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Pilot comparison of customized and generalized hip-knee-ankle exoskeleton torque profiles

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Abstract— **Optimized assistance patterns have produced the greatest exoskeleton benefits to energy expenditure of any strategy to date. This strategy may be effective due to the customization of the applied torque profiles to the user as well as the locomotion condition; however, it is currently unclear how sensitive participants are to their unique torque profile. To investigate, we applied previously optimized hip-knee-ankle torque profiles to expert users (N=3; 1.25 m/s; 0 deg incline). The participants walked with the profile optimized to them, the two profiles optimized to the other two participants, and the average of the three torque profiles while we measured their energy expenditure. Relative to walking with the device turned off, on average, participants experienced a 47.5% (range 12%) metabolic reduction when walking with the torque profile optimized to them and a 46% (range 15%) reduction when walking with the other profiles. Interestingly, within-subject performance was more consistent than across subjects (P1: 52% range 5%, P2: 49% range 6%, P3: 39% range 3%) suggesting that, for expert users of some devices, there may be a range of nearly equally effective torque profiles to reduce the metabolic cost of walking. The torque timing was remarkably similar across the four torque profiles while the torque magnitude varied; participants may be much more sensitive to torque timing than torque magnitude, and there may be a set of torque timing parameters that are generally effective.**

I. Introduction

Lower-limb exoskeletons have the potential to augment human locomotion. These devices apply a torque about a joint, typically the hip, knee, and/or ankle, with the intention of assisting the specified task by acting in parallel to the user's musculoskeletal system. Currently, exoskeleton performance is commonly evaluated by its impact on metabolic cost, an easy-to-measure performance metric [1]. While consciously detecting metabolic changes is challenging, where the just noticeable difference is a 25% reduction [2], people adjust their gait in response to small metabolic rate differences, [3]–[5]. Many devices cannot produce noticeable metabolic reductions let alone overcome the energetic penalty of doffing. This may be due to general assistance strategies that are not customized to the user or the locomotion condition.

Metabolic reductions from exoskeleton assistance have increased due to online optimization of the applied torque patterns. Optimized hip-ankle exosuit assistance reduced the metabolic cost of walking by 17% [6]. Bilateral ankle exoskeletons produced a 39% metabolic reduction [7]. Hip-knee-ankle exoskeleton assistance produced a 50% metabolic reduction [8]–[11] in a variety of gait conditions for expert users.

The need for customized exoskeleton assistance profiles will probably depend on the participant population. One solution is unlikely to work equally well for everyone, for example, adults with osteoarthritis will likely need different exoskeleton assistance than young, healthy, adults because of the difference in kinematics and kinetics [12]. Participants within a heterogenous population, such as those who have had a stroke [13], will probably need customized assistance profiles. Exoskeleton customization may be less impactful within a homogeneous population, such as those within a narrow range of age, BMI, or VO2 max.

Human-in-the-loop optimization (HILO) experiments have produced large metabolic reductions [6], [14], [15]; however, they are time-intensive. HILO is a unique problem where both the exoskeleton and the user are simultaneously adapting at different rates. Ideal assistance patterns vary as participants learn to walk with an exoskeleton, and this moving optimum can increase experimental time. The optimization time only increases with device/assistance complexity. Five optimization sessions (109 minutes) were required for naive participants walking with bilateral ankle exoskeletons [7]. At least three optimization sessions (234 minutes) were required for expert users walking with a hip-knee-ankle exoskeleton emulator [8]–[11], while 10 sessions were necessary for a naive participant using the same system [10]. It is currently unclear if HILO is necessary to deliver meaningful benefits to each new user or if this strategy can be used to determine generally effective assistance profiles with a select number of participants.

In this study, we compared the performance of customized exoskeleton assistance profiles and generalized profiles to determine if participants were sensitive to their unique torque pattern. Participants (N=3, experts) walked in a hip-knee-ankle exoskeleton emulator with assistance profiles that were customized to them in previous studies [8], [10]. In addition to the customized profile, participants walked with the two profiles optimized to the other participants as well as the average of the three profiles. Metabolic rate, applied torques, and exoskeleton joint angles were measured. We hypothesized that the four assistance profiles would produce different metabolic reductions, and participants would receive the largest reduction with their custom profile.

