MASTER THESIS

# THE ONLY WAY IS UP: ACTIVE KNEE EXOSKELETON REDUCES MUSCULAR EFFORT IN QUADRICEPS DURING WEIGHTED STAIR ASCENT


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# Abstract

Firefighters consistently rank stair ascent with gear, which can weigh over 35 kg, as their most demanding activity. Weighted stair climbing requires dynamic motions and large knee torques, which can cause exhaustion in the short term, and overuse injuries in the long term. An active knee exoskeleton could potentially alleviate the burden on the wearer by injecting positive energy at key phases of the gait cycle. Similar devices have reduced the metabolic cost for various locomotion activities in previous studies. However, no information is available on the effect of active knee exoskeletons on muscular effort during prolonged weighted stair ascent. Here we show that our knee exoskeletons reduce the net muscular effort in the lower limbs when ascending several flights of stairs while wearing additional weight. In a task analogous to part of the physical fitness test for firefighters in the US, eight participants climbed stairs for three minutes at a constant pace while wearing a 9.1 kg vest. We compared lower limb muscle activation required to perform the task with and without two bilaterally worn Utah Knee Exoskeletons. We found that bilateral knee assistance reduced average peak quadriceps muscle activation measured through surface electromyography by 32% while reducing overall muscle activity at the quadriceps by 29%. These results suggest that an active knee exoskeleton can lower the overall muscular effort required to ascend stairs while weighted. In turn, this could aid firefighters by preserving energy for fighting fires while also reducing overexertion injuries.

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# 1. Introduction

## 1.1 Research context

This thesis concerns the research performed at the HGN Lab for Bionic Engineering at the University of Utah for my Master's Assignment of the MSc in Biomedical Engineering at the University of Twente. The HGN Lab for Bionic Engineering is specialized in the research and development of assistive wearable devices such as active prosthetics and active orthotics [1], with the latter also being known as active exoskeletons. Previous research from this lab has shown how these exoskeletons can reduce muscle effort for stroke survivors during daily activities like a sit-to-stand transition. However, the HGN Lab for Bionic Engineering envisions that active exoskeletons are not only capable of aiding a patient population in improving their physical capabilities but that these powered devices can also reduce the burden of labor in an occupational context. This thesis provides the first step in realizing that vision by investigating the effectiveness of active knee exoskeletons in reducing muscular effort during the dynamic occupational task of weighted stair ascent.

Several exoskeletons for occupational use are already commercially available, like the IX line of exoskeletons (Ottobock, Duderstadt, Germany) [2]. The currently available commercial devices are mostly passive, redistributing forces spatiotemporally to reduce the burden on critical body parts [3, 4]. However, due to their passive nature, passive exoskeletons can only store, release, or dissipate energy. On the contrary, active exoskeletons have the potential to inject net positive energy into the wearer's motions, reducing the overall energy required from the user [5]. Therefore, it is hypothesized that given an appropriate system design, active exoskeletons have the potential to outperform their passive counterparts in dynamic tasks by virtue of their energy-generating nature.

Although a variety of studies have already evaluated the effectiveness of a broad spectrum of exoskeletons for activities ranging from walking [6] to drilling holes [7], work related to the use of lower-limb exoskeletons for occupational benefit (e.g. tasks involving external weight) is scarce. The few studies on this topic either present mixed results [8] or provide data only on part of the gait cycle [9]. Therefore, this thesis aims to provide a first thorough example of the benefits that lower limb active exoskeletons can provide in physically intensive job activities. In the future, this could lead us one step closer to implementing these assistive devices on relevant job sites, reducing the burden on the United States' ageing and understaffed workforce [10, 11].

## 1.2 Personal contributions

As the work performed for this thesis would not have been possible without the team at the HGN Lab for Bionic Engineering and their continuous advice, it is important to emphasize my contributions to this project to ensure transparency.

I started my time at this lab by finding a suitable occupational use case in which active lower-limb exoskeletons have the most potential to assist wearers. This analysis ranged from understanding the pros and cons of active exoskeletons to analyzing joint biomechanics in relevant movements and unearthing societal bottlenecks caused by physical challenges in various occupations. After a promising use case for active lower limb exoskeletons was found (assistance at the knee for firefighters during weighted stair ascent), the next step concerned the design of an experiment to evaluate the effectiveness of the Utah Knee Exoskeleton in this use case. To this end, I chose an appropriate set of evaluation metrics and a set of sensors that can capture (a proxy of) said metrics. Working with multiple data-capturing systems also necessitated me to design a synchronization protocol that ensured all the different types of captured data could be overlaid and processed. The protocol was refined by performing several test trials guided by a member of the HGN Lab for Bionic Engineering experienced in biomechanical experiments.

Having gotten approval for the experimental protocol from my supervisor, my next contribution revolved around getting the active knee exoskeletons to an operational state for the experiments. This required the two exoskeletons in question to reliably provide high assistive torques for several minutes, something which had not been tried before with these prototypes. Therefore, I made several design iterations for some of the physical interfaces of the exoskeletons, lowering the deformation occurring between the exoskeletons and the wearer while also allowing a wider range of body shapes to fit the device. Moreover, I redesigned the additively manufactured covers of the Utah Knee Exoskeleton to allow for better airflow into the system and easier removal and installation of the exoskeleton's batteries. Furthermore, I was also responsible for fixing any mechanical or electrical issues that sprung up during preliminary testing, at times requiring extensive disassembly and (electrical or mechanical) repair of the devices. I also revamped the existing LabView framework (used to set the controller parameters and record exoskeleton data) to more easily separate variables per controller type. This also included changing the trapezoidal position-based torque profile controller to fit the specific needs of the experiment, like introducing the ability to slowly increase peak torque over time.

