**\subsubsection\*{Optimal device torque and power profiles }**

\begin{figure\*}[ht]

\end{figure\*}

The torque profiles of constrained optimal devices differed from the net joint moments of the hip and knee joints, and the general torque trajectories of these actuators were mostly similar to the torque profiles of the devices with ideal actuators. The ideal and constrained torque profiles of biarticular hip actuators had magnitude differences mostly during mid-stance and terminal-stance to terminal-swing phases. In contrast to the hip actuator, the knee actuator had practically the same profile as the ideal knee actuator during the swing phase, but the path and magnitude of the knee actuator were mostly different from those of the ideal device. This comparison between the ideal and constrained devices is mostly valid when subjects were walking without carrying any load, and the only difference was the magnitude of the constrained profiles.

Both actuators of the constrained monoarticular device demonstrated differences that were more remarkable in comparison with the profiles of the ideal device. The hip actuators had a significant magnitude difference during the load response phase, and most of the optimal constrained hip actuators were saturated during the mid-stance and terminal stance phases, which affected their trajectories as well. The difference between hip actuators became significant during the pre-swing to mid-swing phases, during which the torque trajectories of the constrained hip actuator not only were different from the ideal actuator but also exhibited high variations among optimal constrained actuators. The monoarticular knee had greater resemblance to the ideal knee actuator torque profile when (the 2 actuators you’re talking about) had practically identical profiles during the swing phase, and the differences between them were the higher torque magnitude during the mid-stance and lower magnitude during the pre-swing phases.

Unlike the biarticular exoskeleton, in which the load condition only affected the magnitude of the profiles and there was a close similarity between the torque trajectories in \textit{loaded} and \textit{noload} conditions, the torque profiles of the hip actuator of the monoarticular device assisting the subjects walking without carrying any load had considerable differences with the same device in the \textit{loaded} condition. The hip torque profiles of the device in different load conditions exhibited two mostly different trajectories and magnitudes during all phases of a gait cycle, and their differences were more unambiguous during load response to mid-swing phases.

\begin{figure\*}[ht]

\end{figure\*}

The power profiles of the restrained biarticular and monoarticular devices resembled those of the ideal devices, similar to the torque profiles. The biarticular hip actuator had mostly a similar power trajectory for the hip in both load conditions with considerably lower magnitude for all optimal devices where this magnitude difference is more substantial during mid-stance and initial swing to mid-swing phases. While the knee actuator had a high correlation with the ideal actuator, the loading response and mid-stance phases of the constrained knee actuator was different from the ideal actuator, like the torque profile. The constrained monoarticular hip actuator had a high variation within optimal devices, and most of the optimal configurations had their unique power profile; nevertheless, the optimal devices with the highest peak torque limitations had a close resemblance to the ideal device. In contrast to the hip actuators, the power trajectories of the knee actuator in both load conditions were similar to those of the ideal device with a difference during the pre-swing to mid-swing phases when the peak power consumption occurs about the toe-off in the constrained knee actuators.

The power and torque profiles of optimal biarticular and monoarticular devices reveal that although the optimal monoarticular exoskeletons have a lower energy consumption compared to the optimal biarticular devices, the variation of the torque and power profiles within optimal configurations of the monoarticular device is higher than that of biarticular devices and the load condition of the subjects can considerably affect the profiles of assistive actuators. These variations within optimal monoarticular devices indicate that achieving a generalized design and control policies for assisting subjects in different load conditions, different actuation, and battery life limitations would be genuinely challenging with monoarticular exoskeletons.

We conducted comparisons between devices and load conditions to analyze the torque and power profiles of selected solutions from the Pareto front of devices in different load conditions qualitatively and quantitatively (using a/the root mean square method) in each phase of a gait cycle in \nameref{S4\_Appendix} and discussed them comprehensively. Through these comparisons, we showed that even though two devices can have the same performance on the defined objectives space, they can have different power and torque profiles for delivering the assistance, and these differences also affect the muscle coordination of the assisted subject in whom some muscles like rectus femoris and psoas demonstrated slightly different activation profiles. These comparisons also showed that the biarticular exoskeletons have approximately the same effect on the subjects in different load conditions comparing optimal devices with about the same power consumption.

