscle is idling. By doing this, the activation level of the muscle was linearly scaled to the assistance magnitude. For the U-PMC and the UE-PMC, G_i is updated once every gait cycle based on (1) to maintain a consistent level of target maximum torque even if the user's muscle activation decreases in response to the assistance over time. Firstly, the maximum point of the linear envelope of EMG after the threshold subtraction for every stride, g_j, was stored in a buffer keeping the maximum points of the past 25 strides (N=25). Then, the points in the buffer were averaged using (1) to estimate the adaptive-gain necessary, G_i, to map the maximum point of the linear envelope to the maximum target assistance magnitude. For all PMCs, the commanded assistance torques, τ , were computed for the two Vastii muscles separately and were averaged to represent the final commanded torque for the corresponding leg.

For all controllers, the maximum target commanded assistance torque was set to 5 Nm for level-ground walking. The maximum target commanded assistance torque remained set to 5 Nm for all inclination levels for the U-PMC. However, the maximum target commanded assistance torque was modulated based on the user's walking slope under the UE-PMC which allowed the exoskeleton to provide a large amount of assistance when the user's knee joint produces a higher amount of mechanical effort (Fig. 2) [9]: 7.5 Nm for the 4.5° incline walking and 10 Nm for the 8.5° incline walking.

C. Experimental Protocol

To test the effectiveness of the assistance using different PMCs, we performed human-subject testing on nine able-bodied subjects [8 males/ 1 female, 23.4 (mean) ± 3.9 (standard deviation) years, 176.7 ± 8.0 cm, 73.4 ± 10.9 kg]. The metabolic effect of the different PMC assistance strategies (S-PMC, U-PMC, and UE-PMC) was compared to the unpowered condition. The unpowered condition was used as the baseline where the commanded assistance magnitude remained 0 Nm. Under each walking condition, the user walked for twelve minutes evenly distributed into three inclination levels which were level-ground (0°), incline 1 (4.5°), and incline 2 (8.5°) at the user's preferred walking speed (0.6 ~ 1.2 m/s) determined on the 8.5° inclined surface on a treadmill (TuffTread), and this speed remained static across slope conditions. The order of the exoskeleton conditions (unpowered, S-PME, U-PMC, and UE-PMC) was randomized. The user walked wearing the bilateral exoskeletons on the legs, a control backpack, and a 6.8 kg weighted vest to test a load-carrying scenario. For level-ground walking the maximum target commanded assistance torque was set to 5 Nm for all assistance conditions. For the U-PMC controller, the maximum target commanded assistance torque remained at 5 Nm for all levels of inclination. However, under the UE-PMC, the maximum target commanded assistance torque was modulated based on the inclination level: 7.5 Nm for 4.5° incline walking and 10 Nm for 8.5° incline walking. For the S-PMC, the gain, G, for each muscle was determined to map the maximum muscle activation during level-ground walking to 5 Nm before starting data collection. The gains for the U-PMC and UE-PMC were automatically updated as programmed. For all controllers, the commanded torque above 12 Nm was saturated to 12 Nm, commanding 12 Nm, due to a limitation in hardware.

D. Data Acquisition and Analysis

Indirect calorimetry was used to capture the user's metabolic cost (Parvo Medics, UT). The resting metabolic rate was collected before the start of the first walking condition and was subtracted from the gross metabolic rate from the walking trials to represent the net metabolic cost of walking in each condition. One force sensitive resistor was attached to each of the user's heels to detect the heel-contact of the corresponding leg and segment gait cycles. The raw EMG signals from vastus lateralis and vastus medialis were sampled bilaterally at 1000 Hz using a surface EMG signal acquisition system (Biometrics Ltd, VA). The EMG electrodes were placed parallel to the pennation angle of each muscle. and these signals were transmitted to a NI myRIO unit for real-time control and data acquisition purposes. The activation level of one of the primary knee extensors, Vastus Medialis, was compared between walking conditions. For EMG data analysis, the raw signal was band-passed at 20 Hz to 400 Hz, full-wave rectified, and low-pass filtered at 6 Hz to represent the activation profile of the muscle. The peak muscle activation during each gait cycle was averaged for each condition to compare the activation levels between conditions. The data from the last two minutes of each walking trial at each slope were analyzed for comparison. A one-way ANOVA test with Bonferroni correction was performed to find statistical differences in the data acquired from different conditions. The data are presented in mean ± 1 SEM.