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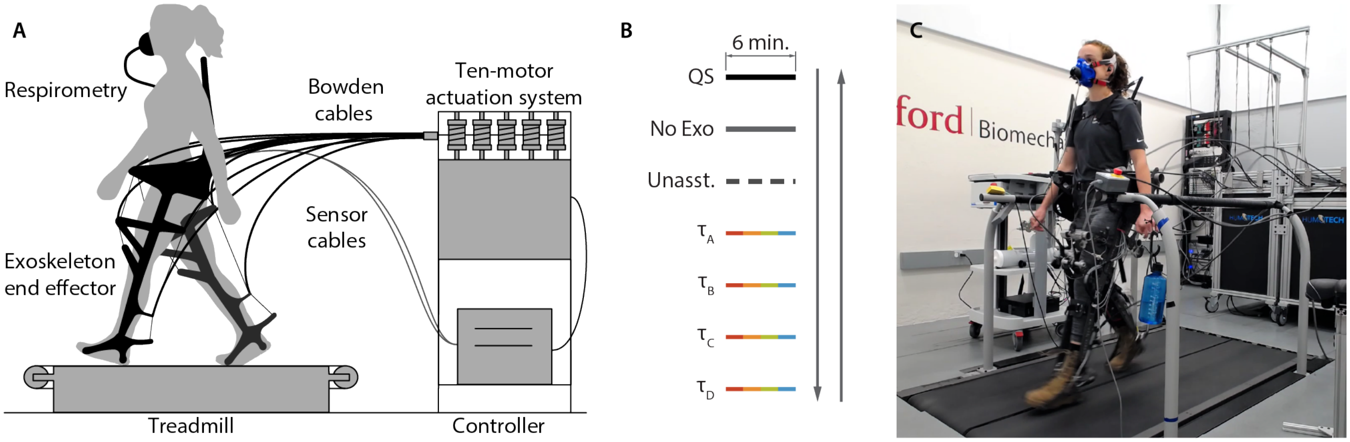

Fig. 1. **Experimental setup.** (A) Exoskeleton emulator system. Powerful, offboard motors actuate the hips, knees, and ankles of the worn end effector through a Bowden cable transmission. (B) Experimental protocol. The participant completed seven unique experimental conditions, each 6 minutes. First, the participant stood quietly, then they walked with no exoskeleton, the exoskeleton turned off, and the four torque conditions in random order. This process was then repeated in the reverse order. (C) Experimental setup. A participant walked on a split-belt treadmill while metabolic rate was measured.

II. Methods

We applied four unique exoskeleton torque profiles to three experienced participants who had previously participated in optimization experiments [8], [10]. Three of the torque profiles were the customized profiles from the optimization experiments, and the fourth was the average of the customized patterns. The average profile was generated by taking the mean of the optimized parameter sets.

Participants (N=3, 1F 2M, age 26-36 years, 60-90 kg, 170-188 cm, expert users) walked on an instrumented treadmill while wearing a hip-knee-ankle exoskeleton emulator [16]. The device has 5 actuated degrees of freedom (hip flexion/extension, knee flexion/extension, and ankle plantarflexion) and 3 passive degrees of freedom (hip ab/adduction, hip internal/external rotation, and dorsiflexion). The participants had previously undergone significant training with this device, which removes the potential for expertise level to confound the results. Two participants completed four optimization studies with this device (P1: 51 sessions; P2: 79 sessions) [8]–[11] and the third completed two optimization studies (P3: 28 sessions) [10], [11]. Both the Stanford University Institutional Review Board and the US Army Medical Research and Materiel Command (USAMRMC) Office of Research Protections approved this experiment. Participants provided written and informed consent.

A. Exoskeleton Hardware

Torque is transmitted from offboard motors to the worn end-effector through a Bowden cable transmission [16] (Fig. 1). The device can be fitted to the participant through length and width adjustability as well as interchangeable boots, and the system results in 13.5 kg of worn mass.