\paragraph{ Effect of optimal devices on the reaction forces and moments} The profiles of reaction forces and moments of subjects assisted by the constrained optimal devices mostly resembled the profiles of reaction forces and moments of subjects assisted by the ideal exoskeletons, and the difference on their maximum suppliable torque did not considerably affect the profiles of the reaction forces and moments. The reaction moments and forces at the ankle joint closely followed the profiles of the ideal devices, indicating that the optimal biarticular and monoarticular devices had practically the same effect on the muscles contributing to the reaction loads and moments of the ankle joint, as represented in Figures 8 and 9 in \nameref{S3\_Appendix}. Similar to the ankle joint, the reaction loads of the hip joint also followed the trajectories of the subjects assisted by the ideal devices. Nevertheless, there was a magnitude difference between the ideal and constrained trajectories, especially during the stance phase of a gait cycle, as shown in Figures 10 and 11 in \nameref{S3\_Appendix}.

The devices had a different effect on the reaction forces and moments at the knee and patellofemoral joints. Unlike the ideal biarticular device, in which the peak reaction moments and forces were reduced in loading response and increased in the late stance phase, the constrained optimal exoskeletons were not able to reduce the peak reaction loads and moments at the loading response phase. Nonetheless, they demonstrated a better performance than the ideal biarticular device in the late stance phase and did not increase the reaction forces as ideal devices (Figures 12 and 13 in \nameref{S3\_Appendix}). It may be reasonable to deduce that this different behavior was due to the changes in the activation of the rectus femoris, iliopsoas, and hamstring muscles, which was shown in the case studies in \nameref{S4\_Appendix} for selected configurations.\\ The constrained optimal monoarticular exoskeletons had better performance on reducing the reaction moments and forces. The reaction moments and forces in this assistive device more closely resembled the trajectories of the reaction moments and forces in the ideal assistance scenario during the loading response, and the constrained devices were able to reduce the peak moments and forces during the late stance phase better than during their ideal configuration (Figures 14 and 15 in \nameref{S3\_Appendix}). The optimal constrained devices had more within device variations in the reaction moments and forces trajectories, which may be due to high within device deviations of their torque profiles. Similar to the differences between the profiles of ideal and constrained biarticular exoskeletons, the differences between the constrained monoarticular and biarticular exoskeletons also may/might be due to the discrepancy in their effect on the muscular activation of assisted subjects.

**\subsubsection\*{Regeneration Effect}**

\begin{figure\*}[t!]

\end{figure\*}

The significant power requirement of the untethered exoskeletons and the finite density of the power source of the proposed assistive devices constrain their assisting duration and make them remarkably dependent on their battery life \cite{140}. The review published by Young and Ferris \cite {36} reported that the maximum functioning duration of portable exoskeletons is 5 hours, indicating several recharging requirements in a day. Harvesting the dissipated energy of assistive devices can address this issue on mobile exoskeletons and help users to be more independent by prolonging the device battery life \cite{140,141,142}. Regenerating and harvesting energy has been utilized in different assistive devices to improve the efficiency of the device, which has been excellently reviewed in several papers \cite{140,141,142}.

Power harvesting had a remarkably positive effect on the devices, enabling some new configurations in both devices to become optimal solutions in the amount of power consumption for delivered assistance, as shown in Figure \ref{Fig\_Paretofronts\_Regeneration\_Efficiency\_Comparison} and \ref{Fig\_Regenerated\_Main\_Paretofronts}. Analyzing the performance of both devices throughout the reported efficiency range demonstrates that the performance of the system can be improved, even with a low power harvesting efficiency. Although designing an assistive device with a regeneration mechanism can be challenging from both mechanical and electrical perspectives, this result, shown in Figure \ref{Fig\_Paretofronts\_Regeneration\_Efficiency\_Comparison}, indicates that the performance and independence of the assistive device can be remarkably improved, even with a relatively weak performance of a regeneration mechanism. Additionally, this power requirement reduction can enable the battery and mechanical designers to reduce the load of the battery and device on the musculoskeletal system of subjects, which causes metabolic power consumption increase in an individual being assisted and requires the device to compensate for this increase ahead of providing assistance to the subject.

