Supporting Document for Studied Specific Cases

We conducted four different case studies to gain more insight into the performance of assistive devices that include studying both assistive devices in a particular load condition or an assistive device in two different load conditions with the same effect on the metabolic power expenditure or the same power consumption of assistive actuators. Investigating these specific configurations of the optimal devices helped us to understand how the profiles of devices with the same performances change in a load condition more systematically. These cases can also help us to gain insight into the effect of load condition on assistive device profiles, and clarify how a particular device can be affected by loading assisted subjects with a heavy load.

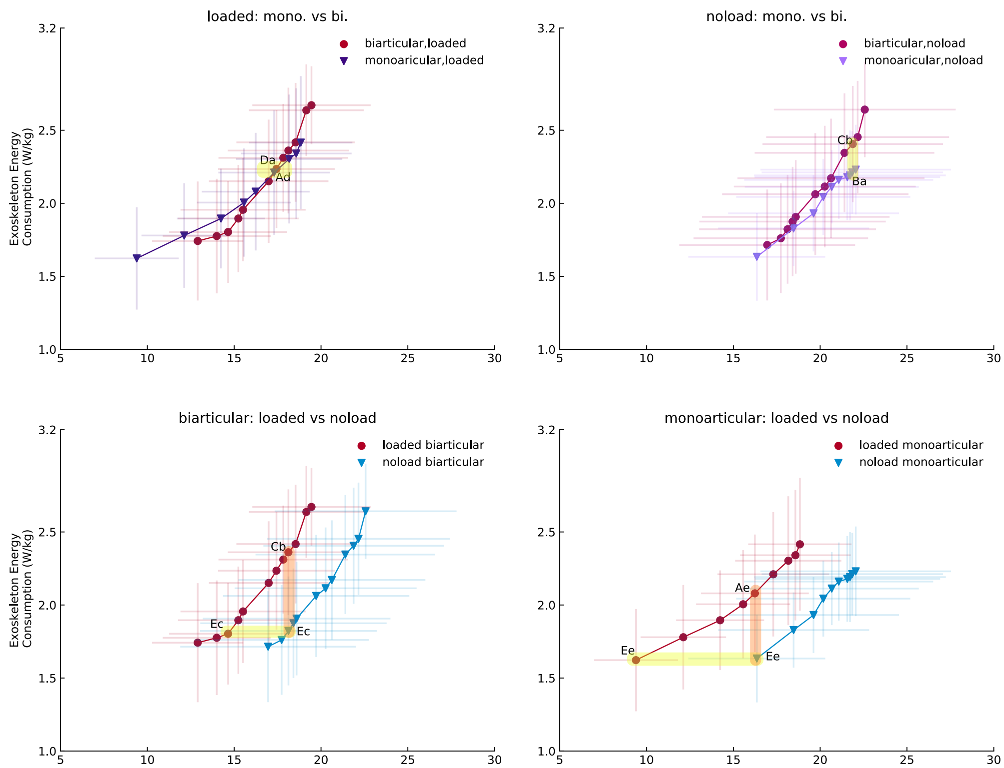


Fig C1. Studied cases chosen from the Pareto front curves. The chosen optimal solutions of each configuration in each of both load conditions used to be studied. The selected solutions on Pareto front curves are resulted from averaging over 7 subjects and 3 trials.

Case 1: Devices Performance in *Loaded* Condition

It was shown that the optimal trade-offs of both exoskeletons are practically the same in the *loaded* condition. To study the performance of each actuator of the devices and their effect on the muscle activity of assisted subjects, we selected two devices on the Pareto front that had nearly the same performance in both metabolic cost reduction and power consumption. The chosen device for the monoarticular exoskeleton has 70 N-m hip peak torque and 40 N-m knee peak torque, which is represented by "Ad" on the Pareto front (Figure C1), and the peak torques of the biarticular device are 40 and 70 N-m on the hip and knee actuators, respectively, represented by "Da " on the Pareto front. As can be inferred from the configurations of devices, although these two chosen devices have the same performance on defined objectives, they have a completely different arrangement on hip and knee actuators. The metabolic rate of assisted

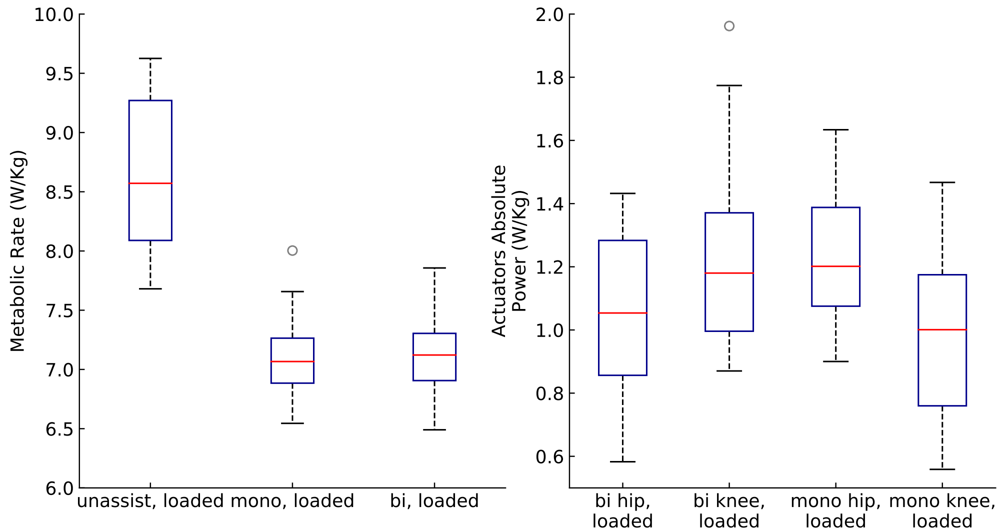


Fig C2. Assistive devices power consumption and their effect on the metabolic rate. The power consumptions of assistive devices and their effect on whole-body metabolic rate of the subjects walking at self-selected speed while carrying a heavy load. Asterisks indicate statistically significant differences (7 subjects, 3 trails, Tukey Post-hoc, $P < 0.05$).

subjects with biarticular and monoarticular devices shows no significant difference between the level of assistance delivered by these devices, which was expected from the Pareto front. While the total power consumptions of these two devices were practically identical on the Pareto front, the power consumptions of the hip and knee actuators between the biarticular and monoarticular devices were not identical, as represented in Figure C2. This dissimilarity indicates that while the devices deliver the same assistance to the subjects carrying a heavy load, their assistance strategies are different; this claim can be more illuminated by analyzing the profiles of actuators.

Although the selected monoarticular and biarticular exoskeletons had the same effect on the metabolic rate of *loaded* subjects, it was shown that their assistive actuator configurations were different, resulting in different power consumptions of the actuators. This variation in the configuration of their actuators affects their mechanical design, especially their required gear train and reflected inertia. We employed the developed modified augmentation factor to assess the performance of the biarticular and monoarticular exoskeletons under the effect of device inertial properties. The computed modified augmentation factors for the monoarticular and biarticular devices were 0.66 ± 1.00 and 1.98 ± 0.71 W/kg, respectively, which indicates that both exoskeletons would be able to deliver assistance to the subjects even under the inertial properties of the devices, causing a greater metabolic burden on the subject. Additionally, the MAF values show that the biarticular device had superior performance to the monoarticular exoskeleton, and the reason for this is rooted in the mass distribution and gear train of the monoarticular device. The inertia calculations show that the monoarticular device had nearly three times more inertia on the thigh than the biarticular device, and according to the inertia location factor of the thigh, the effect of inertia on the thigh is expensive in terms of the metabolic rate increase, which results in a lower MA factor for the monoarticular exoskeleton.