B. Exoskeleton Control

The hip, knee, and ankle profiles were developed in our previous studies [8], [16]. The hip torque profile consists of a period of hip extension during stance and a period of hip flexion during swing (Fig. 2 Hips). Knee assistance has two state-based periods, a virtual spring during stance and a virtual damper during swing, as well as time-based flexion near toe-off (Fig. 2 Knees). Finally, the ankle torque has a period of plantarflexion near toe-off (Fig. 2 Ankles).

The assistance profiles were normalized to body mass and defined by 22 nodes based on our previous work [8], [16] (Fig. 2). Assistance was defined as a function of stride time. Each stride time was measured as the time from one heel strike to the next, and the average stride time was calculated as the weighted average of the previous 20 measurements. The hip profile applied extension torque during heel strike, so to prevent discontinuities at heel strike, the hip profile reset at 84% of stride. The ankle and knee torque profiles reset at 0% of stride.

The applied torque at each exoskeleton joint was measured through load cells and strain gauges. At the ankle joint, strain gauges directly measured applied torque. The torque at the hip and knee was calculated by multiplying the measured cable force from the load cells by the angle-dependent exoskeleton lever arm.

When torque was commanded, the device was controlled through closed-loop, proportional control with iterative learning [17] and velocity compensation [16]. When torque was not desired, the offboard motors tracked the user's joint angle trajectories. To not apply torque caused by friction in the Bowden cable transmission, we added some slack to the system when not applying torque.

C. Experimental Protocol

Participants completed four experimental conditions and two control conditions (Fig. 1 B). The participant first completed the control conditions (no exoskeleton, and unassisted). Then they completed the four torque conditions in a random order (Fig. 3). The participant repeated the seven conditions in reverse order to account for fatigue and ordering effects (ABCDEFFEDCBA). The four torque conditions were the optimized torque from Participant 1 (T1), the optimized torque from Participant 2 (T2), the optimized

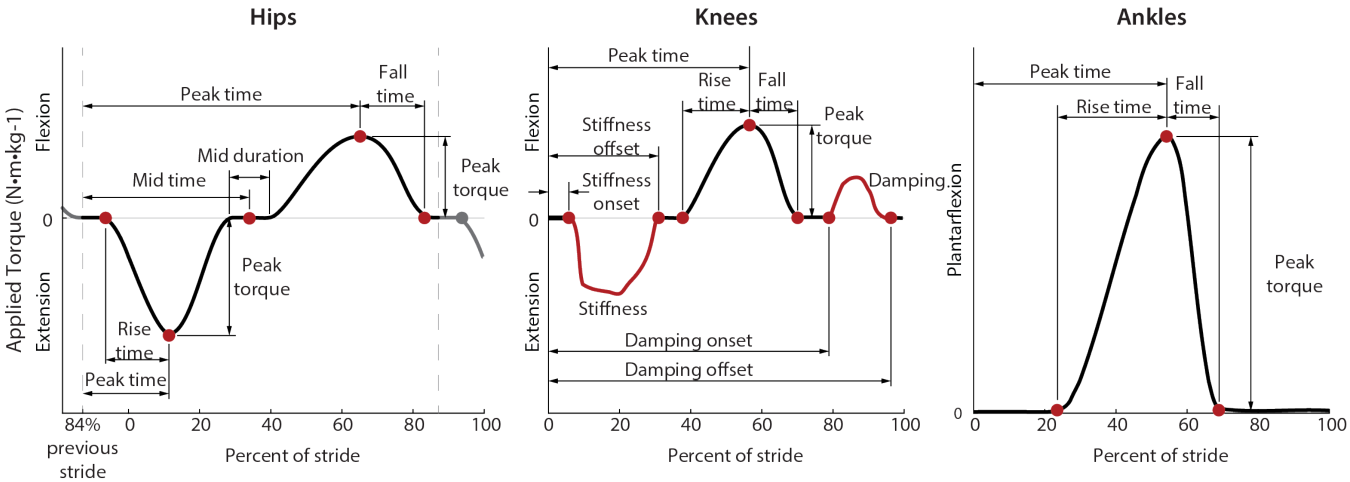

Fig. 2. **Parameterization of the hip, knee, and ankle exoskeleton profiles.** Assistance profiles defined torque as a function of stride time with periods of state-based torque at the knees. Hip assistance is defined by eight parameters, knee assistance by ten parameters, and ankle assistance by four parameters for a total of 22 parameters. The optimization algorithm adjusted the labeled nodes or state variables (red). The hip stride time is reset at 84% of stride to avoid discontinuities in the desired torque profile during heel strike.

torque from Participant 3 (T3), and the torque profile created from the overage of the 22 optimized parameters (Avg.) (Fig. 2). The applied profiles were normalized by body mass.