Conducting comparisons between the devices and load conditions shows that the regenerated monoarticular devices in different loaded conditions performed differently in that the performance and optimal configurations of monoarticular devices were considerably different in different load conditions. The regeneration, additionally, changed the slope of the biarticular Pareto front curves and enabled this configuration to achieve more optimal solutions on the higher torque requirement regions, while the performance of the monoarticular exoskeleton was not affected considerably in high peak torque regions, as shown in Figure \ref{Fig\_Regenerated\_Main\_Paretofronts}.

**\subsection\*{Optimal Devices Inertial Properties Effect}**

One of the main challenges with designing mobile exoskeletons is the effect of their mass and inertia augmented to the extremities of assisted subjects on the metabolic rate of subjects. The effect of mass and inertia on the metabolic cost has been studied by several researchers \cite{133,134} It has been shown that the metabolic rate of the subject changes considerably by adding mass and inertia \cite{133,134,135}.

The proposed exoskeletons in this study have remarkably different inertial properties due to their kinematic designs, and this difference results in a different effect on the metabolic power consumption of subjects. Since the current neural control algorithm of OpenSim is not able to simulate any variations in the musculoskeletal model that has not been captured by experimental data \cite{92}, we simulated the effect of the mass and inertia offline using the model proposed by Browning et al.\cite{133} for the effect of mass and inertia on the metabolic cost of subjects.

The study accomplished by Browning et al.\cite{133, proposed a linear model for the effect of adding mass and inertia on each segment of the lower limb by experimentally capturing and analyzing the effect of adding mass to different segments of the lower extremities and their inertia on the metabolic energy expenditure of the subjects. In this study, subjects walked at 1.25 m/s without carrying any heavy load on their torso, which is similar to the data captured from the subjects in the walking with \textit{\textit{noload}} condition at their self-selected speed\cite{93}. This qualitative match between the data and experimental protocols of Browning et al. and Dembia et al. enabled us to employ the developed model of \cite{133} to study the effect of mass and inertia added by assistive devices through offline simulations for the subjects walking at free speed without carrying any external load (i.e.{\it noload} condition).

\begin{figure\*}[ht]

\end{figure\*}

The effect of inertial properties of devices on the simulated devices was remarkable, changing most of the solutions on the optimal trade-off curve, as represented in Figure \ref{Fig\_Paretofronts\_Mass\_Regeneration\_Effect\_Comparison}. The highest peak torques in both monoarticular and biarticular exoskeletons are 60 N.m for the knee and 50 N.m for the hip joint. Since the inertia of devices was affected by altering the peak torque, these results indicate that the reflected inertia has a considerable effect on the optimality of a device and power consumption of an exoskeleton with high peak torque is not efficient in comparison to the amount of metabolic cost reduction they provide.

\begin{figure\*}[ht]

\end{figure\*}

While the biarticular exoskeleton showed a better performance than the monoarticular skeleton when the inertial properties of exoskeletons were considered (Figure \ref{Fig\_Paretofronts\_Mass\_Regeneration\_Effect}), the slope of Pareto front for both exoskeletons indicates that devices with higher torque capacity does not considerably/significantly change the amount of assistance that the device can provide to the subjects due to the inertia effect, which can be seen more obviously for the monoarticular device.