With the Utah Knee Exoskeletons sufficiently robust and operational with a controller that could potentially assist in stair climbing, I performed the data collection trials mostly myself with the participants. This process included fitting the Utah Knee Exoskeletons to each participant, training them to climb the stairmill with the exoskeletons while tuning their controller parameters, donning and doffing the participants with the required sensors, and monitoring the various software packages used during the experiments. With the first data collected, I created a data processing pipeline to convert the raw sensory data into meaningful biomechanical metrics. Finally, I discussed the possible interpretations of the processed data with experienced colleagues. With these insights in mind, I took on the task of creating the necessary figures and writing the research paper (and by extension this thesis), forming the capstone of my time at the HGN Lab for Bionic Engineering.

# 2. Research Paper

# The Only Way Is Up: Active Knee Exoskeleton Reduces Muscular Effort in Quadriceps During Weighted Stair Ascent

Vincent S. Boon, Brendon Ortolano, Andrew J. Gunnell, Margaret Meagher, Rosemarie C. Murray, Lukas Gabert, and Tommaso Lenzi, *Member, IEEE*

**Abstract— Firefighters consistently rank stair ascent with gear, which can weigh over 35 kg, as their most demanding activity. Weighted stair climbing requires dynamic motions and large knee torques, which can cause exhaustion in the short term, and overuse injuries in the long term. An active knee exoskeleton could potentially alleviate the burden on the wearer by injecting positive energy at key phases of the gait cycle. Similar devices have reduced the metabolic cost for various locomotion activities in previous studies. However, no information is available on the effect of active knee exoskeletons on muscular effort during prolonged weighted stair ascent. Here we show that our knee exoskeletons reduce the net muscular effort in the lower limbs when ascending several flights of stairs while wearing additional weight. In a task analogous to part of the physical fitness test for firefighters in the US, eight participants climbed stairs for three minutes at a constant pace while wearing a 9.1 kg vest. We compared lower limb muscle activation required to perform the task with and without two bilaterally worn Utah Knee Exoskeletons. We found that bilateral knee assistance reduced average peak quadriceps muscle activation measured through surface electromyography by 32% while reducing overall muscle activity at the quadriceps by 29%. These results suggest that an active knee exoskeleton can lower the overall muscular effort required to ascend stairs while weighted. In turn, this could aid firefighters by preserving energy for fighting fires while also reducing overexertion injuries.**

***Index Terms*— Assistive robots, EMG, stairs, wearable robots**

## I. INTRODUCTION

FIREFIGHTERS rank climbing stairs with their equipment, which weighs around 35 kg, as the most physically challenging part of their job on the firegrounds [1, 2]. This is not surprising, considering that energy expenditure increases by almost 50% when climbing stairs with the additional weight of firefighter gear [3]. Moreover, the average firefighter of 39 years old has an aerobic limit close to the acceptable minimum for their profession [3, 4, 5]. Premature exhaustion can lead to firefighters being unable to complete their tasks at a desired pace [6]. Besides aerobic considerations, over half of all reported non-fatal firefighting injuries are related to overexertion causing muscle-tendon strain and ligament sprains [7]. Considering these challenges, reducing the effort of weighted stair climbing for firefighters can potentially offer benefits for both task completion rates and long-term health.

In stair ascent, the average peak knee torque is more than doubled with respect to level ground walking, being the largest increase in any of the lower limb joint [8, 9]. This torque requirement only increases further with added weight. Therefore, providing assistive power around the knee joint specifically is hypothesized to yield the most benefit in the context of weighted stair ascent. Exoskeletons can offer this assistance by augmenting the wearer's power output around their joints [10]. Whereas passive exoskeletons redistribute the release of energy generated by the wearer spatiotemporally, active exoskeletons can inject net positive energy into the human-robotic system by using actuators [11]. Previous studies have shown reductions in metabolic rates and muscular effort utilizing both types of exoskeletons in various locomotion activities [12, 13, 14], including stair ascent [15]. However, regarding weighted stair ascent, we are not aware of any work that provides insight into the effect of knee exoskeleton use past the analysis of a few strides [16, 17, 18] or only part of the gait cycle [19]. Therefore, although some of these studies indicate that knee exoskeletons can decrease muscular effort at the quadriceps, we are not aware of any conclusive research on the effects of active knee exoskeletons during operational weighted stair ascent.

In this study, we investigate the effect of a bilaterally worn

This paragraph of the first footnote will contain the date on which you submitted your paper for review, which is populated by IEEE This work was supported in part by the National Institute for Occupational Safety and Health under grant 5T42OH008414 and in part by the National Science Foundation under award 2046287. (Corresponding author: Tommaso Lenzi). This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Institutional Review Board of The University of Utah under Protocol No. 00120712, and performed in line with the Declaration of Helsinki.

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active exoskeleton on muscular effort during prolonged weighted stair ascent. Eight participants climb stairs for 3 minutes with a 9.1 kg weighted vest (similar in weight to a firefighter's hose bundle) with and without the exoskeletons. Throughout the trial, both kinematic joint data and EMG data related to the lower limbs are collected. The results of this study offer insights into the potential benefit of using active knee exoskeletons during weighted stair ascent.

## II. Methods

### A. Powered Knee Exoskeleton

In this study, we use two prototypes of the Utah Knee Exoskeleton, shown in Fig. 1. This exoskeleton is characterized by its relatively low weight of 2.6 kg (including battery and physical interfaces) while being able to apply a torque up to 55 Nm, provided by a geared brushless DC motor (EC-22 120W and 3.8:1 gearbox, Maxon Motors, Switzerland). To achieve this high torque density, the knee exoskeleton houses a passive variable transmission (PVT) consisting of a five-bar linkage with an elastic element similar to the drive train described in [20]. In short, this configuration yields a torque-sensitive transmission ratio between the motor and output joint, allowing for a wider range of torque-velocity profiles than would be achievable through a fixed gear ratio.