Quiet standing was collected before and after the experiment to measure baseline metabolic rate. The quiet standing measurement was subtracted from each of the walking conditions to calculate the net metabolic rate. To remove the thermal effect of food, participants fasted for at least four hours before the collection. Each experimental condition was 6 minutes long to measure steady-state metabolic rate. To eliminate transient effects from the previous conditions, participants rested at least 3 minutes before any walking bout. Before the two quiet standing conditions, participants rested for at least 5 minutes to return to baseline.

D. Measured Outcomes

We measured metabolic rate, exoskeleton joint angles, and exoskeleton joint torques during validation experiments. The outcome measurements were averaged over the final 3 minutes of each condition to remove transient behavior.

1) Metabolic Rate: Participant energy expenditure was measured through indirect calorimetry (Quark CPET, COSMED). In a similar fashion to our previous studies [14], [18], [19], we calculated the metabolic rate with a modified Brockway equation [20].

$$E_{met.} = 16.477 \cdot V_{O^2} + 4.484 \cdot V_{CO^2} \qquad (1),$$

Metabolic rate was normalized to participant body mass for accurate comparison. This experiment was completed during the COVID-19 pandemic, so participants wore an additional cloth or paper mask under the collection mask following safety procedures. In previous experiments, we evaluated the impact of the cloth or paper mask and found that it slightly lowers the measurement by a similar offset across experimental conditions [8]–[11]. The metabolic results were calculated as percent changes from the control conditions which minimizes the effect of the cloth or paper mask.

2) Exoskeleton Joint Angles: We estimated user kinematics through the measured exoskeleton joint angles. The exoskeleton hip, knee, and ankle joint angles were directly measured with rotary encoders, which reasonably approximated user joint angles. Measuring kinematics with motion capture systems is challenging and inaccurate with our set up because of marker occlusion and ghost markers caused by reflective elements on the device.

3) Exoskeleton Joint Power Calculation: To calculate the exoskeleton joint power, we multiplied the calculated joint velocity by the measured torque. Exoskeleton joint velocity was calculated by taking the time derivative of the measured joint angle, then low-pass filtered at 50 Hz.

$$P_{j,exo} = \tau_{j,exo} \cdot \dot{\theta}_{j,calc} \qquad (2),$$

To calculate positive applied power, we averaged the calculated power when greater than zero. The joint power was summed between the two legs, and the total power was the sum of the joint powers.

4) Significance Analysis: A functionally meaningful difference in metabolic cost while walking is 10% [21], and the standard deviation in metabolic cost while walking on level ground is 0.4 W/kg [22]. For each participant, we compared whether there were measurable differences between the four torque conditions, as well as between the control conditions and the assistance conditions.

III. Results

A. Metabolic Reductions

On average, participants experienced a 47.5% ± 6.0% metabolic reduction when walking with the torque profile optimized to them relative to walking with the device turned off and a 46% ± 6.2% reduction when walking with the other profiles (Fig. 4). The assistance profiles produced similar metabolic changes within participants, and the average torque profile produced nearly identical metabolic reductions to