The analyses accomplished for the effect of the regeneration and device inertial characteristics on the performance of assistive devices can provide a qualitative perspective for mechatronic systems designers for designing assistive devices. The general outcome of these analyses shows that selecting an actuator with a high torque density is essential to reduce the reflected inertia effect of gear train on the power expenditure of subjects. The active inertia compensation methods by designing a controller also can be helpful in reducing the impact of inertia on the metabolic expenditure increase of subjects, which has been used by \cite{148} for controlling a one degree of freedom knee exoskeleton to compensate its inertia; however, this method comes with some coupled stability issues \cite{146,147,148} that need to be addressed while designing the controller.

The results also showed that keeping the actuator mass near proximal joints and assisting the joint of interest distally has a considerable impact on the metabolic power consumption of subjects. As we discussed previously, this mechanical design conclusion was already studied on the human musculoskeletal system, and it was shown the biarticular muscles enable human musculoskeletal structures to keep muscle volume near to the trunk and transfer power to the distal joints to reduce the inertia and mass of the leg. Consequently, the bio-inspired biarticular and multi-articular configurations of the assistive devices can provide a promising improvement in their performance.

Although assistive design with highly effective regeneration requires system-level optimization and complicating the design of assistive devices \cite{140}, even a qualitative comparison between devices with regeneration and without regeneration shows a remarkable difference in power consumption. This implies the necessity of regeneration, especially for untethered devices, to improve their independence and their operational duration.

As was discussed, the main difference between the inertial properties of the exoskeletons was the location of augmenting the knee actuator affecting the performance of the monoarticular exoskeleton significantly. According to the developed model, the performance of the monoarticular exoskeleton can be improved by embedding the knee actuator to the upper part of the shank or thigh while its kinematic remains constant. Therefore, we examined the performance of the monoarticular exoskeleton with two different inertial characteristics.

\begin{figure\*}[ht]

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As can be seen in Figure \ref{Fig\_MonoarticularExoskeleton\_DifferentConfigs}, both proposed alternative locations for embedding the knee actuator can improve the performance of the monoarticular exoskeleton under their mass and inertia effect. Among these configurations, attaching the knee actuator proximal to the waist performs better than the other configurations of the monoarticular exoskeleton. The proposed alternative monoarticular configurations can be used to design a monoarticular exoskeleton that enables designers to avoid biarticular device design complexities while achieving a performance superior to that of the typical monoarticular exoskeleton design. Additionally, the weight of the distal segments can be compensated to improve the performance of the device; however, compensating only for weight can come with adverse effects on the gait of subjects \cite{179}, which needs a careful assessment of the gravity compensation mechanism.

**\paragraph{Case studies.}** The performance assessment of the same biarticular configuration ( {\it loaded} and {\it noload} "Ec") in two different load conditions using introduced MAF showed that the performance of the biarticular exoskeleton in {\it loaded} condition was improved (MAF = 1.40$\pm$0.80 W/kg) in comparison with the {\it noload} condition in which MAF was 1.01$\pm$0.70 W/kg. Although the increase in the positive power of the device in the {\it loaded} condition was expected due to the discussed characteristics of power profiles in the {\it loaded} condition, this MAF improvement shows that the positive power in the {\it loaded} condition was delivered to the subjects effectively. Comparing the devices with the same effect on the metabolic cost reduction of subjects in different load conditions (i.e., {\it loaded} "Cb" and {\it noload} "Ec") showed the superior performance of the biarticular device in the loaded condition in which devices in {\it loaded} and {\it noload} circumstances had 2.08$\pm$0.69 and 1.01$\pm$0.69 W/kg MAF values. This finding requires validation through experimental investigations.

Conducting a comparison for the monoarticular exoskeleton configuration in different load conditions ( {\it loaded} and {\it noload} "Ee") using MAF resulted in -0.20 +/- 0.43 W/kg and -0.15 +/- 0.56 W/kg MAF values in {\it noload} and {\it loaded} conditions, respectively. These values show that although the performance of the monoarticular exoskeleton, similar to that of the biarticular device, was improved by loading subjects, the monoarticular devices in the low torque region were not able to improve the metabolic power consumption of subjects in comparison to the metabolic expenditure of subjects in the no assistance condition without the effect of inertial properties of devices on their power consumption. Exploring the MAF values of two different devices with the same effect on the metabolic power consumption of subjects not carrying and carrying (i.e., {\it noload} "Ee" and {\it loaded} "Ae") showed that the monoarticular exoskeleton with high torque capacity in the hip actuator could deliver assistance under the effect of its inertial properties.