The DC motor is actuated by a current driver (Everest Core, Ingenia, Spain), which uses an absolute encoder (RM08, RLS, Slovenia) for commutation. Absolute encoders are also housed on the input joint of the elastic element (iC-MU & MU2S, iC-Haus, Germany) and the main output joint (iC-MU & MU18S, iC-Haus, Germany). An inertial measurement unit (IMU, MTi-1, Xsens Technologies B.V., The Netherlands) is attached to the wearer's thigh, yielding the thigh's angle relative to vertical for high-level control purposes. The peripherals are interfaced through two 32-bit microcontrollers (PIC32, Microchip Technology, USA). Finally, an embedded computer (Raspberry Pi 4 Compute Module) is connected to the microcontrollers, which is used for telemetry over WiFi through a graphical user interface.

To efficiently transfer the assistive loads generated by the active knee exoskeleton onto the user, a set of rigid interfaces is used. The thigh cuff, shank interface, and footplate in Fig. 1 exhibit a total of eight adjustable settings. During fitting, these settings can be altered and consecutively fixed to minimize misalignment with the user's biological knee joint without sacrificing the rigidity required for maximal power transfer.

### B. Assistive Controller

The assistance provided by the active knee exoskeleton is dictated by the controller schematically represented in Fig. 3, which has feedforward compensations as well as a state-dependant assistive torque profile. This controller type was chosen over more sophisticated options like neuromechanical model-based control [21], delayed output feedback control [22] or temporal convolutional networks [23] for several reasons. First and foremost, the simplicity of this controller allows for intuitive implementation and tuning while the position-based


**Fig.** 1: A picture of the Utah Knee Exoskeleton, including covers and physical interfaces. A total of eight physically adjustable settings are present to improve fitting for a wider range of wearers.

nature gives the user control over the speed of the activity. By extension, the ease of implementation benefits the reproducibility of the experiments. Secondly, biomechanical data on firefighter stairclimbing is currently scarce. Hence, controllers that rely on simulations or prior data for training are impractical to implement. Thirdly, the use of electromyography as an input for the controller, as necessary for neuromusculoskeletal control, is difficult to justify in an occupational context where it is paramount that assistive devices are as "plug and play" as possible.

The feedforward compensation torque

$$\tau_c = -\left(I\ddot{\theta}_m + b_{PVT}\dot{\theta}_{PVT} + b_m\dot{\theta}_m\right) \qquad (1)$$

lowers the apparent impedance of the exoskeleton to the wearer. $\dot{\theta}_{PVT}$ (rad/s) is the velocity of the elastic element's joint whereas $\dot{\theta}_m, \ddot{\theta}_m$ (rad/s, rad/s$^2$) are the motor velocity and acceleration, used in conjunction with the feedforward inertia $I$ (kgm$^2$) and the PVT- and motor damping terms $b_p, b_m$ (Nm/s). The PVT


**Fig.** 2: Left, a still of a participant on the step mill with the active knee exoskeletons worn bilaterally while walking up the stairs with a 9.1 kg weighted vest. Right, the placement of Xsens (kinematic) sensors and Delsys (surface electromyography) to gather biomechanical data.

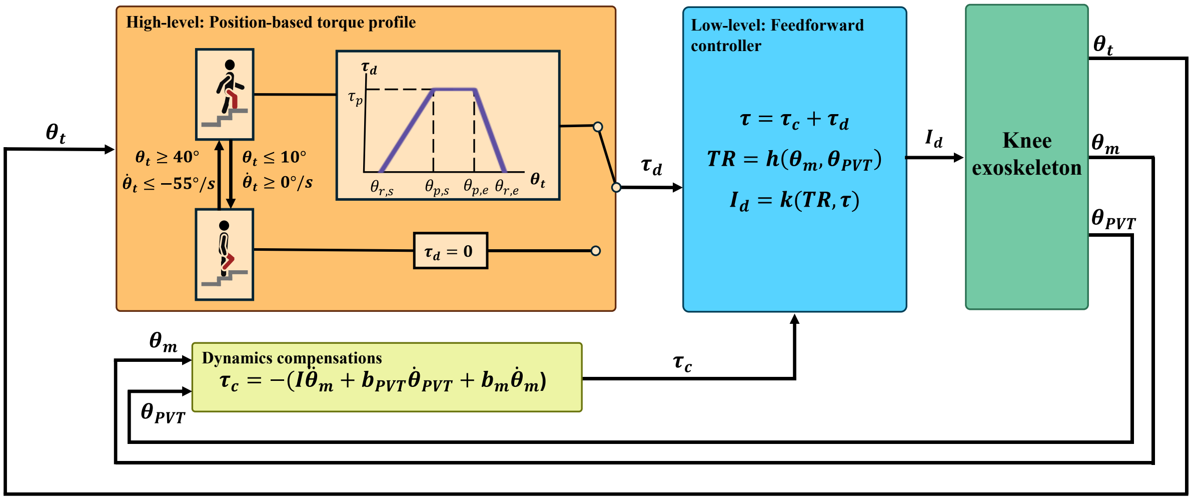


**Fig.** 3: Block diagram of the stair ascent controller. A finite state machine determines if the wearer is in flexion or extension. During extension, a position-based torque profile is employed, while no assistance is given in flexion. Additional impedance compensation torque is provided based on the motion of the motor and passive variable transmission. The (desired) feed-forward current is computed using the instantaneous transmission ratio of the drive train.