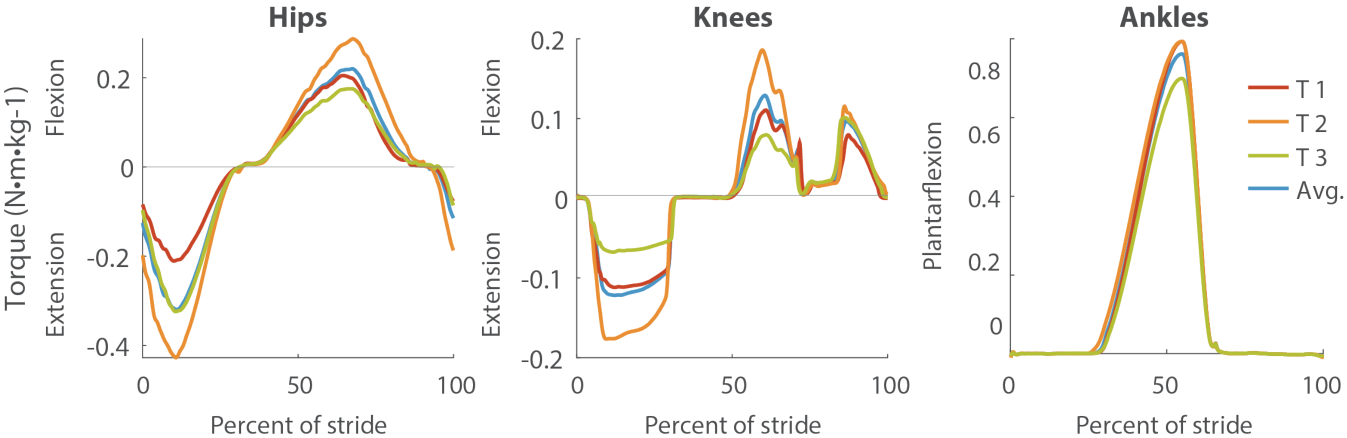

Fig. 3. **Applied hip, knee, and ankle torque profiles.** The torque profiles optimized for Participant 1 (T1, red), the profiles optimized for Participant 2 (T2, orange), the torque profiles optimized for Participant 3 (T3, orange), and the profile generated from the average optimized parameters (Avg., blue) are shown. The profiles are applied as a percentage of stride and normalized by body mass.

the customized profile for each participant. Exoskeleton assistance reduced the metabolic cost of walking relative to unassisted walking by 52% for Participant 1 (custom profile: 53%, average profile: 53%, range 48%-53%), 49% for Participant 2 (custom profile: 49%, average profile: 51%, range 45%-51%) and 39% for Participant 3 (custom profile: 41%, average profile: 38% range 38%-41%). These percent reductions correspond to a 2.32 W/kg reduction for participant 1 (range: 2.15 - 2.39 W/kg), a 1.13 W/kg reduction for participant 2 (range: 1.04 - 1.18 W/kg) and a 1.64 W/kg for participant 3 (range: 1.60 - 1.71 W/kg). When compared to walking with no exoskeleton, assistance reduced the metabolic cost of walking by 30% for participant 1 (range 25% - 32%), 41%for participant 2 (range 36% - 43%), and 24% for participant 3 (range 23% - 27%). The range of metabolic rates across the torque profiles was less than the clinically relevant metabolic difference (10%).

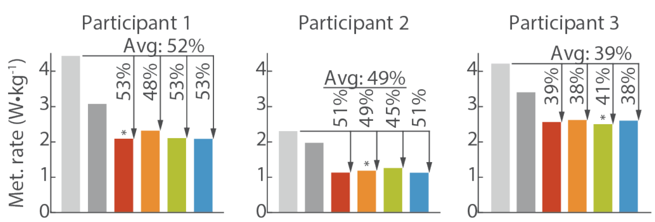

Fig. 4. **Metabolic cost results.** The metabolic cost of walking with no torque applied (Unasst., light gray), walking without the device (No exo, gray.), walking with the torque profile optimized to Participant 1 (T1, red), walking with the torque profile optimized to Participant 2 (T2, orange), walking with the torque profile optimized to Participant 3 (T3, green), and walking with the average torque profile (Avg., blue). The bar corresponding to the participant's customized profile is shown with a star. The four torque profiles reduced the metabolic cost of walking relative to both the unassisted condition and the no-device condition for all three participants. Quiet standing has been subtracted from all scores.

B. Joint angles

For all participants, joint angle trajectories varied from the unassisted condition when torque was applied (Fig. 5). However, there were no noticeable changes between torque conditions within participants.