Studying the selected cases from the optimal Pareto front curves confirms our claims on the overall performance of the monoarticular and biarticular exoskeletons and also shows that designing a monoarticular device requires a careful selection of actuators and their gear ratio to compensate the negative effect of the device on the metabolic power consumption of subjects and deliver assistance.

**\subsection\*{Study Limitations}**

This simulation-based study of an assistive device has some limitations that need to be considered for any interpretation of the results. One of the main limitations is kinematics and ground reaction forces for the assisted subjects. Although experimental studies reported that an exoskeleton could make minor \cite{42,79,91,114,115,116} and significant \cite{80,117,118,119} changes on assisted subjects’ kinematics and joint moment, the OpenSim current neural algorithm (i.e., CMC algorithm) does not capture these changes, and it was assumed that unassisted and assisted subjects have the same kinematics, ground reaction force, and joint moment. Nonetheless, it has been reported that metabolic cost may not substantially be affected by kinematics changes\cite{120}. This limitation recently has been addressed by employing dynamic optimization methods for performing simulations, which can capture the changes in the kinematics and dynamics of assisted subjects. However, , since altered kinematics can have several side effects such as increasing joint loads, the kinematic adaptation may not be desirable in some conditions \cite{93}.

Secondly, as was earlier stated, the assistive devices that we modeled were assumed to be massless without any actuator and link mass and inertia; however, in practice, exoskeleton actuation modules mass and their reflected inertia on the links are large and one of the main challenges on mechanical design of exoskeletons; it was also proven that adding mass to the lower limbs can considerably change the metabolic cost of the subjects. The exoskeleton attachments on the body is also one of the central performance limiting factors of assistive devices \cite{121}, which is not modeled in ideal exoskeletons.

Another significant limitation of this study is limitations on musculoskeletal modeling. sSome influential restrictions on muscles modelingaffect assistive device simulation results. One of these restraints is extortionate passive force generated by the muscles \cite{92,93}, which can result in extortionate muscular activities, which was observed in similar work \cite{93} comparing simulation and experimental muscular activities. Another critical issue in Hill-type muscles modeling is that it does not take into account muscle fatigue, which is an effective factor in muscle recruitment strategies \cite{92}. Rectus femoris, which is more vulnerable to fatigue due to its fiber properties \cite{123} experienced extreme activations in all of the assistance scenarios, which in practice may cause subjects muscle fatigue\cite{122}. Tendon modeling, constant force enhancement, short-range muscle stiffness, training effect, and other factors \cite{92} are limiting factors of the muscles and models that affect the musculoskeletal models and simulations results which need to be considered for any interpretation of this study's results. Apart from all these general limitations, the dataset and models of unassisted subjects used in this study had some inconsistency with the experimentally collected data affecting some of the results and we would refer to \cite{93}, which discussed that study’s limitations.

Furthermore,\cite{92} provides comprehensive information about all aspects of the OpenSim simulations and proposes some recommendations for any interpretation and validation of the simulation results, which that can be beneficial in obtaining an accurate interpretation of our results. It is not feasible to expect to obtain a close quantitative match between the results of our simulations and experiments without acknowledging the discussed limitations and other practical matters.

**\section{Future work}**

In consideration of our study limitations, the monoarticular and introduced biarticular exoskeletons can be modeled in simulators with dynamic optimization neural algorithms \cite{110,111,112,180} by considering their inertial properties to study their effect on the power expenditure of subjects’ muscles activity and how adding inertia can affect the torque and power profiles while performing different tasks.

