**\subsubsection\*{Devices Speed, Torque and Power}**

One of the main objectives of this section is to validate the kinematic modeling of the exoskeletons on the OpenSim. As it is already discussed, there is a linear mapping between the monoarticular and biarticular exoskeletons, and if the modeling of the devices were correct in the musculoskeletal simulator, this kinematics relation requires to be held.\\

Figure \ref{Fig\_IdealExo\_Speed} represents the velocity profiles of the biarticular and monoarticular exoskeletons in both load conditions. From the Eqn. \eqref{Eqn\_Mono\_Bi\_Jacobian} we were expecting to have the same angular velocity profiles at the hip actuator, and since the hip actuator in both exoskeletons were supposed to be attached directly to the hip joint, the velocity of the joint and actuators in both devices have practically the same profiles as is shown in Figure \ref{Fig\_IdealExo\_Speed} for the hip joints in both load conditions.\\

\begin{figure\*}[ht]

\end{figure\*}

The main difference between the two configurations of the exoskeleton is on the knee joint in which the biarticular device assists the joint through a parallelogram mechanism, and the velocity profiles of the knee actuator were supposed to be different according to their jacobian. This difference can be seen in Figure \ref{Fig\_IdealExo\_Speed} in which the monoarticular knee actuator follows the knee joint velocity profile, but the biarticular actuator is showing different profile than the knee joint profile due to Eqn. \eqref{Eqn\_Mono\_Bi\_Jacobian}. Analyzing the velocity profiles of the devices validating the mapping between two exoskeletons and modeling of them through the OpenSim.\\

According to the jacobian between these two devices expressed in Eqn. \ref{Mono\_Bi\_Torque\_Mapping}, the knee and hip actuators were expected to have the same and different torque profiles respectively, which is evident in Figure \ref{Fig\_IdealExo\_Torque} for both \textit{loaded} and \textit{noload} conditions. The generated optimal torque profiles of the ideal exoskeletons did not resemble the net moment of the assisted joint, which was also observed by \cite{93,2} for the simulation-based study of the walking with heavy load and running respectively. The torque of assistive actuators in both hip and knee joints exceeded the corresponding net joint moment and resulted in opposing muscles generated moment and device torque in the joint.\\

\begin{figure\*}[ht]

\end{figure\*}

This opposition was more significant on the knee joint than hip during the mid-stance to the mid-swing phase, with the highest opposition on the onset of the pre-swing phase. The hip joint had significant actuator and muscle torque opposition during the pre-swing to terminal swing phases, indicating that different than the knee joint in which major portion of antagonism occurred during the stance phase, the hip got into muscle and actuator torque contraction during swing phase. \\

The analysis of the torque profiles of a device, represented in Figure \ref{Fig\_IdealExo\_Torque}, in different load conditions indicates that the loading subject with a heavy load does not result in substantial changes in the torque profiles of the assistive devices. The main changes between \textit{loaded} and \textit{noload} conditions are the timing and magnitude of the profiles, which is due to the change of the joints kinematics and kinetics.

Nevertheless, the standard deviation of assistive devices and assisted muscles generated torques are considerably greater in the \textit{loaded} condition, and it is more evident in the knee joint where the net joint moment has a remarkable deviation during the stance phase. This high within-subject deviation of torque profiles indicates that the assistance of subjects carrying heavy load requires the subject-specific design and control of the exoskeletons \cite{2}.\\

\begin{figure\*}[ht]

\end{figure\*}

Due to the discussed kinematic differences between two configurations, the power profiles of the exoskeletons were different in both actuators, as is represented in Figure \ref{Fig\_IdealExo\_Power}. The profiles of power consumption of the biarticular actuators are different during the gait cycle except in the loading response and partially in mid-stance phases. The load carried by subjects causing different timing and magnitude than subjects walking with no load and the deviation of the profiles are higher for the \textit{loaded} subject, which both are observed in torque profiles as well.\\

Although the power profiles of hip actuators were roughly following the net joint power profile, the knee actuator profiles did not resemble the knee joint power. The mechanical work performed by the assistive devices were mostly positive work for both knee and hip actuators. The negative mechanical work in the biarticular exoskeleton can be harvested mostly during the initial-swing and mid-swing phases for the knee actuator and terminal phase for the hip actuator. Unlike the biarticular device, the monoarticular hip actuator performed practically no negative mechanical work, and the regeneratable work of the knee actuator is within both mid-stance and late-swing phases.