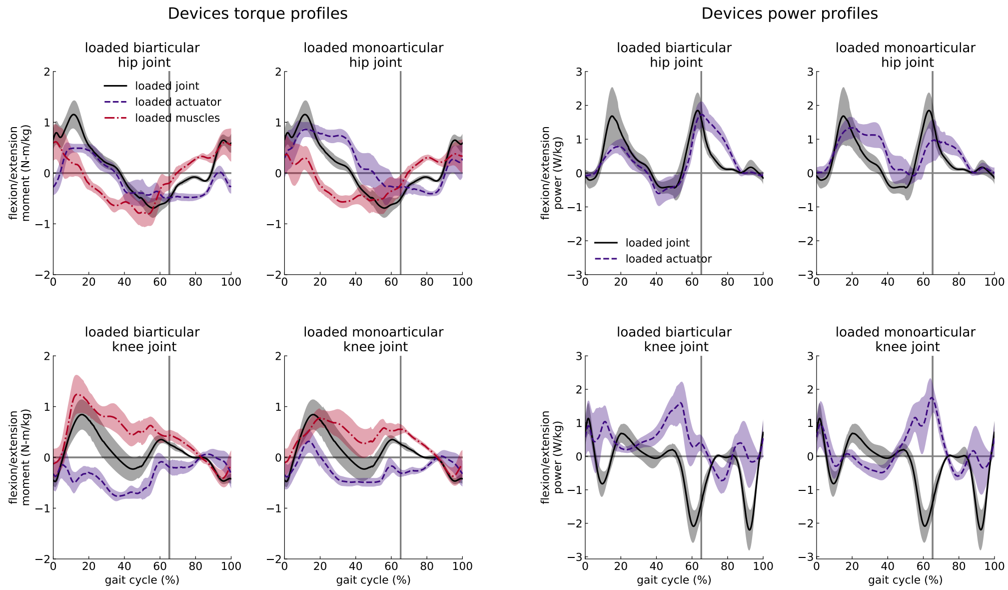


Fig C3. Assistive devices torque and power profiles. The actuator torque and power for subjects carrying heavy load (dark purple) , and net joint power and torque profile for *loaded* (black) condition are shown for each actuator of the devices. The torque profile of moment generated by muscles (dark red) is shown for each joint for both devices. The curves are averaged over 7 subjects with 3 trials and normalized by subject mass; shaded regions around the mean profile indicate standard deviation of the profile.

The overall trend between the torque profiles of the monoarticular and biarticular devices was similar, which can be seen in Figure C3 qualitatively, and the root mean square error between the profiles of the selected devices during a gait cycle also supports this claim quantitatively, as shown in Figure C4. A detailed analysis of the gait cycle shows that the main difference between the torque profiles of the hip actuators occurred during the stance phases, and the monoarticular device delivered hip extension torque greater than the biarticular device, especially during the mid-stance and terminal-stance phases (Figure C3 and C4). Although the difference between the profiles of the hip actuators followed similar trajectories during the swing phase, the terminal-swing phase of these actuators was considerably different in that the monoarticular exoskeleton delivered extension while the biarticular one provided flexion torque to the joint.

Although the biarticular and monoarticular knee actuators had almost identical trajectories during the swing phase, as their RMSE shows in Figure C4, there were some significant differences between these two actuators during the stance phases. While the biarticular knee actuator opposed the torque generated by muscles around the knee joint during the all stance phase, the monoarticular knee actuator assisted torque generated by the knee muscles during the loading response and mid-stance phases.

These remarkable differences between the torque profiles of the two devices affected the torque trajectories generated by muscles around the knee and hip joints, indicating muscular activation differences between subjects assisted by the monoarticular and biarticular exoskeletons; nevertheless, according to the root mean square error between

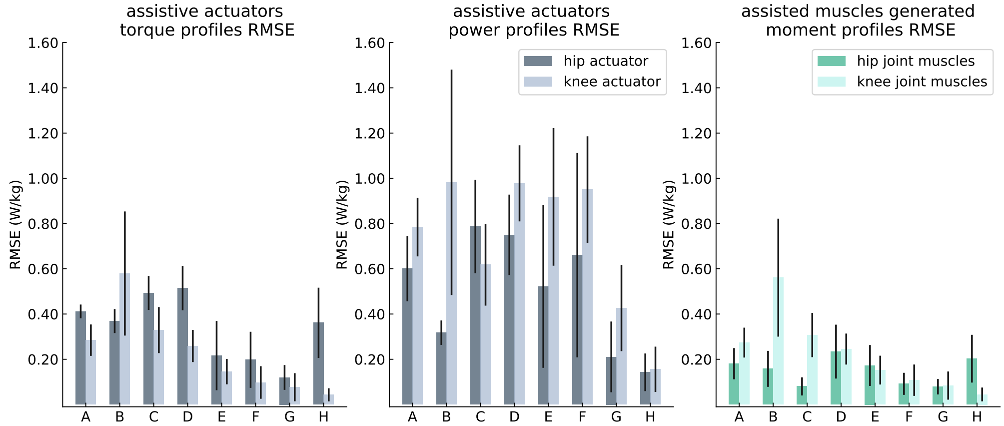


Fig C4. Assistive devices torque, power, and muscles generated moment profiles root mean square error. The root mean square error between actuators of biarticular and monoarticular devices and the muscles generated moment of subjects assisted by these devices. The RMSE was calculated during a total gait cycle (A), loading response (B), mid stance (C), terminal stance (D), pre swing (E), initial swing (F), mid swing (G), and terminal swing (H) phases.

torque trajectories generated by hip and knee muscles, the differences were not substantial except on the loading phase of the knee joint.

The comparison between the muscular activation of the *loaded* subjects assisted by ideal devices and constrained devices indicates substantial differences in some muscles. The activation of rectus femoris and psoas as two primary muscles on the hip and knee was considerably different in the ideal and constrained devices. The constrained biarticular and monoarticular exoskeletons also had different impacts on these two muscles (Figure C5). The difference between the activation of rectus femoris and vasti muscles of subjects assisted by the constrained optimal biarticular and monoarticular devices during the loading response phase explains the difference between the muscles generated moments of subjects assisted by the biarticular and monoarticular devices. Another difference between the ideal and torque limited devices was in the gastrocnemius medial head muscle; the activation of this muscle increased during the loading response to terminal stance phases. Due to the higher activation of the gastrocnemius muscle, it provided a greater moment on the ankle joint. Consequently, the activation of the soleus muscle did not increase considerably to compensate for the inadequacy of the moment generated by gastrocnemius at the ankle joint. The differences between the muscular activation of subjects assisted by monoarticular and biarticular constrained devices were not limited to the rectus femoris and psoas muscles. The other representative muscles also demonstrated some differences, as shown in Figure C5; nevertheless, the differences between them were not as considerable as those of the rectus femoris and psoas muscles.