Simulations based on the Pareto front had limitations as highlighted in the previous section, which should be addressed in future work. The search large discretization might well be be addressed using the normal boundary intersection method \cite{108}, which is designed to resolve these issues on computationally expensive problems, resulting in a more accurate Pareto front with fewer discretization problems for the subjects.

Simulation outcomes are beneficial as prior information to assist the subjects, and they can be used on the human in the loop (HIL) optimization \cite{109} as a prior profile to start optimization with the torque profiles of simulations, which may result in less optimization time by increasing the convergence rate of the optimization. We are planning to establish experimental setups and partially validate our results using the outcomes of the experiments. Although we do not anticipate obtaining a quantitative match between the results of the simulations and experiments due to the discussed limitations, we expect to obtain qualitative matches between these results. Along with these confirmations, muscle fatigue, muscle activities, and training effects that could not be addressed through simulations can be assessed through the experiments.

**\section\*{Conclusions}**

In this study, a novel biarticular exoskeleton assisting the hip and knee was proposed and compared to another typical monoarticular exoskeleton. We then modeled the mechanism of the proposed exoskeleton through a musculoskeletal simulator to study the performance of the device and its difference from a monoarticular exoskeleton and analyze its effect on the assisted musculoskeletal models.

In the first phase of this study, we conducted simulations with ideal exoskeletons to verify the modeling and to study the performance of devices under the ideal condition. These simulations showed that despite the same metabolic reduction effect, the devices have different power consumption, and the monoarticular device was affected more considerably by loading the subjects with a heavy load. Additionally, we showed that loading subjects with a heavy load changed the profiles of devices only bye magnitude and timing, and the trajectories did not change considerably. We showed that the devices affected the hip abduction and ankle indirectly by analyzing the effect of devices on the activation of muscles. Finally, joint reaction load analysis revealed that assistive devices considerably affected the reaction forces and moments of assisted and unassisted joints, and the kinematic difference between the biarticular and monoarticular exoskeletons resulted in their different effect on the reaction loading of the knee joint.

We organized another stage in this study in which we introduce a novel Pareto simulation framework to conduct fair comparisons among different optimal configurations of the exoskeletons by taking advantage of Pareto optimization methods. Through this phase of the study, we showed that both constrained devices could provide practically the same assistance delivered by the ideal exoskeletons using lower power in both actuators. Although both devices demonstrated similar optimal trade-off curves, we showed that the optimal configurations of these devices were considerably different and had different effects on the muscle activations of assisted subjects. Additionally, it was shown that optimal monoarticular exoskeletons tended to operate in the high peak torque regions compared to the biarticular exoskeleton, especially on the hip joint. The biarticular device showed more robust performance by loading subjects than the monoarticular device. The monoarticular hip actuator demonstrated high within optimal devices and between load condition variations, which complicates designing a general device. Moreover, through joint reaction forces and moments analyses, we demonstrated the resemblance of the reaction forces and moments of subjects assisted with constrained devices to reaction forces and moments of subjects assisted with ideal devices. Finally, the analysis of regeneratable power helped us gain insight into how regeneration could considerably affect the power consumption of devices.

The last stage of this study was dedicated to investigating the effect of inertial properties of optimal devices (i.e., optimal solutions on Pareto front curves) on the metabolic expenditure of assisted subjects on the metabolic rate of assisted subjects. To conduct this study, we adopted the model developed to estimate the metabolic rate of subjects mounted with the additional masses and performed offline simulations. Along with these simulations, we proposed a modification of the augmentation factor to measure metabolic power change due to carrying the exoskeleton to assess the performance of the optimal device. Through this phase of the study, we showed that both devices were substantially affected by considering their inertial properties, and this effect was more severe on the monoarticular device due to its kinematic design. Although high torque was required for the monoarticular exoskeleton to have a positive effect on the subjects, none of the high torque devices showed optimal performance, which complicates designing an efficient monoarticular device. We showed that keeping the knee actuator near to the hip joint or grounding it to the shank instead of the thigh can notably improve the performance of the monoarticular device.