**\subsubsection\*{Effect of Devices on Muscle Coordination}**

The muscular activation of the subjects assisted by ideal assistive devices was considerably adjusted. Adding a set of ideal actuators with high optimal force (i.e., low penalization cost) to the musculoskeletal model changes the solution of the optimizer for finding a set of actuators to track the kinetics and kinematics of the joints.\\

Appending ideal actuators does not necessarily decrease the activity of all muscles, and it can be more economical for the complete set of actuators to increase the activity of specific muscles during some phases of a gait to decrease the activity of less cost-effective muscles. Since metabolic power of muscles is a function of their activity, and their fiber properties \cite{106}, the reduction in the activity of the entire set of muscles is resulting in gross whole-body metabolic cost reduction.\\

Despite the kinematic difference between the two configurations of the assistive device, the applied torques to the joints were practically identical, and it resulted in an identical effect on the muscular activation of the subjects.\\

\begin{figure\*}[ht!]

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\begin{figure\*}[ht]

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The cause for this issue rooted in the ideal nature of the actuators and devices, meaning that there are no constraints on the torque actuators can provide, the devices are assumed to be massless, and actuators do not have any reflected inertia effect. The muscle activation in Figure \ref{Fig\_IdealExo\_MusclesActivation}, which shows the effect of the biarticular device on the activation of representative muscles, is sufficient and can be generalized for both configurations of the assistive device. \\

The devices affected the activity of muscles of the lower extremity. This effect was significant on the Bicep femoris short head, Semimembranosus, and vasti muscles in which their activation was replaced by another set of actuators, including muscles and ideal actuators. The rectus femoris, which is a large knee extensor and a hip flexor biarticular muscle, was considerably increased during the stance phase. This increase occurred so that the optimizer could take advantage of the rectus femoris high force-generating capacity to exert hip flexion and knee extension moments more economically. In the meanwhile, this high muscular activity of the rectus femoris resulted in high knee extension and hip flexion moments exceeding net joint moment of the joints which was neutralized by ideal actuators that can extremely economical for applying high torques due to its high optimal force assignments.\\

This set of activation in which hip flexion and knee extension required moment could be applied by more cost-effective muscles and actuators, resulted in a substantial reduction in the activity of psoas and iliacus muscles as two major hip flexor muscles and the vasti muscles (vastus lateralis, vastus intermedius, and vastus medialis) as knee extensor set of muscles. The semimembranous muscle is another biarticular muscle contributing to hip extension and knee flexion moments, which was affected by the assistive devices, and the new set of actuation was practically replaced its activity. The medial gastrocnemius as a critical knee flexor and ankle plantar flexor muscle was influenced by the assistive devices in which its activity was substantially reduced, yet the muscle remained partially active to supply an ankle plantarflexion moment. The reduction of ankle plantarflexion moment was compensated by increasing the soleus activity as another primary ankle plantarflexor muscle. The assistive devices affected the activity of the gluteus medius muscles as well, which are not only responsible for a significant fraction of hip abduction moment, but also they contribute to hip motion in the sagittal plane as well and hip rotation.\\

The anterior and posterior portions of the gluteus medius muscle, besides their primary contribution to hip abduction, were supporting the hip extension and flexion and its lateral and medial rotations. Though, the contribution of these muscles to the hip sagittal moment replaced by assistive devices and a modified set of activations in assisted subjects, and it resulted in their muscular activity reduction.\\

The main differences between the muscular activity of the subjects walking normally and subjects walking while carrying a heavy load on the torso were the magnitude and timing of the muscular activations, which were observed already in other profiles as well. This load condition affected partially some muscles like Semimembranosus in which the muscles were not entirely replaced by the ideal devices.