Unlike the moment profiles of devices in which the difference was significant only in some specific phases, the power profiles of the biarticular and monoarticular devices had significant differences, as shown qualitatively and quantitatively in Figures C3 and C4, respectively. The difference between the power profiles of these two devices was notable in the knee actuator in which the devices followed different trajectories during the gait cycle. Similar to the knee actuators, the hip actuators had roughly different power profiles, and their maximum power consumption occurred in two completely different phases, similar to the knee actuators. The difference between the trajectories and

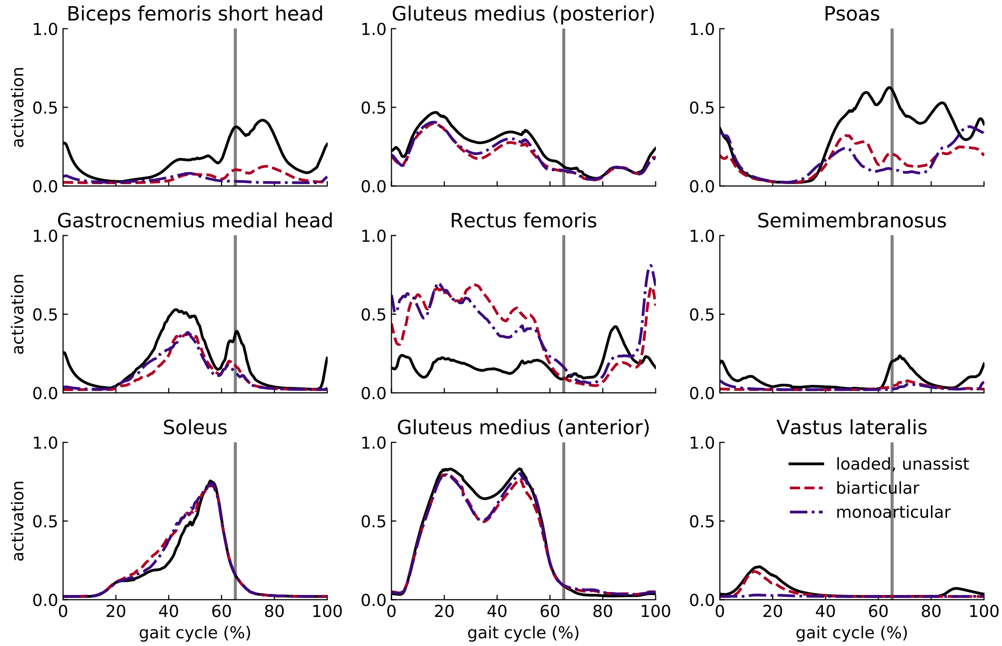


Fig C5. Activation of representative lower limb muscles of assisted and unassisted subjects. The activation of unassisted subjects carrying heavy load (black), and subjects assisted by "Ad" monoarticular (dark purple), and "Da" biarticular (dark pink) devices are shown for nine important muscles. The curves are averaged over 7 subjects with 3 trials.

magnitudes of the power profiles also explains the difference observed between the power consumption of the monoarticular and biarticular exoskeletons (Fig.C2).

Studying the selected monoarticular and biarticular devices proves that devices with the same total power consumption can have different power consumption in different joints. Additionally, we showed that optimal devices with the same performance could follow different moment and power profiles, even under kinematic similarities due to the arrangement of the actuators. Although the devices were selected from the ideal Pareto front with the same performance, employing the modified augmentation factor for the monoarticular and biarticular devices with different mass and inertia characteristics indicated the superior performance of the biarticular device. This emphasizes the discussion held in the "Effect of Optimal Device Inertial Properties on Subject Metabolics" section in that the biarticular device could deliver the same amount of assistance to the subjects more effectively than the monoarticular configuration.

Case 2: Devices Performance in *Noload* Condition

In the second case study, we selected two devices with similar power consumption and the same effect on the metabolic rate of subjects walking without any external load, which are shown as "Cb" and "Ba" on the Pareto front curves of the biarticular and monoarticular devices, respectively, in the *noload* condition. Similar to the first case study, while the total performance of these two biarticular and monoarticular devices was similar in mean values on the trade-off curves, the actuators had different power consumptions.

The power consumption of the hip actuators in both exoskeletons had a high

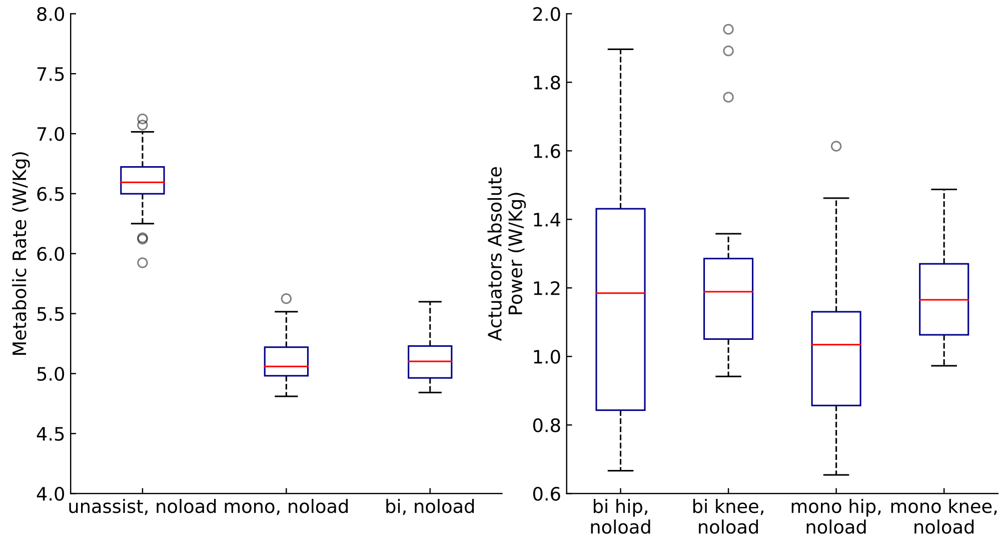


Fig C6. Assistive devices power consumption and its effect on the metabolic rate. The power consumptions of assistive devices and their effect on whole-body metabolic rate of the subjects walking at self-selected speed without any additional load. Asterisks indicate statistically significant differences (7 subjects, 3 trails, Tukey Post-hoc, $P < 0.05$).

within-subject deviation, as shown in Figure C6, which explains the absence of statistically significant differences between the actuators. Additionally, the metabolic rate of subjects assisted with the monoarticular and biarticular devices had no significant differences. However, the metabolic cost reduction caused a significant difference between the metabolic expenditure of unassisted and assisted subjects, as represented in Figure C6. Despite the large variation between the devices' power consumption and the absence of significant difference among their actuators, employing the modified augmentation factor indicates the different performance of the monoarticular and biarticular exoskeletons delivering assistance to the subjects. The selected devices had a different design in the actuators in which the biarticular device could provide maximum 50 and 60 N-m torque in the hip and knee actuators, respectively, while the maximum moments in the hip and knee actuators of the monoarticular device were 60 and 70 N-m, respectively. The computation of MAF under the mentioned configurations of these two devices resulted in 1.57 ± 0.72 and 0.42 ± 0.85 W/kg for the biarticular and monoarticular devices, respectively, indicating the superior performance of the biarticular device similar to the first case. The performance of these two devices can also be discussed based on the Pareto front of devices under inertia and mass effect, as shown in Figure 17 in the paper. According to this analysis, the studied configuration of the monoarticular device became a dominated solution in Pareto simulations under the inertial properties of devices, while the chosen biarticular device could maintain its efficiency under the negative effect of its inertial properties on the metabolic rate of subjects. This analogy between the Pareto front under the effect of the inertial properties of the devices and MAF can also confirm the extension of the augmentation factor.