damping is set to zero when the elastic element flexes and set to a negative value when extending, which dampens the release of energy stored in the elastic element. The motor damping slope is conditional, based on a speed threshold $|\dot{\theta}_t|$ (rad/s) and described through

$$b_m = \begin{cases} b_{m,1} & \text{if } |\dot{\theta}_m| > \dot{\theta}_t, \\ b_{m,2} & \text{otherwise.} \end{cases} \quad (2)$$

For the assistive torque profile, a state machine splits the wearer's gait into a flexion and extension phase, with the state transition being dependant on the thigh's angular position relative to the vertical ($\theta_t$) and the thigh's angular velocity ($\dot{\theta}_t$). An IMU on the wearer's thigh provides this information. Assistive torque is only applied by the exoskeleton during the extension phase, as nominal biomechanic data of stair ascent indicates little knee torque during flexion [24]. A position-based trapezoidal torque profile determines the desired assistive joint torque $\tau_d$, which also uses the thigh angle obtained from the IMU as an input. The trapezoidal profile is described with five parameters: The start and end angles of the profile $(\theta_{r,s}, \theta_{r,e})$ the start and end angle of the profile's plateau $(\theta_{p,s}, \theta_{p,e})$, and the peak torque $\tau_p$ (Fig. 3). All parameters are heuristically set during training sessions in which participants ascend a stairmill while wearing the Utah Knee Exoskeleton bilaterally. Ideally, support activates after the wearer's leg is ready for weight acceptance. Setting $\theta_{r,e}$ too high causes wearer to be pushed backwards by the extension torque, whereas a late starting angle causes higher quadriceps activation, yielding to lost opportunity for effective support. Similarly, $\theta_{p,e}$ is set such that the assistive torque ramps up sufficiently fast to support the increased weight on the stance leg without jerking said leg. Afterwards, the end of peak torque $\theta_{p,s}$ and the end of support $\theta_{r,s}$ are set such that the wearer is eased into full stance without overextending their knee. Finally, starting from zero, the peak torque is gradually increased during the training session until the wearer notes that the change in torque feels undesirable. The additional weight carried by the participant during the experiment is assumed to proportionally effect the required (assistive) knee torque.

To realize the total desired torque $\tau = \tau_c + \tau_d$, a kinematic model of the drive train is used in combination with a dynamic model of the motor to compute the desired current $I_d$. This desired current is the input to the motor driver's internal closed loop current control. The instantaneous transmission ratio of the drive train is calculated based on the encoder data from both the motor and the elastic element [20]. The low-level microcontrollers perform the abovementioned calculations at a rate of 2 kHz. The relevant controller parameters (trapezoidal profile, dynamic compensations, etc.) can be set via the embedded computer's high-level control loop, which runs at 500 Hz. An engineer's PC can be connected to the embedded computer via WiFi.

The open-loop torque control method described above yields a steady state error of 1.2% in devices with a similar drive train design, which we deem sufficient to omit the need for closed loop torque control [20].

### *C. Participant Information*

A convenience sample of eight able-bodied participants were recruited for this study (64 ± 8.4 kg, 1.72 ± 0.10 m, 23 ± 3 years old, four males, four females). The Institutional Review Board at the University of Utah approved the study protocol (reference number 00120712). The participants provided informed consent to participate in the study, as well as the use of photos and videos from the experiment. The only relevant exclusion criterion for this study based on the convenience sample is an incompatibility with the physical interfaces.

### *D. Experimental Protocol*

Prior to the experiment, participants familiarized themselves with the device during two 45-minute training sessions. During these sessions, the exoskeleton interfaces are adjusted to the user. After familiarization, the participant dons sensors that allow for bio-mechanical data acquisition in preparation for the experimental trials, schematically shown in the right image of Fig. 2.

Surface electromyography (sEMG) is used as a proxy for muscular effort [25]. A total of four wireless sEMG sensors are placed on each leg at the rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL) (Trigno Quattro Sensor, Delsys Inc., United States of America), and gastrocnemius medialis (GM) (Trigno Avanti Sensor, Delsys Inc., United States of America). The ground units of the Trigno Quattro sensor also house an inertial measurement unit (IMU) used for segmentation in post-processing. The SENIAM guidelines [26] are followed during sensor placement and skin preparation. The location of the sEMG sensor at the gastrocnemius medialis is slightly adjusted from these guidelines to eliminate physical contact with the exoskeleton's interfaces, reducing the chance of motion artifacts. The sensors on the upper leg are wrapped in self-adherent cohesive to also minimize motion artifacts from interface deformation. The sEMG data is acquired through EMGWorks Analysis v4.8.0 (Delsys Inc., United States of America).

To measure joint kinematics, participants wore eight wireless inertial motion capture sensors on their lower limbs, pelvis, and sternum (Xsens MVN Awinda, Xsens Technologies B.V., Enschede, The Netherlands), shown in orange in the right image of Fig. 2. A separate computer records the kinematic joint data through MVN Analyze 2020.0 (Xsens Technologies B.V., Enschede, The Netherlands).

After sensor placement, a calibration sequence is performed. A dedicated trigger panel from DelSys synchronized the data acquisition between the capturing systems [27]. The sensory data acquired by the Utah Knee Exoskeleton is recorded separately on the exoskeletons' embedded computers.

The experimental procedure is inspired by the first part of the Candidate Physical Abilities Test (CPAT) for firefighters in the United States [28]. Participants ascend a step mill (StepMill 3, StairMaster, United States of America) for three minutes while wearing a 9.1-kilogram weighted vest at a set speed of one step per second, without using the handrails for weight bearing. The trial is performed twice: Once with the exoskeleton providing assistive torque and once without the exoskeleton. In both trials, participants get a ninety second warmup while the step mill slowly accelerates to one step per second. During this time, the assistive peak torque slowly increases in the exoskeleton trial, aiming to ease the wearer into the assistance to avoid excessive co-contractions [29, 30]. If either knee exoskeleton displays abnormal behavior during a trial (e.g. a faulty state trigger), the trial is prematurely stopped and excluded from further analysis. In between trials, the participants rest for at least thirty minutes. The inertial motion capture sensors are moved and recalibrated between trials to accommodate the exoskeleton interfaces or lack thereof. The order of trials is evenly distributed between participants in an effort to average the effects of fatigue between conditions.