III. Results

A. EMG

The primary knee extensor, Vastus Medialis, significantly

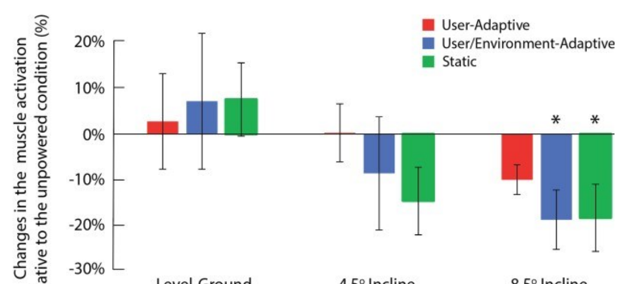

Fig. 3. Average reduction of the Vastus Medialis for each assistance condition compared to the unpowered condition. The error bar represents ± 1 SEM. The asterisk (*) above the bars indicate statistically significant differences compared to the unpowered condition.

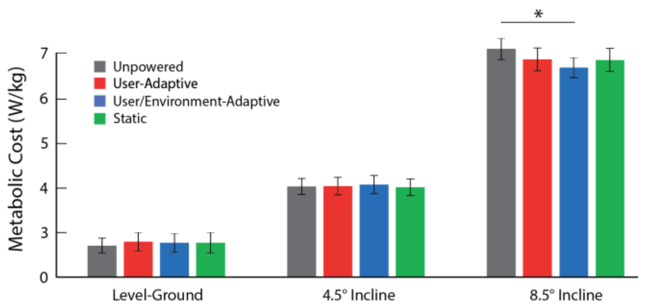

Fig. 4. Averaged net metabolic result across users. The metabolic result was generally similar across walking conditions during all slope conditions. The metabolic cost was significantly lower with the user/environment-adaptive PMC compared to the unpowered condition. The error bar represents ± 1 SEM. The asterisk (*) between the bars indicate statistically significant differences between the assistance conditions.

reduced its activation under the UE-PMC (19.0 ± 5.0% reduction) and the S-PMC (18.5 ± 7.3% reduction) compared to the unpowered condition during the 8.5° incline condition (Fig. 3). However, the reduction of the knee extensor activation level with the U-PMC was not as large and consistent as the UE-PMC and the S- PMC, which seems to be primarily led by the difference in the assistance magnitude. The knee extension assistance during the 4.5° incline condition and the level-ground condition was not as effective as it was during the 8.5° incline condition in reducing the user's muscle effort. This seems primarily due to the positive power generation by the knee joint is more apparent during the early stance phase of the gait cycle during the 8.5° incline condition compared to the 4.5° incline condition and the level-ground condition.

B. Metabolic Cost

During the 8.5° incline condition, the metabolic reduction was largest under the UE-PMC (5.8 ± 3.1% reduction) even though the S-PMC also yielded a similar peak torque and the reduction of the knee extensor activation was similar under the two PMCs (Fig. 4). On the other hand, the peak torque under the U-PMC might have been too low compared to the UE-PMC to yield a larger metabolic reduction during the 8.5° incline condition. There were no significant differences in metabolic cost with different PMCs compared to the unpowered condition for both level-ground and the 4.5° incline condition.