The analysis of moment profiles of assistive devices in the *noload* condition shows that the differences between these two devices were similar to the difference between the biarticular and monoarticular devices in the *loaded* circumstance, which is represented in Figure C7. Nevertheless, the variations of moment generated by muscles of the

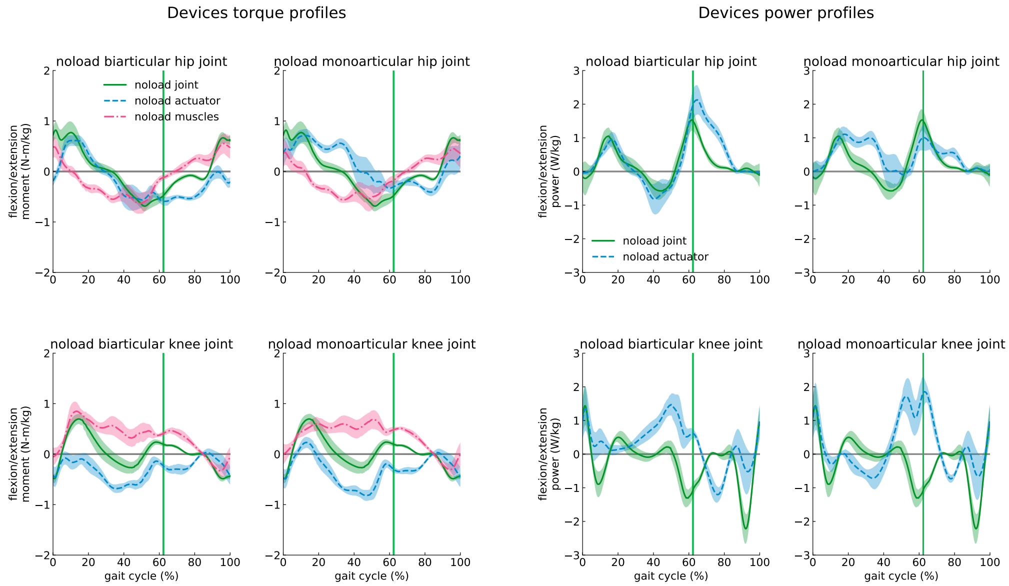


Fig C7. Assistive devices torque and power profiles. The actuator torque and power for subjects walking without an additional load (blue) , and net joint power and torque profile for *noload* (green) condition are shown for each actuator of the devices. The torque profile of moment generated by muscles (rose pink) is shown for each joint for both devices. The curves are averaged over 7 subjects with 3 trials and normalized by subject mass; shaded regions around the mean profile indicate standard deviation of the profile.

assisted subjects were negligible in the *noload* condition (Figure C8), which signals similar muscular activation of subjects assisted by these two exoskeletons. Unlike the moment profiles, the devices' power profiles were different in the *noload* condition. It can be seen from the power profiles that the devices followed remarkably different trajectories during a gait cycle to deliver assistance to the subjects and these profiles in the hip actuators, similar to those in the knee actuators, had the highest contrast during the pre-swing phase, according to their RMS error, as shown in Figure C8.

Studying specific optimal monoarticular and biarticular exoskeletons in two load conditions, chosen from the Pareto fronts, shows that even though the devices had practically the same performance in the optimal trade-off between the device total power consumption and metabolic cost reduction curves, their provided moments during a gait cycle had considerable differences, which could cause a different effect on the muscular activation of assisted subjects as well. These two case studies also show that the power profiles of monoarticular and biarticular devices are considerably different, while they have a similar total power consumption.

The study on the performance of selected devices in both loading conditions by developed MAF factor supports the discussion in the "Effect of Optimal Device Inertial Properties on Subject Metabolics" section on the effect of mechanical design on the performance of devices and also showed that the performance of the monoarticular exoskeleton was highly affected by the inertial properties of the device. Although the mechanical design of a biarticular device can be complicated, its performance under the device inertial properties seems promising in both loading conditions, according to the performance of the studied cases. The studied cases and general Pareto front of the monoarticular device under the effect of its inertial characteristics show that this type of device needs to be designed thoughtfully to reduce the effect of inertia and mass effect

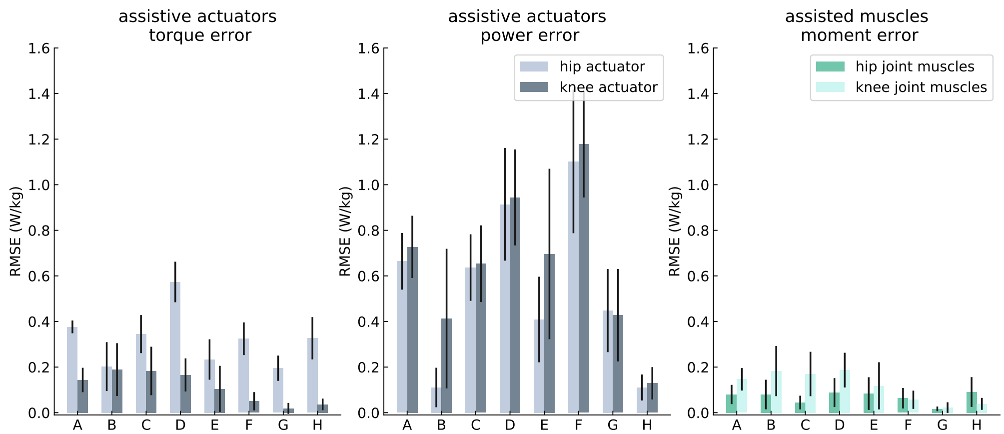


Fig C8. Assistive devices torque, power, and muscles generated moment profiles root mean square error. The root mean square error between actuators of biarticular and monoarticular devices and the muscles generated moment of subjects assisted by these devices. The RMSE was calculated during a total gait cycle (A), loading response (B), mid stance (C), terminal stance (D), pre swing (E), initial swing (F), mid swing (G), and terminal swing (H) phases.

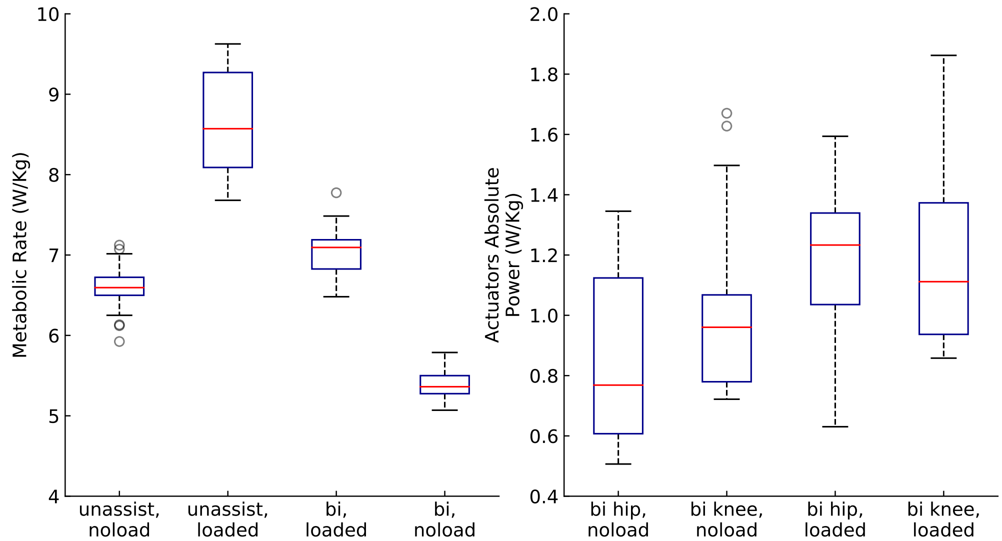
of the device on the metabolic burden of subjects, complicating the design procedure, and ignoring the mechanical design results in delivering no assistance to the subject, or increasing their metabolic burden.