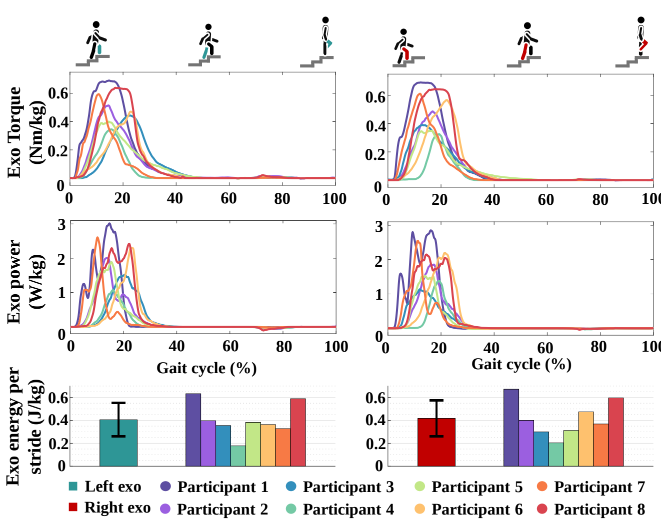


**Fig.** 4: Mean bodyweight normalized exoskeleton torque (top row), power (middle row) and energy generated per stride (bottom row) on the left (left column) and right (right column) side for each participant. Extension is defined as positive. The gait cycle is segmented based on the maximum angle of the thigh per stride.

## *E. Post-processing*

The majority of the data analysis is performed in Matlab R2022b (MathWorks, United States of America). sEMG data are exported using EMGWorks Analyze (Delsys Inc., United States of America), whereas the joint trajectories are exported

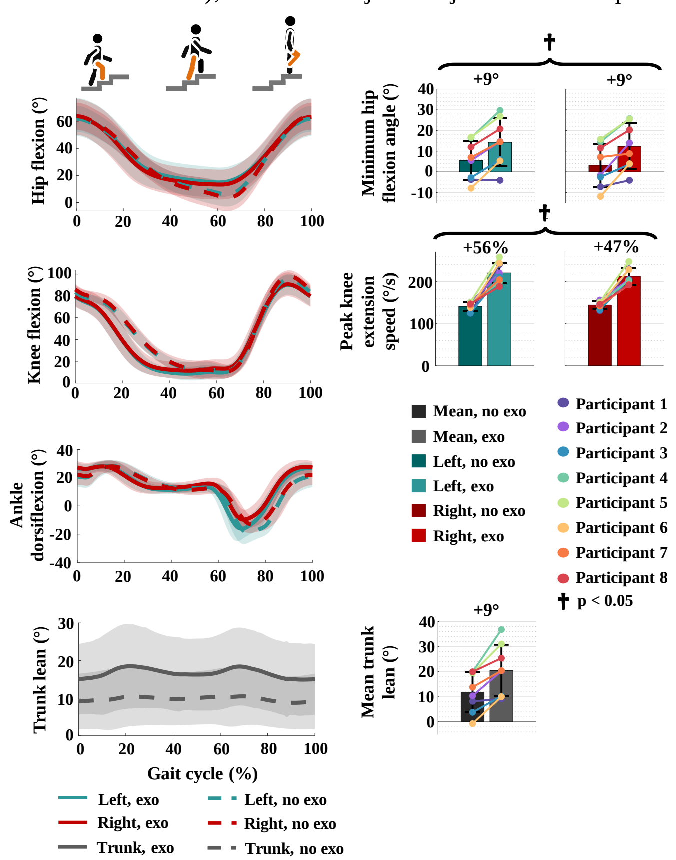


**Fig.** 5: The mean kinematics of the trunk and lower limb joints (left column) plotted over a gait cycle for right leg (red) and left leg (blue) during the exoskeleton trial (solid) and no exoskeleton trial (dashed). The colored shading indicates a range of one standard deviation above and below the mean. Trunk kinematics are segmented based on the strides of the left leg. Some notable mean joint kinematic characteristics are plotted in bar graphs (right columns). Individual results are overlaid with dots. Crosses above the bar plots indicate statistically significant differences between the two testing conditions ($p < 0.05$).