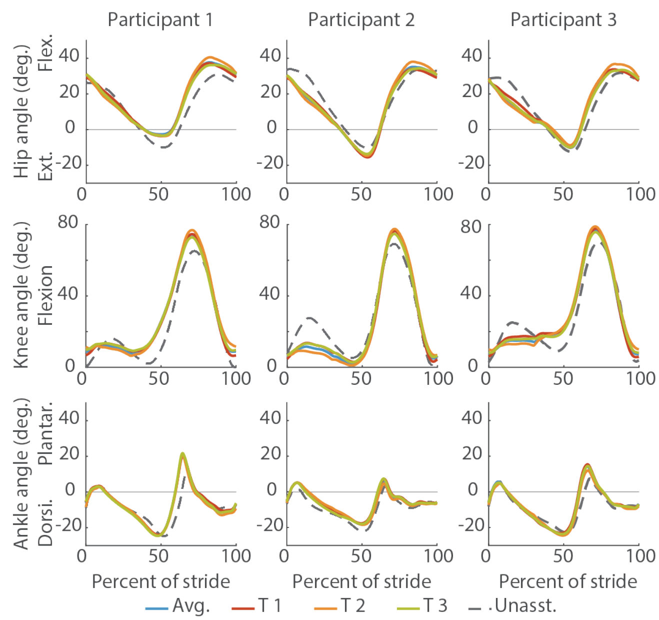

Fig. 5. **Average hip joint angle trajectory for each participant and exoskeleton condition.** Assisted (solid) and unassisted (dashed) The joint angle trajectories are shown for the four torque profiles (T1 (red), T2 (orange), T3 (green), and Avg. (blue)). The joint angle trajectory of the unassisted condition (walking with no torque applied) is shown by the dashed gray line. Joint angle trajectories changed with assistance relative to walking unassisted, but the changes were consistent across the torque profiles within a participant.

C. Exoskeleton Joint Power

The four assistance profiles resulted in different amounts of positive exoskeleton power (P1: 1.08 - 1.52 W/kg, P2: 0.81 - 1.20 W/kg, P3: 0.92 - 1.38 W/kg) while negative exoskeleton power was more consistent within a participant (P1: -0.27 - -0.32, P2: -0.19 - -0.26, P3: -0.22 - 0.29)(Fig. 6). Ankle torque produced similar positive power levels for all four assistance profiles, and the positive power at the hips varied. Knee and ankle torque resulted in low levels of negative power that was consistent within a participant.

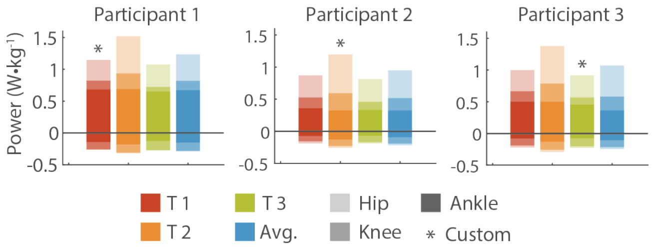

Fig. 6. **Applied exoskeleton power.** The total positive and negative exoskeleton power from the four torque profiles (T1 red, T2 orange, T3 green, and Avg. blue) for each participant. Power contributions from the hips (light), knees (medium), and ankles (dark) are shown, and the profile customized to the participant is denoted with a star. The power was summed across both legs.

IV. Discussion

We hypothesized that participants would experience different metabolic reductions with the four torque profiles, but the results of this study show that, for expert users with similar age and fitness levels, both customized and generalized exoskeleton torque profiles produced similar metabolic reductions. In addition, there was no measurable difference when a participant was walking with their optimized torque pattern as opposed to walking with any of the other three torque patterns. The consistent metabolic reductions within participants suggests that, for this homogeneous population, people are not sensitive to their customized profile, and there may be a range of torque profiles that are similarly effective.

Participants may be more sensitive to specific assistance profile features. The applied torque profiles had similar timing parameters while the torque magnitude varied. Some magnitude parameters for one profile were at least double those for another profile. For example, hip extension torque magnitude for T2 relative to T1. In addition, the torque profiles applied different amounts of power to the user while walking, which suggests that applied power does not directly correlate with metabolic reduction. There may be a range of torque profiles and applied power that produce large metabolic reductions for young, able-bodied participants. Too little power may not be noticeable, and too much power may be overwhelming. The profiles applied in this study may be in a sweet spot in between the two extremes. The torque timing parameters were consistent across the applied torque profiles, so participants may be more sensitive to torque timing parameters than other facets of the assistance profiles.