Case 3: Biarticular Exoskeleton Performance

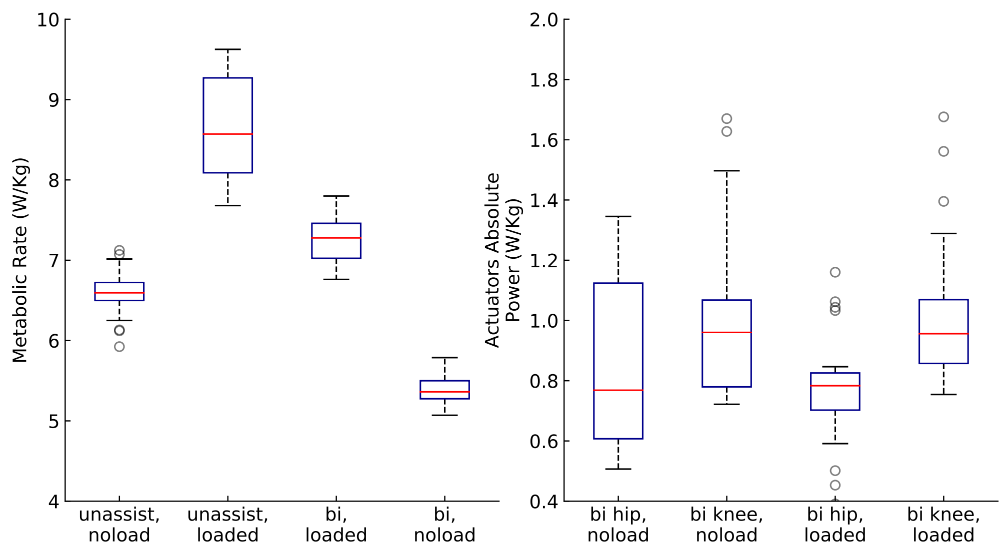
To study how the performance of a biarticular exoskeleton changes by loading subjects with a heavy weight on torso more specifically, we chose two cases in which the biarticular exoskeletons had the same effect on the metabolic rate of subjects in one case and had the same power consumption in another case.

The selected configuration of the biarticular exoskeleton in the *noload* condition was "Ec" with 30 and 50 N-m peak torque in the hip and knee actuators, respectively, and it was compared to the same configuration (i.e., "Ec") in the *loaded* condition in which they had practically the same power consumption. In order to conduct a comparison with the similar metabolic burden reduction, the same configuration of the device in the *noload* condition (i.e., "Ec") was compared to the biarticular exoskeleton with 50 and 60 N-m peak torque in the hip and knee actuators represented by "Cb" on the Pareto front of the *loaded* biarticular exoskeleton.

Comparing the metabolic rate of assisted subjects by the biarticular devices in two conditions, which were similar metabolic cost reduction conditions and the same power consumption condition, were represented in Figure C9. The metabolic rates of subjects in both conditions show that the metabolic rate of the *loaded* subjects was reduced considerably, and there was no significant difference between subjects walking with no load and the *loaded* subjects assisted by the biarticular device. The power consumption of the hip actuators and knee actuators showed no significant differences when the selected device consumed a similar amount of the power in different loading conditions. Additionally, a similar performance in power consumption of two different configurations of the biarticular exoskeleton delivering a similar amount of assistance in different load conditions was observed, which is represented in Figure C10(a).



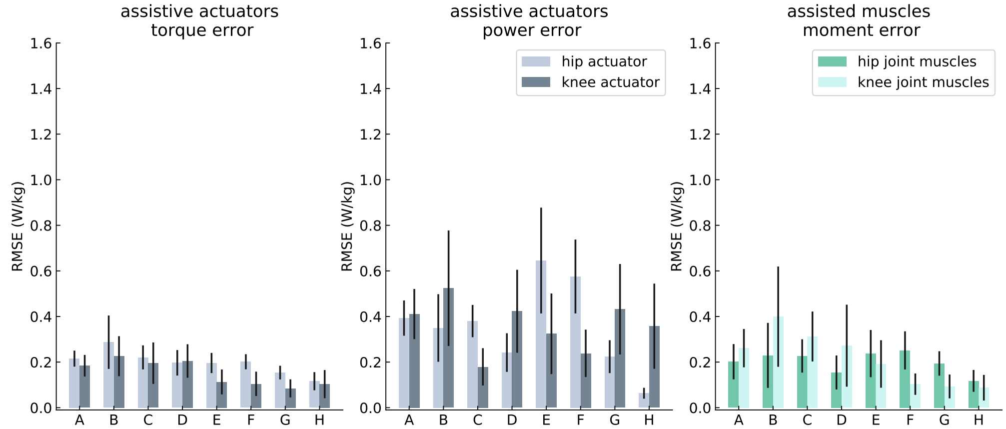
(a) Biarticular exoskeleton with the same effect on the metabolic consumption of subjects



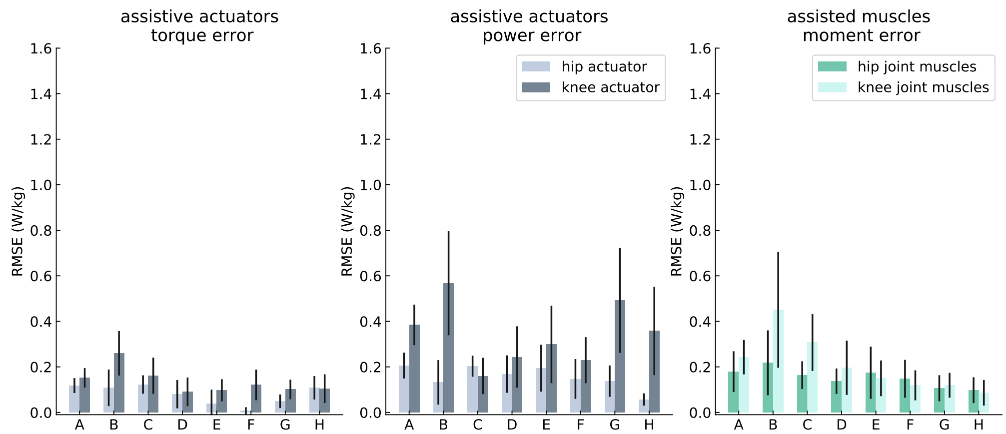
(b) Biarticular exoskeleton with the same total power consumptions

Fig C9. Biarticular exoskeleton power consumption and its effect on the metabolic rate in different load conditions. The power consumptions of biarticular exoskeleton and their effect on whole-body metabolic rate of the subjects walking at self-selected speed in both *loaded* and *noload* condition. Asterisks indicate statistically significant differences (7 subjects, 3 trails, Tukey Post-hoc, $P < 0.05$).

The absence of a significant difference between the two devices with different configurations and different load conditions can facilitate designing a battery with a robust performance to the different load conditions. This performance can also help to achieve a general mechanical design for an exoskeleton to assist subjects in different load conditions and assistance levels. Nevertheless, the high within-subject deviations and outliers of power consumption indicate high contrast within-subjects, which can complicate obtaining general power profiles for the device.