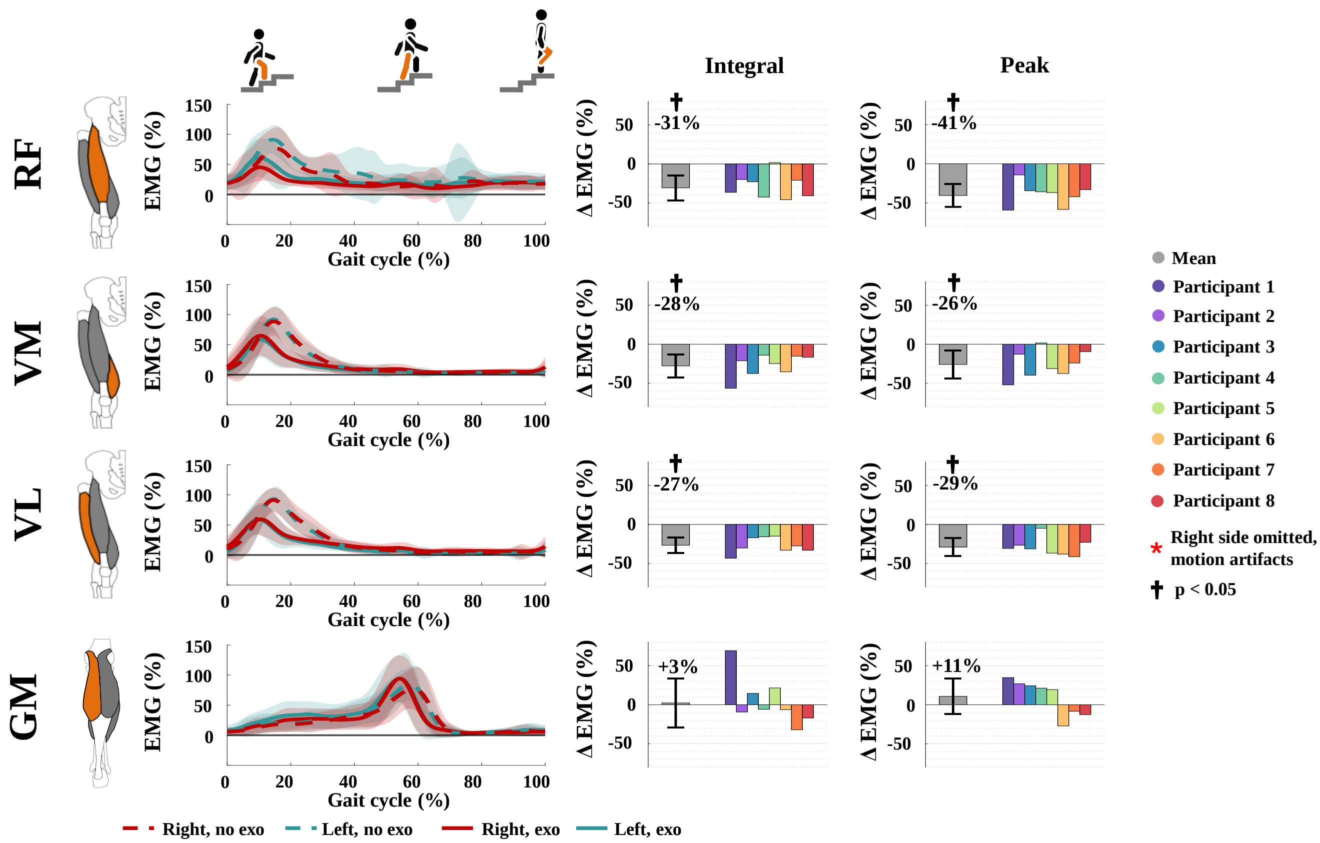


**Fig.** 6: The left column displays the mean muscle activation for the right leg (red) and the left leg (blue) during the exoskeleton trial (solid) and no exoskeleton trial (dashed). The muscles measured include the rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), and gastrocnemius medialis (GM). EMG data is normalized by the mean peak activation during the no-exoskeleton trial. The colored shading indicates a range of one standard deviation above and below the mean. The bar graphs indicate the percentage change between the no-exoskeleton trial and the exoskeleton trial regarding mean peak activation and the mean area under the curve per stride, averaged over both legs. The grey bars are averaged over all participants, with crosses indicating statistical significance ($p < 0.05$).

through Xsens' Matlab library [31] and the exoskeleton data are converted using a custom LabView function (National Instruments Corporation, United States of America). Biological joint kinematics are high-pass filtered at 0.15 Hz using a second order Butterworth filter. Moreover, the sEMG data are first band-pass filtered (20-450 Hz, second order Butterworth filter), rectified, and low-passed using a 5 Hz second order Butterworth filter. Peak activation and area under the curve are computed for each stride, with the latter being a proxy for total muscular effort. The calculated exoskeleton output torque is low-pass filtered with a 100 Hz second order Butterworth filter, while output joint velocity data is low-pass filtered with a 40 Hz second order Butterworth filter. Exoskeleton computed output torque and the output power are bodyweight normalized. All filter orders and frequencies of interest are based on visual inspection of the raw and filtered data in both the temporal- and frequency domain.

After filtering, the data are segmented by stride using the peak thigh angle relative to vertical. Strides are then time normalized and resampled to be 1001 datapoints. Finally, the filtered sEMG data are normalized with respect to the mean peak activation of the respective muscle in the no-exoskeleton condition. Finally, the characteristic metrics of the kinematics and sEMG data between the exoskeleton and no-exoskeleton condition are tested for a statistically significant difference with paired t-tests. The alpha level for all statistical tests is set to $\alpha = 0.05$. To account for the multiple comparisons problem, the alpha level is adjusted for data subsets tested more than once to $\alpha = 0.05/n_{tests}$, with $n_{tests}$ being the number of tests for the particular subset [32].

## III. Results

### A. Exoskeleton kinetics

Fig. 4 displays the output torque and power of both exoskeletons as a function of the gait cycle for each participant. The trapezoidal position-based torque profile manifests as a smoother and more bell curve-like profile in the time domain. The mean peak assistive torque ranged from 0.38 Nm/kg (participant 4) to 0.71 Nm/kg (participant 1) based on tuning, which corresponds to approximately 42% - 79% of the nominal peak biological torque for unweighted stair ascent [24]. On average, the torque profiles (including compensations) start around 4% of the gait cycle and end at 40%.

The mean peak bodyweight normalized power ranged from 1.2 W/kg (participant 4) to 2.9 W/kg (participant 1), which is 56% and 133% of the peak biological knee power during unweighted stair ascent, respectively [24]. As described in Section II, the controller only provided positive power during the stance phase, with negligible power generated during swing.