\paragraph\*{Effect of devices on reaction forces moments of joints.} The change in the muscle coordination and augmenting assistive device to the subjects affected the reaction forces and moment of both assisted and unassisted joints. This relationship between the muscle activity and joint reaction forces has been proven through different literatures\cite{171,170,173,174}. The modified coordination of muscles in the ankle joint reduces the reaction forces and moments of the ankle in the swing phase while increasing them slightly during the stance stage as shown in Figure 1 and 2 in \nameref{S3\_Appendix}. The effect of muscle recruitment change was evident in the medial-lateral reaction force and extension-flexion reaction moment of the ankle. The study accomplished by Veen et al. \cite{170} shows that increase in the activation of rectus femoris and gastrocnemius muscles along with a decrease in activation of the soleus muscle can reduce the reaction forces of the ankle joint. Although assistive devices increased the activation of the rectus femoris, the effect of devices on the gastrocnemius and soleus muscles was not favorable to reduce the reaction force, especially during the stance phase. This coordination of muscles explains the behavior of reaction moments and forces of the ankle joint.\\

The effect of devices and altered muscle recruitment strategy on the reaction moments and forces of the patellofemoral and knee joint was substantial. The reaction forces of the patellofemoral and knee joints decreased during the early stance phase, and the increased during the late stance as shown in Figure 3 and 4 in \nameref{S3\_Appendix}. The analysis of muscle effect on the tibiofemoral forces showed that the hamstring muscles have a significant impact on the reaction forces of the knee during the early stance, while the gastrocnemius, rectus femoris, and iliopsoas muscles affecting the reaction forces during the late stance stage \cite{171,170}.\\

The increase in the activation of the soleus and decrease of the activation of hamstring muscles (i.e., semimembranosus, semitendinosus, and biceps femoris muscles) reduced the reaction force of the knee in early stance phase. During the late stance, we hypothesize that the substantial promotion and reduction of the rectus femoris and gluteus medius activities, sequentially, became dominant to the reduction of activities of other muscles and resulted in the tibiofemoral reaction force increase. Since the behavior of the other reaction force components in both the patellofemoral and knee joints was practically identical with the tibiofemoral performance, we think that the muscle arrangement had the same effect on tother reaction forces. However, since the hip muscles' effect on the knee reaction force was proven \cite{170,171}, this claim needs to be justified in a more isolated condition, such as assisting a joint condition. Although the reaction moments in both joints were roughly following the reaction forces' behavior, the effect of devices on the reaction moments was slightly different in which the biarticular exoskeleton was able to reduce the reaction moments and have lower peaks than the monoarticular device on the extension-flexion reaction moment which is represented in Figure 5 in \nameref{S3\_Appendix}.\\

Although the reaction forces and moments of the knee joint were increased during the late stance phase, the assistive devices were able to reduce the most of the maximum or peak reaction forces and moments on the knee joint. Additionally, the modified muscle recruitment effect on the reaction forces and moments during the swing phase was remarkably lower than its effect on the stance phase; nonetheless, the tibiofemoral force experienced considerable reaction force reduction during the swing phase compared to other reaction forces.\\

The reaction forces of the hip joint were affected by the activity of a group of muscles mentioned in \cite{170}, including the gluteus minus, gluteus medius, iliopsoas, and rectus femoris muscles. The increase in the activity of the rectus femoris incorporation with iliopsoas and gluteus medius muscle activity reduction decreased the reaction forces of the hip joint. This reduction was considerable during late stance, and early swing phases and subjects in {\it noload} condition were more substantially affected than the subjects in {\it loaded} condition as shown in Figure 6 and 7 in \nameref{S3\_Appendix}. \\

These modifications in the reaction moments and forces of the assisted subjects can improve the health of joint tissues \cite{178}. The large joint loads are identified as an essential factor of onsetting and progressing osteoarthritis \cite{172,176,177} and joint pain \cite{175} and reduction in the reaction forces and moments can prevent and decline these joint pain and arthritis onset and development.

**\subsection\*{Pareto Simulation Results}**

**\subsubsection\*{Optimal devices performance}**

The analyzed assistive devices in the previous section and their effect on metabolic cost and muscular activation of the subjects in two different load conditions were studied under the assumption that the assistive actuators have no bounds on the amount of the moment they can supply to the musculoskeletal model. However, this assumption does not a descriptive assumption for the real-time designing and controlling assistive devices because the devices and especially untethered exoskeletons have some constraints on the amount of moment that their actuation unit can provide to assist the joint of interest and the suppliable power from the battery for untethered exoskeletons is limited by the battery life. \\

One of the main intentions of Pareto simulations was addressing this limitation through a simulation-based study in which we can analyze the performance of assistive devices under the limitation of their actuators on providing torque to the joints.