C. Controller

The peak commanded torque was very consistent across PMCs during level-ground waking, which was around 5 Nm (Table 1 and Fig. 5). Across all slope conditions, the U-PMC

TABLE I. AVERAGE (1 STANDARD ERROR OF THE MEAN) PEAK COMMANDED TORQUE ACROSS SUBJECTS UNDER EACH PMC DURING LEVEL-GROUND, 4.5° AND 8.5° INCLINED SURFACE WALKING CONDITIONS.

Slope condition	User-adaptive PMC	User/environment-adaptive PMC	Static PMC
Level-ground	4.9 (0.1)	4.8 (0.3)	5.1 (0.4)
4.5°	4.8 (0.1)	6.8 (0.2)	6.5 (0.6)
8.5°	5.2 (0.3)	9.4 (0.2)	9.6 (0.8)

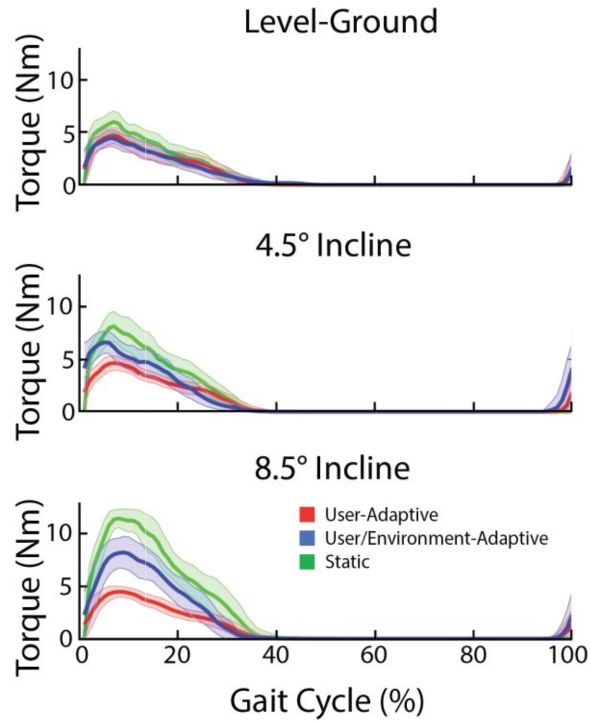

Fig. 5. The commanded torque profiles of a representative subject under the static PMC (green), user-adaptive PMC (red), user- and environment-adaptive PMC (blue) during walking on level ground (top), 4.5 ° incline (middle), and 8.5° incline (bottom). The shaded region represents ± STD.

maintained consistent peak commanded torque as expected since the peak commanded torque was not modulated depending on the slope level. The UE-PMC increased the peak commanded torque with the increase of the slope level. Additionally, the S-PMC also allowed an increase in the peak assistance torque with the increase of the slope level, which was directly caused by the increase in the knee extensor activation level. The trend in the peak commanded torque with the change of slope was very similar between the UE-PMC and the S-PMC. However, the variance in the peak commanded torque with the S-PMC across subjects was almost three to four times larger than the variance with the UE-PMC (Table 1).

IV. DISCUSSION

In general, our robotic knee exoskeleton using an adaptive control strategy (both U-PMC and UE-PMC) was able to effectively off-load the user's biomechanical joint demand by reducing the relevant knee extensor's muscle activation during the 8.5° incline condition. However, only the UE-PMC, which adjusted the target assistance level across different slope conditions and the user's adaptation time, was effective in significantly reducing the user's metabolic cost compared to the unpowered condition ($p < 0.05$).

While the average commanded torque magnitude across subjects was relatively similar between UE-PMC and S-PMC, S-PMC had a greater inter-subject variation (3~4 times greater than the UE-PMC) in the assistance level, indicating that the performance was less consistent and less reliable depending on the subject. This is potentially due to the limitation in S-PMC not being capable of accommodating the user's feedback (changes in the user's muscle activation level during adaptation). Conversely, the UE-PMC was consistent in providing reliable assistance levels for all subjects, which may have contributed to a more consistent behavior across subjects in reducing the user's metabolic cost. This reliable behavior observed in using our UE-PMC supports our initial hypothesis that a control strategy that comprehends the user's underlying biomechanical demand would outperform a static, user-agnostic controller.

