In the inertia position factor represented in Eqn.\eqref{Eqn\_Inertia\_Factor}, $A\_i$ is the multiplier of $I\_{ratio}$ in Browning models for foot, thigh and shank segments, and $I\_{unloaded}$, $m\_{subjects}$, and $MC\_{unloaded}$, which were obtained from the \cite{133} models and plots, are the inertia of the leg without any external load, the mean weight of subjects, and the metabolic rate of subjects walking without any load on their segments, respectively. The modified augmentation factor (MAF) was obtained by adding the effect of inertia on the metabolic detriment part of this factor and can be expressed as Eqn.\eqref{Eqn\_MAF}.

\begin{align}

\end{align}

where $p^{+}$, $p^{-}$, and $p^{disp}$ represent mean positive, negative, and dissipated power calculated through Eqn.\eqref{Eqn\_disp\_power}. The $\beta\_i$ in MAF stands for the location factor of the device mass, which is 14.8, 5.6, 5.6, and 3.3 $W/kg$ from the foot to waist, respectively \cite{41,133}; the $\gamma\_j$ represents the location factor of the device inertia, which is 47.22, 27.78, and 125.07 $W/{kg.m^2}$ from the foot to thigh, respectively. Consistent with augmentation factor procedure \cite{41}, MAF uses muscle-tendon efficiency,$\eta = 0.41$ to convert the mechanical assistive power to metabolic power determined empirically by Sawicki and Ferris\cite{149} and Malcolm et al.\cite{40}. Finally, we normalized modified augmentation factor by the weight of each subject.

**\subsection\*{Validation of Simulations}**

The comprehensive validation procedure of the OpenSim simulations was published by Hicks et al. \cite{92, which explained how to validate modeling and simulation results at each stage. Additionally, Dembia et al. \cite{93} explained simulation verification for their simulations of assistive devices. We followed the same procedures explained in \cite{92,93} to validate our results from the simulations.\\

As was already discussed, the adjusted model, adjusted kinematics, and processed ground reaction forces, which had previously been evaluated and validated by Dembia et al. \cite{93,were used to accomplish this study. The muscular activation resulting from the simulations of unassisted subjects was validated with experimentally recorded electromyography (EMG) signals \cite{92,93}, in which there were some timing and magnitude discrepancies between simulated and experimentally collected activation of some muscles due to excessive passive forces in knee and ankle joints.\\

The \textit{loaded} and \textit{\textit{noload}} joint kinematics and kinetics were compared with the results of the studies accomplished by Huang and Kuo \cite{131} and Silder et al.\cite{132} and validated qualitatively. Since our simulations of the unassisted subject for \textit{loaded} and \textit{noload} conditions were the same as those of the Dembia et al. simulations, we reproduced their simulations and compared them with their results to validate the results of our reproduced simulations. Additionally, since we used the provided RRA results to perform the CMC simulations, the joint moment and joint kinematics represented in this paper were already validated.\\

The other source of error during simulations is kinematics error, which was analyzed to be within the recommended thresholds by Dembia et al.\cite{93}. Since the inverse kinematics stage of the simulation was not reproduced in this study, the markers error was not examined, and we relied on the previously performed verification of this error source. The analysis of Dembia et al. \cite{93} on residual errors showed that the residual forces lie below the threshold recommended by Hicks et al. \cite{92}; however, the residual moments exceeded the Hicks et al. thresholds, but since the joint moments matched with \cite{92}, it was claimed that these exceeding residual moments do not affect the interpretations \cite{131,132}.

Another error source in these simulations could be additional moments introduced to compensate for any unmodeled passive structures and muscle weakness, which were checked to confirm that they were within their recommended thresholds of less than 5\% of net joint moments in peak and RMS \cite{93}.\\

To ensure that our simulations in both ideal and Pareto-optimization phases did not deviate from the defined error source thresholds, we analyzed the kinematics of all simulations and checked their divergence from the adjusted kinematics resulting from the RRA simulations. Additionally, some simulations of the Pareto-optimization part were selected randomly, and their residual and reserve moments and forces were analyzed.