(a) Biarticular exoskeleton with the same effect on the metabolic consumption of subjects

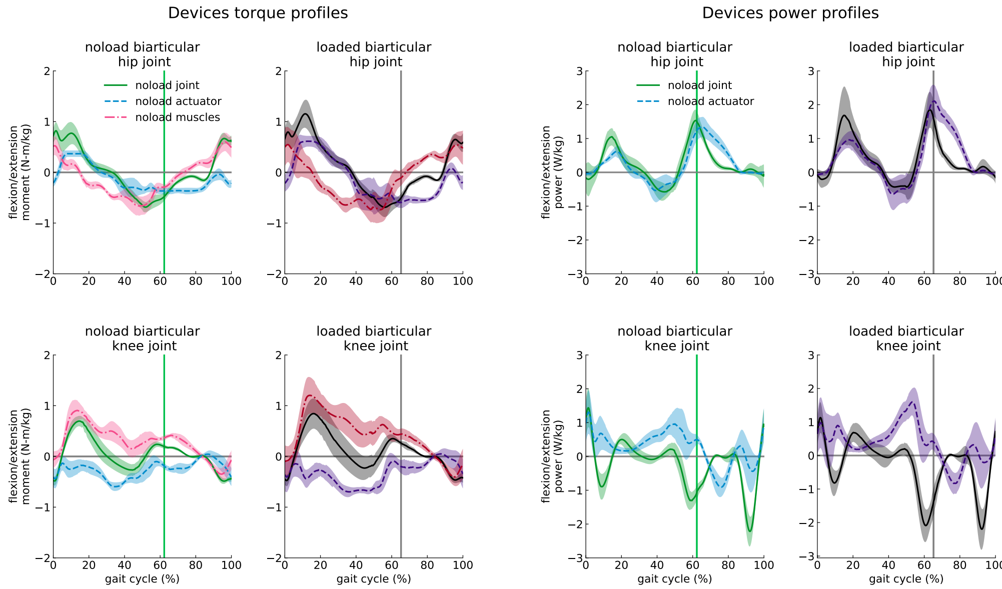


(b) Biarticular exoskeleton with the same total power consumptions

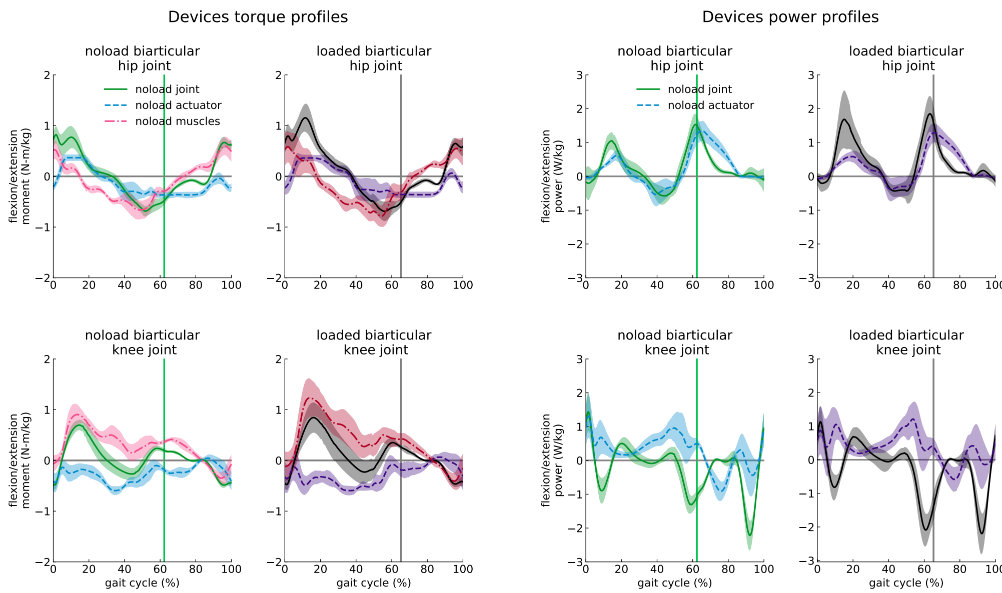
Fig C10. Biarticular exoskeleton torque and power and muscles generated moment profiles root mean square error in different load conditions. The root mean square error between actuators of the biarticular exoskeleton and the muscles generated moment of subjects assisted by this device *loaded* and *noload* conditions. The RMSE was calculated during a total gait cycle (A), loading response (B), mid stance (C), terminal stance (D), pre swing (E), initial swing (F), mid swing (G), and terminal swing (H) phases.

The performance assessment of the same biarticular configuration in two different load conditions by employing MAF showed that the performance of the biarticular exoskeleton in the *loaded* condition was improved (1.40±0.80 W/kg) in comparison with the *noload* condition, in which the MAF value was 1.01±0.70 W/kg. Although the increase in the device's positive power in the *loaded* condition was expected, improvement of the MAF value shows that this increase in positive power was delivered to the subjects effectively. In the meanwhile, comparing the devices with the same effect on the metabolic cost reduction of subjects in different load conditions showed the superior performance of the biarticular device in the *loaded* condition in which devices in the *loaded* and *noload* circumstances had 2.08±0.69 and 1.01 ±0.69 W/kg MAF values, which can indicate the inefficiency of noload device power profiles in general.

The quantitative and qualitative analyses of power and moment profiles between the pair of selected devices not only show a moderate variation between the torque profiles of the compared biarticular devices; the power profiles also demonstrated considerably



(a) Biarticular exoskeleton with the same effect on the metabolic consumption of subjects