Exoskeleton power oscillated somewhat during peak assistance, particularly for participants with higher torques.

Finally, the bottom row of Fig. 4 indicates that both exoskeletons generated an average bodyweight normalized energy of approximately 0.4 J/kg over a stride.

### *B. Joint kinematics*

The joint kinematics are presented in Fig. 5. The temporal plots indicate relevant joint kinematics averaged over all participants for each condition. At the knee level, Fig. 5 shows an increased knee extension velocity at the start of gait. Each participant increased their knee extension speed. On average, the knee extended 52% faster ($p = 3.4\ 10^{-13}$, $n_{tests} = 2$), with no significant difference between the two legs (no exoskeleton condition: $p = 0.53$, exoskeleton condition: $p = 0.50$, $n_{tests} = 2$). Moreover, the knee extension phase shortens by approximately 10% of the gait cycle in the exoskeleton condition. On the other hand, more time is spent around the maximum knee extension angle before the leg is lifted to the next step.

Regarding hip kinematics, the most notable difference is that the minimum hip flexion angle decreased by an average of 9° in the exoskeleton condition versus the no-exoskeleton condition, although this metric is significant ($p = 0.02$, $n_{tests} = 2$). As with the knee, there is no significant difference between the left and right side for the minimum hip flexion angle (no exoskeleton: $p = 0.67$, exoskeleton: $p = 0.73$, $n_{tests} = 2$).

Based on the trunk lean data, it appears that participants lean further forward when wearing the exoskeleton during stair ascent by an average of 9°, although this change is not statistically significant ($p = 0.08$, $n_{tests} = 1$).

Finally, the main variation in the ankle kinematics between conditions appears during the swing phase. Here, peak plantarflexion is held for a shorter period of time in the exoskeleton condition while peak dorsiflexion is held longer, from the end of swing until weight acceptance.

### *C. Muscle Activation*

The sEMG data are shown in Fig. 6. The right rectus femoris data of participant 4 and participant 6 are excluded due to substantial motion artifacts. In the gait cycle plots on the left of Fig. 6, a good agreement can be observed between the mean activation patterns of the left and right muscles in the same condition. Temporally, the mean peak of all activation patterns occur roughly 5% earlier in the exoskeleton condition than the no-exoskeleton condition.

When wearing the exoskeleton, the participants reduced activation of all three measured quadriceps muscles. The mean integral of the sEMG signal, a proxy for total muscle activation, significantly decreased during the exoskeleton trial compared to the no-exoskeleton trial (RF: -31%, VM: -28%, VL: -27%, $p < 0.05$). Similarly, the peak sEMG value, analogous to the peak muscle activation, reduced significantly for all quadriceps muscles while participants wore the active knee exoskeletons (RF: -41%, VM: -26%, VL: -29%, $p < 0.05$).

In contrast, the gastrocnemius muscle may activate more when using the exoskeleton. The mean integral gastrocnemius medialis activation increases slightly when participants wear the active knee exoskeletons, although this change is not significant (GM: +3%, $p > 0.05$). Similarly, the peak activation of the GM increased by 11% during the exoskeleton trials with respect to the no-exoskeleton trials, although this change was not significant ($p > 0.05$).

## IV. Discussion

Out of all tasks on the fire ground, firefighters rank stair ascent with protective gear and various tools as the hardest [1, 2]. In this paper, we investigated the effect of bilaterally wearing an active knee exoskeleton on muscular effort during prolonged weighted stair ascent. In this experimental study with eight individuals, we found that the overall muscular effort in the quadriceps decreased by an average of 29% during a three minute weighted stair ascent trial while utilizing two Utah Knee Exoskeletons. Moreover, the mean peak activation of the quadriceps decreased by 32%. These is the first results known to the researchers that indicates how active knee exoskeletons can provide a net decrease in muscular effort for individuals in an occupational setting. By decreasing the overall muscular effort during weighted stair climbing, we hypothesize that firefighters will have more stamina remaining to fight the fire once on location. Furthermore, by decreasing the peak activation in this strenuous activity, the resulting peak load on the lower-limb muscle-tendon units might also decrease. This implies that utilizing an active knee exoskeleton might not only reduce the task effort, but that it can also mitigate the chances of overexertion injuries during weighted stair climbing. Finally, as firefighters in the US are close to the aerobic metabolic limit necessary to perform their duties [3, 4, 5], the metabolic impact of these devices during weighted stair ascent could be investigated in future studies to get a holistic overview of the impact active knee exoskeletons have on this task [32].

Regarding the kinematic results, we observed an increase in maximum knee extension speed by an average 52% when participants were wearing the active knee exoskeletons. While the knee extension phase of gait seems to be reduced, participants spent a longer time with their knee fully extended in stance in return. Since the stairmill constrained the step speed of the participants, we hypothesize that this longer time in full knee extension would disappear when climbing stairs freely, with the overall step speed increasing instead. If this is the case, the active knee exoskeletons can not only reduce the muscular effort required to complete a weighted stair climbing task, but also increase the task completion speed. However, future studies will need to show if an individual's climbing speed increases when wearing active knee exoskeletons by using a free staircase instead of a stairmill.

More closely observing the joint kinematics, the decrease in maximum hip extension angle seems closely related to the increase in the trunk lean angle of participants. Although the exact reason for this behaviour is unknown, we present two potential contributing factors to this phenomenon. First, recall that individuals had only a brief period of training with the exoskeletons. This lack of accommodation time might have caused individuals to more closely focus on their gait by looking at their feet placement on the stairmill, an action which causes one to lean further forward. Second, by leaning further forward participants can potentially increase their stability on the stairs, moving their center of mass further towards the front of the base of support as to not fall backwards when the exoskeletons provide extension torque. In the future, these deviations in kinematics might be prevented by longer training times, minimizing the impact of active knee exoskeletons to the rest of the wearer's body.