For this population, the metabolic rate cost function landscape may have a flat optimum, where there may be a variety of similarly effective parameters. The optimization process may produce a range of useful assistance profiles. However, arbitrarily setting assistance parameters has no guarantee of being equally effective, and a systematic approach, such as optimization, is needed to determine nodes within the flat optimum. The cost landscape may vary for different participant populations, cost functions (balance, speed, comfort, etc.), and/or gait conditions.

Optimizing the torque profile to the population and gait condition may be key factors in determining effective assistance. The profiles applied in this study were optimized for walking at 1.25 m/s on level ground. However, exoskeleton assistance customized to other walking conditions, such as uphill walking, may not be as effective. During uphill walking, optimized torque magnitudes for hip and knee extension were roughly double the largest torque magnitudes seen in this study [9]. While this study did not see metabolic changes with assistance profiles that had large torque magnitude differences, there may be a point at which the discomfort of large torque profiles overcomes the metabolic benefits. In addition, there is no guarantee that the torque profiles that are effective for young, healthy adults will be effective for other populations, for example, older adults with chronic stroke. Also, exoskeleton torque customization may be especially beneficial for heterogeneous populations.

This study could be strengthened by evaluating participant preference as well as increasing sample size. While we did not measure physiological changes within a participant from different assistance profiles, there may have been preference differences. Anecdotally, participants mentioned that some assistance patterns felt like marching while the kinematic effect of others was not as noticeable. Systematically evaluating participant preference could give greater insight into the differences between the torque profiles. In addition, performing this experiment on additional participants would improve statistical power and has the potential to demonstrate trends we did not see in such a small participant pool. However, we were limited by experimenting on only expert participants, which we chose to account for confounding factors from learning to use an exoskeleton [7]. The protocol also required that participants had already completed exoskeleton torque optimization experiments, which are typically at least 40 hours of data collection time per participant. Even with the small sample size, the difference in metabolic reductions within each participant was less than the functionally measurable difference (10%) [21] and the standard deviation of metabolic rate while walking (0.4 W/kg) [22].

This study has the potential to inform future exoskeleton products as well as research. The consistent metabolic reductions from different torque profiles suggests that devices for homogeneous populations might not need to be customized to deliver large metabolic benefits. Future implementations of HILO could penalize unnecessarily large torque magnitudes (for example, by minimizing the combined human and exoskeleton energy expenditure), which could be beneficial for mobile device development. Exoskeleton products could provide a baseline assistance profile that is generally useful, allow users to fine-tune to their preference [23], and still produce effective torque patterns. However, we should confirm this assessment with additional exoskeleton devices as well as a wider variety of participant populations. Future research could determine how sensitive additional populations are to customized assistance, for example, do the torque profiles optimized for young, healthy adults translate to older adults or those who have had a stroke?

V. Conclusion

In this study, we applied previously optimized hip-knee-ankle torque profiles to three expert users. The participants walked with the profile optimized to them, the two profiles optimized to the other two participants, and the average of the three torque profiles while we measured their energy expenditure. We hypothesized that participants would receive the largest metabolic reduction with the profile customized to them and that the four torque profiles would produce a variety of metabolic reductions. However, participants experienced consistent metabolic reductions across the four profiles. The torque profiles had similar timing parameters and a wide variety of torque levels, which suggests that torque timing may matter more than torque level. In addition, mechanical power is not sufficient to predict metabolic rate where participants received different levels of mechanical power from the four profiles but experienced similar metabolic reductions. There may be a large set of equivalently effective profiles within a given population, and the set may be larger for more homogeneous populations, like the one tested here, than for heterogeneous populations. This study showed that there appears to be generally effective exoskeleton assistance strategies for homogeneous populations. Future studies could evaluate customization in novel tasks, such as stair climbing or manual material handling tasks, or with heterogeneous participant populations.

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