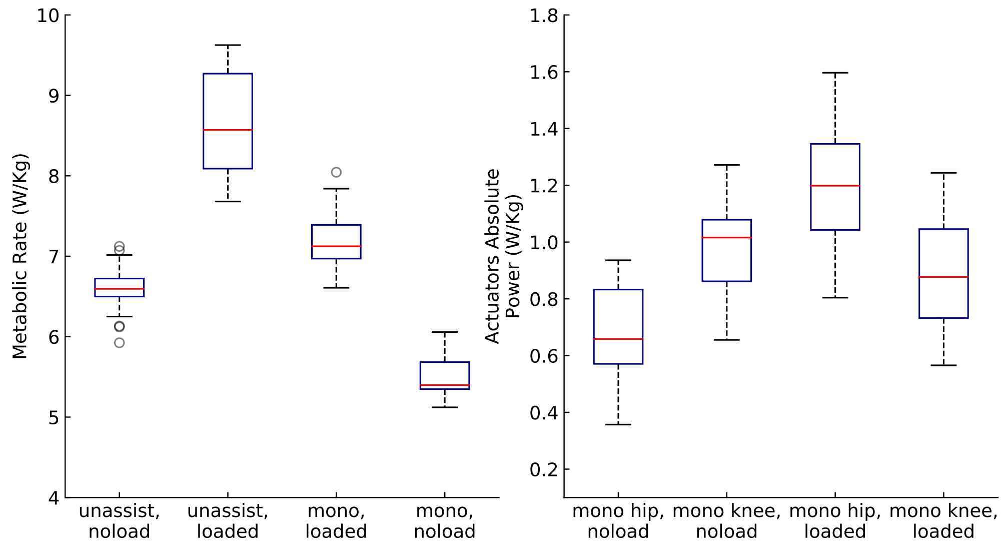


(b) Biarticular exoskeleton with the same total power consumptions

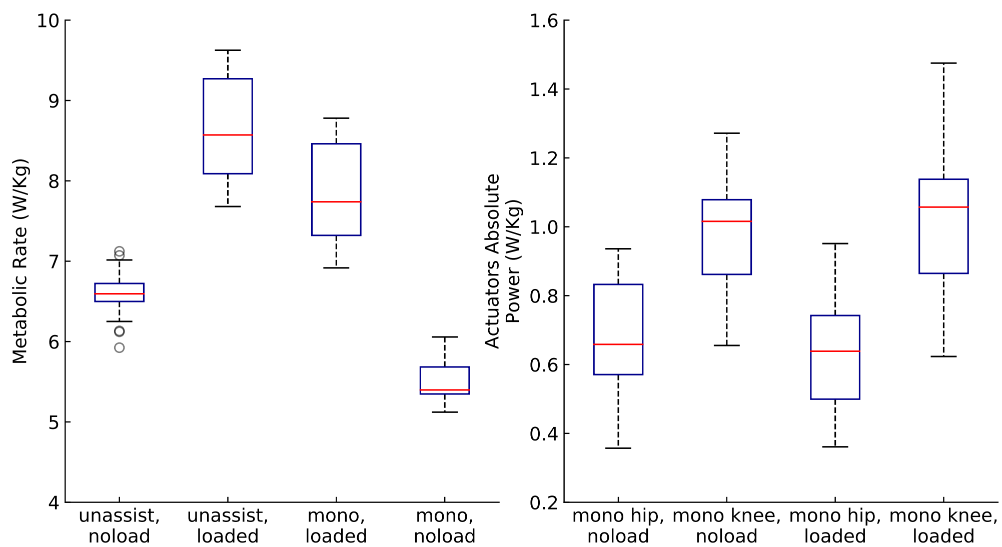
Fig C11. Biarticular exoskeleton actuators torque and power profiles and muscles generated moment of assisted subjects.

low diversity in different load conditions, as shown in Figure C10(a), C10(b). Additionally, Figure in C11 supports our claims regarding the high resemblance between profiles of the biarticular exoskeletons in the *loaded* and *noload* conditions. As shown in Figure C11, the difference between the power and moment profiles of the pair of devices in *load* and *noload* conditions were nearly limited to the magnitude and timing based on the toe-off difference except on the pre-swing and initial-swing phases of knee profiles in which the trajectories had differences between biarticular devices with the same effect on the metabolic power consumption of assisted subjects.

Case 4: Monoarticular Exoskeleton Performance



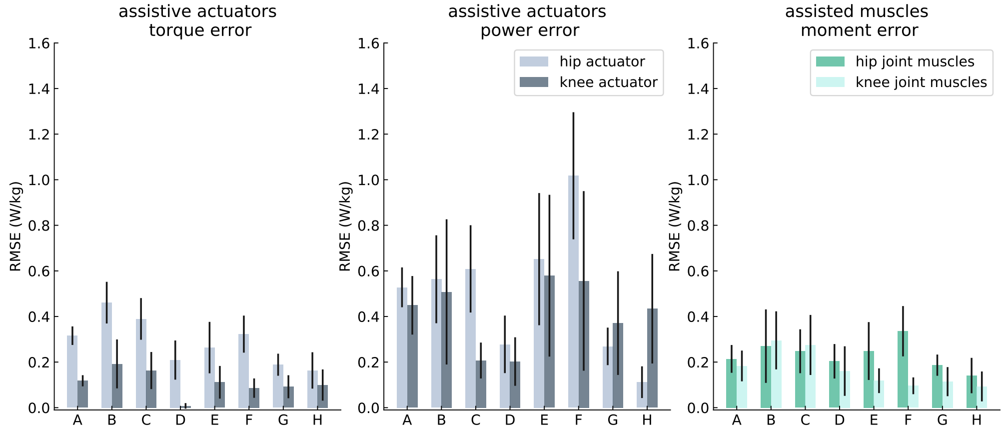
(a) Monoarticular exoskeleton with the same effect on the metabolic consumption of subjects



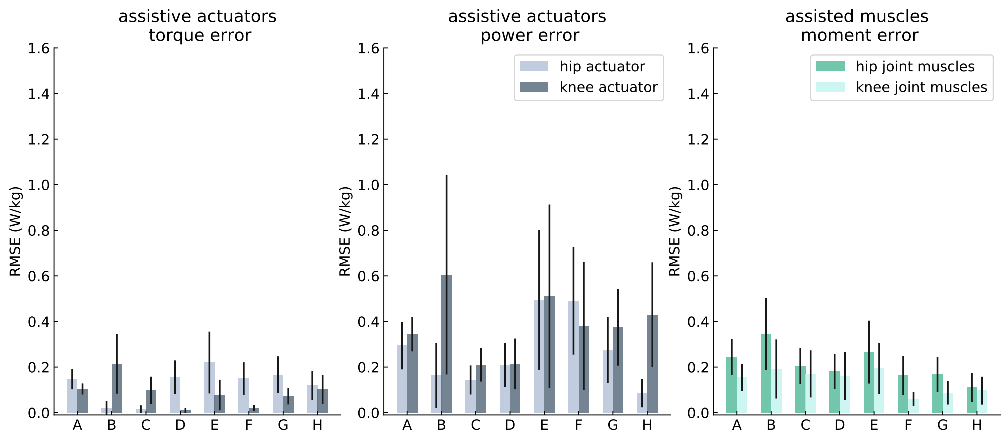
(b) Monoarticular exoskeleton with the same total power consumptions

Fig C12. Monoarticular exoskeleton power consumption and its effect on the metabolic rate in different load conditions. The power consumptions of biarticular exoskeleton and their effect on whole-body metabolic rate of the subjects walking at self-selected speed in both *loaded* and *noload* condition. Asterisks indicate statistically significant differences (7 subjects, 3 trails, Tukey Post-hoc, $P < 0.05$).

The same analyses on the biarticular exoskeleton, discussed in the previous case study were performed on the monoarticular exoskeleton to gain an in-depth insight into this type of exoskeleton. To conduct the comparisons between a pair of monoarticular devices with a similar effect on the metabolic cost or with similar power consumption, we chose the "Ee" configuration of the monoarticular device in the *loaded* and *noload* conditions and the "Ae" monoarticular exoskeleton from the Pareto front of the



(a) Monoarticular exoskeleton with the same effect on the metabolic consumption of subjects