The study was accomplished by constraining the maximum torque the assistive actuators can provide, and the optimal trade-off between the metabolic cost reduction and power consumption of the devices was obtained. The average Pareto front for the biarticular and monoarticular exoskeletons for both loading conditions of the subjects are represented in Figure \ref{Fig\_Main\_Paretofronts}.\\

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One of the immediate indications provided by the Pareto front curves is that the constrained devices in their maximum torque bounds where hip and knee restricted to 70 N.m peak torque, can mostly provide assistance that was provided by the ideal exoskeletons with lower power consumption in both load conditions (Table \ref{Table\_Device\_Performance\_Comparison}). Both constrained devices were able to assist the \textit{noload} subjects as much as the ideal exoskeletons were assisting, whereas the power consumption in the biarticular exoskeleton was considerably reduced compared to its ideal actuation, but the monoarticular energy consumption was practically the same with its ideal configuration.\\

The performance of devices under peak torque limitation and their comparison with the ideal devices implies that the comparison of assistive devices with unlimited actuation units is not a legitimate comparison suggesting that the comparison between assistive devices should be conducted using optimal trade-off points of devices in which a device has its optimal performance in both power consumption and metabolic cost reduction criteria.\\

\begin{table}[ht]

\end{table}

The biarticular and monoarticular exoskeletons show practically similar performance in assisting subjects in both {\it loaded} and {\it noload} conditions. Nevertheless, analyzing the power consumption of the devices on the Pareto front reveals that the monoarticular device considerably was affected by the load condition of subjects in which the performance of monoarticular exoskeleton became practically identical with the biarticular device when subjects were in \textit{loaded} condition, whereas the monoarticular device had superior performance in \textit{noload} condition.

The detailed analysis of the monoarticular device shows that both actuators of this device were affected by loading subjects with a heavy load (Figure \ref{Fig\_Main\_Paretofronts} and \ref{Fig\_Paretofronts\_Actuators\_EnergyBarPlot}), and unlike the \textit{noload} condition where the power consumption of the knee actuator was dominant to the hip in all optimal devices, loading subjects increased the amount of mechanical work performed by the hip actuator. In contrast, the power consumption of the biarticular knee and hip actuators was not affected noticeably by loading subjects as Figure \ref{Fig\_ \_Actuators\_EnergyBarPlot} representing the energy consumption of optimal devices. Additionally, the optimal configurations of the biarticular exoskeleton were mostly similar in the subjects walking while carrying a heavy load and walking normally, whereas monoarticular exoskeleton had different configurations in both load conditions.\\

The practically similar configurations and performances of the biarticular exoskeleton in both load conditions can facilitate designing a biarticular device and developing a generic controller to assist the subjects in different load conditions.\\

**\paragraph\*{Case studies.}** Studying and conducting comparisons between the {\it "Ad"} and {\it "Da"} configurations of the monoarticular and biarticular devices respectively in {\it loaded} condition in which they had practically the same performance in both optimization criteria. Nevertheless, analyzing the power consumption of actuators and comparing them shows that they had actuator recruitment, and there is a pairwise statistically significant difference between the power consumption of the hip and knee actuators, as shown in Figure 2 in \nameref{S3\_Appendix}. On the other hand, performing the same comparison between the {\it "Cb"} biarticular and {\it "Ba"} monoarticular in {\it noload} condition shows no significant difference between the power consumption of actuators as shown in Figure 5 in \nameref{S3\_Appendix}, indicating that the power consumption strategies of the devices change by loading the subjects. The performance of the same type of devices with the same power consumption or metabolic cost reduction in different load conditions also has been studied in Case 3 and Case 4 in \nameref{S3\_Appendix}. These cases along with the other two cases, comparing the monoarticular and biarticular exoskeletons in different load conditions, have been discussed comprehensively in \nameref{S3\_Appendix}.