Besides these outcomes, it is of interest to note the symmetry of the participants in response to the exoskeleton support, both in terms of kinematics (Fig. 5) and muscular activation patterns (Fig. 6). Intriguingly, the assistive profiles in Fig. 4 indicate that some difference exists between the two exoskeletons in the timing and amplitude of the provided support. This discrepancy between assistive symmetry and biomechanical symmetry might have various causes. It should be noted that each exoskeleton was tuned individually, prioritizing participants' perceived symmetry of support over setting the parameters equal on both controllers. In the end, the two exoskeletons will have slightly different dynamic behavior due to differences in fabrication. Moreover, a slight difference in the placement or orientation of the IMUs used to determine the thigh angle might have also impact the temporal support profile of the exoskeleton during stair ascent. Therefore, where applicable we advise those studying the use of bilaterally worn assistive devices in the future to tune for the symmetry of output metrics rather than symmetry of controller values.

There are several factors in this study that limit the insight that can be provided. In future studies, it should be considered if these limitations can be overcome. For example, none of the eight participants in this study were a firefighter nor had a history with firefighting. Thus, the results of this study might not be fully generalizable to the firefighter population due to variations in age, physical abilities, and potential variations in body shape. Furthermore, a lighter weight of 9.1 kg was chosen to represent the firefighter gear which can weigh around 35 kg, as it was deemed unrealistic to let this participant pool successfully perform a test intended for (fit) firefighters without any notable prior training. A future study that represents the presented use case better could aid in creating a stronger foundation on which the effectiveness of active knee exoskeletons for firefighters can be evaluated.

Besides the limitations of the experimental protocol, the Utah Knee Exoskeleton also offered insight into the electromechanical challenges that are associated with active exoskeletons. Most prominently, the physical interfaces (Fig. 1) exhibited some motion relative to the user, which can reduce the effective assistance provided at the wearer's knee as well as redistributing the provided assistance spatiotemporally [33]. The current interface design also made data recording of some relevant muscles, like the biceps femoris, impractical due to large motion artifacts that would show up when interface deformation occurred. The long head of the biceps femoris works directly antagonistic to the quadriceps, being engaged in both knee flexion and hip extension in stair ascent [24]. Finding a way to include more of these relevant muscles while retaining a high signal-to-noise ratio would thus offer a more complete understanding of the effect active knee exoskeletons have on the wearer. In short, we deem that research into rigid interfaces between exoskeletons and wearer's is vital, as power transfer might become a bottleneck as increasingly more powerful exoskeletons are invented.

In conclusion, this study shows that active knee exoskeletons can reduce the overall- and peak muscular effort of eight individuals performing weighted stair ascent for several minutes. In the future, these assistive devices can potentially reduce the muscular loads on stair ascending firefighters, while increasing stair climbing speed. The results of this study provide the first step in understanding the role active knee exoskeletons can fulfill in assisting during strenuous occupational activities such as weighted stair ascent.

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# 3. Discussion

The work presented above offers a first step in understanding the benefits that active knee exoskeletons could provide to assist firefighters during the toughest part of their job on the fire grounds [12, 13]. It should be emphasized that this research utilizes the Utah Knee Exoskeleton only as an example; Both the device and its controller should not be seen as the penultimate solution to provide assistive power at the knee but rather offer insight into the potential of such devices to realise assistance. Based on these results, I believe we have shown the first convincing use case for active knee exoskeletons in an occupational setting. I hope these results inspire future studies into novel active knee exoskeletons and different types of stair climbing controllers to potentially enlarge the benefits shown in this study, slowly moving towards implementation of these devices there where they can make an impact on society.

This study focussed on assisting the knee joint based on the biomechanical requirements of this joint during stairclimbing. However, other studies have shown that hip exoskeletons are able to reduce metabolic cost during the adjacent activity of unweighted stair climbing [14, 15] with a fraction of the torque supplied in this study. Putting the assistive device's mass more proximal to the body might outweigh the benefits of assisting the joint that has the largest change in required torque with respect to walking, although a conclusive comparison does not yet exist. Nevertheless, in demonstrating that active knee exoskeletons can reduce muscular effort while increasing extension speed in weighted stair climbing, this research shows that knee exoskeletons are not automatically less capable for providing support than hip exoskeletons. Ideally, support is provided to multiple joints in the future, potentially decreasing the wearer's required effort to an extent that is not feasible by supporting a single joint.

During this study, findings showed both spatial and temporal misalignments had an increasingly negative effect on assistance effectiveness at higher torques, with small differences causing large chances in the wearer's response at times. For example, the timing misalignment between the desired biological joint torque and the provided assistive torque is hypothesized to have caused (reflexive) muscle co-contraction. This way, inappropriately timed and scaled torques acted like a disturbance instead of assistance. Similarly, undesired interface constraint forces and torques might have effected the wearer's muscle activation, as interface misalignment is inherent to wearable devices. Furthermore, due to the limited rigidity of the human-robot interface, interface deformation causes spatiotemporal shifting of the supplied assistance [16] and potentially increases joint axis misalignment at higher torques. Indeed, I believe that the necessity of rigid interfaces which can be adjusted to the individual user (or are even customed designed for each user) are an underexposed topic in the research field with considerable implications for the effectiveness of exoskeleton assistance. As exoskeletons are becoming lighter and more power dense, I truly believe that the parasitic interface dynamics can cause a bottleneck for exoskeleton research in the near future.

All in all, this study hopefully serves as a stepping stone for those wanting to improve the world of occupational health and safety through assistive devices, giving a clear indication of the potential benefits of active knee exoskeletons in an occupational context.

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