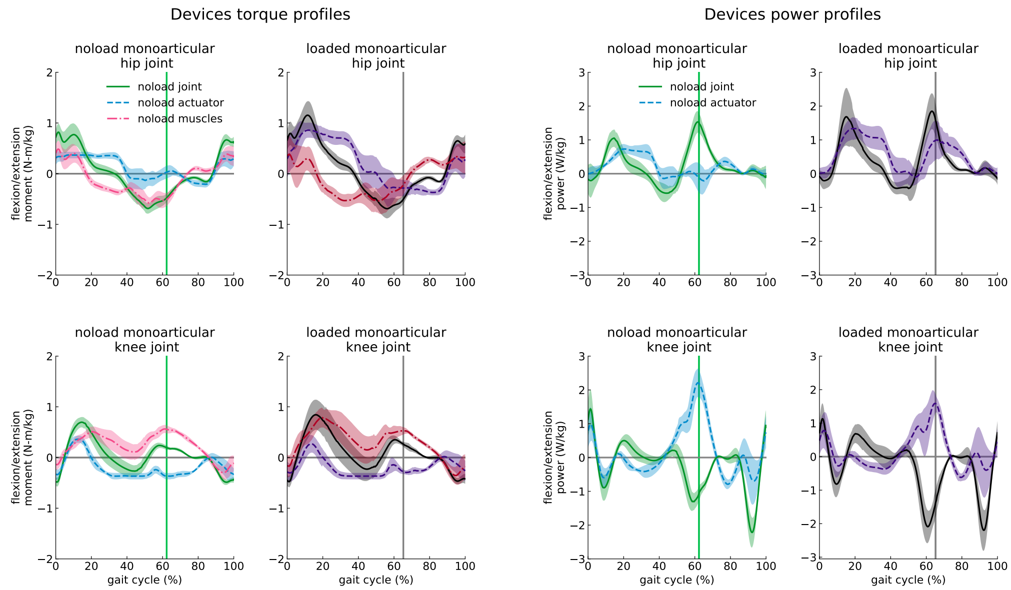


(b) Monoarticular exoskeleton with the same total power consumptions

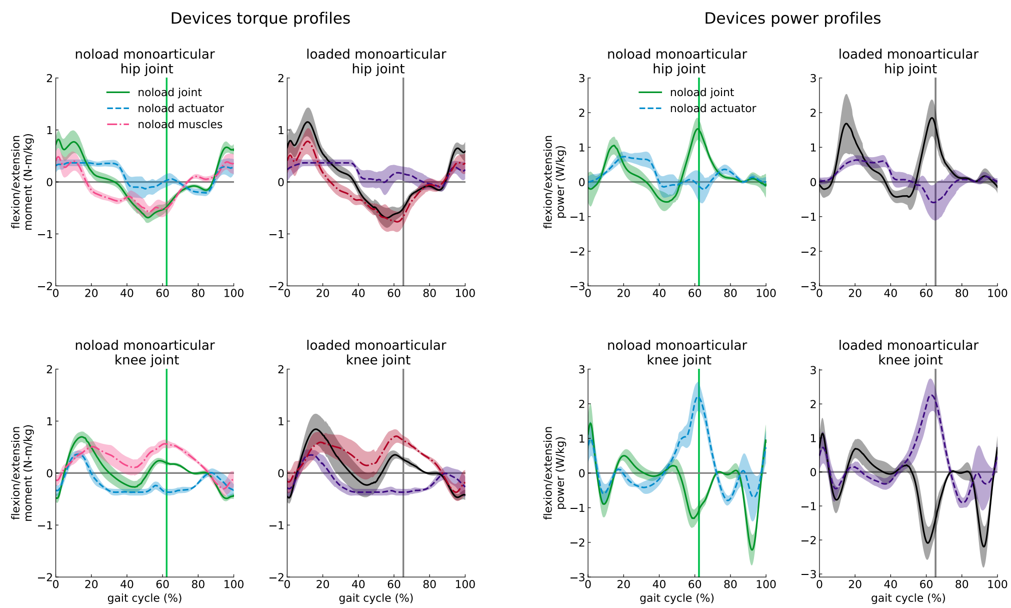
Fig C13. Monoarticular exoskeleton torque and power and muscles generated moment profiles root mean square error in different load conditions. The root mean square error between actuators of the biarticular exoskeleton and the muscles generated moment of subjects assisted by this device *loaded* and *noload* conditions. The RMSE was calculated during a total gait cycle (A), loading response (B), mid stance (C), terminal stance (D), pre swing (E), initial swing (F), mid swing (G), and terminal swing (H) phases.

monoarticular device in the loaded condition. The selected "Ee" and "Ae" configurations have 30 and 30 N-m, and 70 and 30 N-m peak torque constraints on the pair of hip and knee actuators, respectively.

The comparison of actuators power consumption between the "Ae" *loaded* and "Ee" *noload* devices, which have a similar metabolic cost reduction, showed a statistically significant increase in power consumption of the *loaded* hip actuator. Despite the reduction in the knee power consumption of the *loaded* knee actuator, the difference between the pair of knee actuators was not significant. One of the observed critical issues was that even though the within-subject variation of monoarticular devices was generally low, the deviation of actuator power consumptions between the load condition and between the configurations was considerable. As shown in Figure C12(a) along with Figure C12(b), while the power consumption of the knee actuator in the low torque availability (i.e., "Ee") was higher than that of the hip actuator, this was changed in higher torque constraints, and the hip actuator became dominant power consumer indicating considerable changes in the power profile of the monoarticular device in different arrangements of its actuators.



(a) Monoarticular exoskeleton with the same effect on the metabolic consumption of subjects



(b) Monoarticular exoskeleton with the same total power consumptions

Fig C14. Monoarticular exoskeleton actuators torque and power profiles and muscles generated moment of assisted subjects.

Despite the similar percentage of the metabolic rate reduction, the metabolic rate of assisted subjects in the *loaded* condition has a significant difference with that of the unassisted *noload* subjects, which indicates that the monoarticular device was not able to sufficiently compensate the cost of carrying a heavy load. Comparing the metabolic rate of subjects assisted by two monoarticular devices with a similar power consumption also shows that compensating additional load is more costly for monoarticular devices.

The metabolic rate of the compared pairs of monoarticular devices is shown in Figures C12(a) and C12(b). The pair of devices selected for conducting the comparisons were evaluated by the modified augmentation factor to assess their performance under their mass and inertia effect. Unlike the other case studies in which all evaluated biarticular and monoarticular devices delivered a positive power to the human musculoskeletal system, the monoarticular exoskeletons evaluated in this case study both caused an increase in the metabolic expenditure of subjects according to MAF performance factor. The modified augmentation factor of the monoarticular device with 30 N-m peak torque constraints in both hip and knee actuators showed -0.20 $\pm$ 0.43 W/kg in the *noload* condition while it increased to -0.15 $\pm$ 0.56 W/kg when subjects were loaded. These MAF values of the "Ee" monoarticular device represent a high variation of device effect on the subjects and improvement of the device performance by loading subjects, which was also observed in the biarticular exoskeleton. Even with the improvement of device performance in the loaded condition, the device in both loading conditions would cause subjects to consume more power due to wearing these devices. The same analysis on the monoarticular device "Ae" configuration (70 N-m hip 30 N-m knee) in the *loaded* condition showed relative improvement compared to the "Ee" device.

According to the MAF value, the monoarticular exoskeleton requires a large torque capacity at the hip actuator to deliver assistance to the subjects. According to the slope of the Pareto front of the monoarticular device under the devices inertial properties effect, we claimed that it might be beneficial to keep the torque capacity of the monoarticular exoskeleton more moderate, yet, we observe in this case study that the monoarticular device cannot inject positive power to the human musculoskeletal system in low torque capacity. According to the MAF value and our previous observations, it might be reasonable to conclude that designing an optimal monoarticular exoskeleton that can be used for different assistance levels and load conditions is complicated.

The moment and power profiles of the selected monoarticular exoskeleton did not show a similar resemblance that we observed in the biarticular device between the pair of actuators. The hip actuators of devices with the same metabolic reduction effect followed completely different trajectories, and it was not surprising that their power profiles had significant differences. Although the moment profiles of the knee actuators had a higher resemblance than the hip actuators, their power profiles showed relatively different paths during pre-swing and initial swing phases. The comparison of the same device (i.e., "Ee") profiles in different load conditions showed a considerable divergence between the profiles of the hip actuator after the mid-stance phase of the gait cycle and their maximum difference occurred during the pre-swing and initial swing phases. Although the knee actuators showed similar torque profiles in different load conditions, their power profiles demonstrated remarkable differences during pre-swing and initial swing phases. These described moment and power profiles; their quantitative differences, using the RMSE, in both comparison cases are represented in Figures C14 and C13.

This case study confirms our discussion about monoarticular exoskeleton in which we claimed that obtaining a generic control policy for this device would be challenging. Also, designing an optimal battery under its life and weight considerations highly depends on the selected configuration, and a generic battery would not perform optimally for the monoarticular device at different assistance levels and in different load conditions.