During level-ground walking, the three tested PMCs for the knee exoskeleton in this study did not lead to the user's metabolic reduction relative to the unpowered condition whereas other previous PMC studies for assisting the ankle joint did [12, 14, 15]. One of the reasons could be that the magnitude of peak assistance torque for the knee exoskeleton in this study could potentially have been much smaller compared to the previous ankle exoskeleton study (by around six times) [15]. Another reason could be that achieving the user's metabolic reduction by assisting the knee joint seems more challenging during level-ground walking than targeting the ankle joint since human relies much more heavily on the ankle joint for positive power generation than the knee joint during level-ground walking [16, 17]. This was also seen in the previous work by *Franks et al*. where the human-in-the-loop optimization-based assistance yielded more than two times greater metabolic reduction by assisting the ankle joint compared to assisting the knee joint during level-ground walking[18].

During the steepest incline walking condition (8.5°), only the UE-PMC and the S-PMC were effective in significantly reducing the knee extensor muscle activation compared to the unpowered conditions ($p < 0.05$). On the other hand, the U-PMC performance was likely due to the provided assistance magnitude (~5 Nm while the other two controllers had ~10 Nm), which is a natural limitation of U-PMC, namely that it adapts back to the level-walking torque assistance level over time by default regardless of environmental context. Interestingly, this reduction in the user's knee extensor muscle activation did not directly translate to the user's metabolic cost benefit (no significance was shown using the S-PMC). One potential reason for this discrepancy between the muscle activation and metabolic cost is muscle co-contraction (antagonistic muscle group), such as knee flexors (hamstrings). Literature has shown excessive/unreliable exoskeleton assistance can induce instability in the user's gait, leading to muscle co-contraction [3]. The S-PMC adapts the torque level based on muscle activation level, rather than the way the knee typically scales torque in inclines, which is a non-linear function with muscle activation [9]. Our environmental-aware PMC successfully captured this non-linear behavior which may have been a mild benefit to users. In addition, metabolic and EMG results suggest that human augmentation using a knee exoskeleton is more likely to be beneficial for the condition where the knee joint produces a significant level of positive power, the 8.5° incline condition, than the condition where the role of the knee joint is highlighted much less, 4.5° incline and the level-ground conditions.

While our study results showcased the viability of designing a user and environmental context adaptive controller, there are a few limitations that need to be addressed in a future study. For this study, the ground slope incline was manually adjusted using a keyboard input from the experimenter. While this was suitable for an experiment in a laboratory setting, a ground slope incline estimator can be implemented to enable outdoor overground locomotion for the controller to automatically adjust the maximum target torque based on the ground slope. Previously, we developed a deep learning-based slope incline estimator for a knee exoskeleton that can continuously predict the user's slope incline in real-time [19]. By integrating this framework, we can modulate the target assistance torque for our UE-PMC, allowing for a fully autonomous exoskeleton system. Lastly, our study only examined the muscle activation level around the knee extensor muscle group. Thus, future exoskeleton studies should look into muscle groups in other lower-limb joints (e.g., hip extensors and knee flexors). This multi-perspective EMG analysis would allow for a holistic understanding of the relationship between the user's muscle activation level and the user's metabolic response.

V. Conclusion

In this work, we proposed a novel user and environmental context adaptive proportional myoelectric controller that can automatically modulate the exoskeleton assistance level based on the user's EMG signal and slope inclines. We have tested and validated our controller in multiple sloped walking conditions and have showcased the robustness of our approach by reducing both the user's muscle activation and metabolic cost of walking, outperforming the conventional PMC approaches. This control framework highlights the viability of implementing an assistance paradigm that can dynamically adjust to the user's biological demand, allowing for a more personalized assistance paradigm.

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