**\subsection\*{Performance Metrics}**

For the purpose of attaining a methodical analysis and comparison between the assistive devices and load conditions, some performance metrics were defined. As discussed earlier, the ultimate goal of each assistive device is to reduce the energy consumed by the subject in performing a task, which in this study was walking with and without a/the load at self-selected speed.

Therefore, the normalized gross whole-body metabolic rate of each subject in two different assistive devices and load conditions was calculated, and then metabolic cost reduction was computed using the metabolic rate of assisted and unassisted subjects. This procedure of metabolic cost calculation was repeated for all seven subjects in three trials to obtain the total average metabolic energy expenditure and metabolic cost reduction for each assistance scenario and load condition.\\

Another important metric for assessing the performance of an exoskeleton is the power it consumes to assist a subject. To analyze this metric in the simulated exoskeletons, we computed the absolute power consumed by all actuators; the reason for considering the absolute value is the absence of the regeneration mechanism as a general case for the exoskeletons. Additionally, to analyze the amount of power available for regeneration, we computed the negative energy of the exoskeletons. The power consumption metric is a crucial part of an exoskeleton analysis to estimate the efficiency and battery life for untethered devices.

In the Pareto optimization part of the study, these two introduced metrics (i.e., metabolic cost reduction and power consumption) were optimized simultaneously, and their set of optimal solutions was represented for each device and load condition. This Pareto front represents the set of different configurations for each exoskeleton in terms of power consumption and the assistance the device can provide. For more detailed analyses, some specific configurations in each exoskeleton and load condition were selected and studied exhaustively.\\

Additionally, the regeneratable power and inertial properties of the exoskeletons, as two critical metrics for analyzing the performance of assistive devices, were studied in simulations of multi-criteria optimization by reflecting their effect on the Pareto-curves and obtaining the Pareto sets under the effect of these two metrics.

Muscular activations of the lower limb key muscles were extracted and studied to gain insight into how an assistive device can change muscle activities and, consequently, metabolic cost resulting from muscle activities.

Lastly, the reaction forces and moments of the joints were computed to gain insight into the effect of assistive devices on the reaction forces of the joints and understand how the proposed devices can change the reaction forces and moments of the joints.

**\subsection\*{Statistical Analysis}**

To conduct methodical comparisons among scenarios with the discussed metrics, we employed statistical analyses. Since the simulations were performed on seven subjects with three trials in five different scenarios, repeated measures analysis of variance (ANOVA) and Tukey Post-hoc were applied to test the statistically significant difference between the selected metrics and scenarios. Additionally, for the Pareto simulations, statistical analyses were performed for the specific points selected from the Pareto front for further investigation. Nevertheless, the standard deviations of all points on the Pareto front were explicitly plotted for both criteria. We used a significance level of 0.05 and SPSS \cite{spss} to perform the tests.

**\section\*{Results and Discussion}**

**\subsection\*{Ideal Exoskeletons Results}**

**\subsubsection\*{Device Performance}**

Both biarticular and monoarticular configurations of the ideal exoskeleton reduced the metabolic energy consumption of subjects walking without and with carrying a heavy load at self-selected walking speed. The biarticular and monoarticular exoskeletons decreased the metabolic rate of subjects carrying a heavy load by 20.49$\pm$2.87\% and 20.45$\pm$2.81\%, respectively. The monoarticular and biarticular configurations of the exoskeleton were able to reduce the gross whole-body metabolic cost of the subject in the \textit{noload} condition by 22.38$\pm$4.91\% and 22.47$\pm$4.89\%, respectively. These exoskeletons were expected to have the same performance on reducing the metabolic energy consumption of the subjects due to their kinematic relation, and the results represent an expected performance for these two devices.\\

\begin{figure\*}[ht]

\end{figure\*}

The assistance of both exoskeletons on subjects carrying a heavy load was able to compensate for their demanding metabolic energy to carry the heavy load. As can be seen in Figure \ref{Fig\_IdealExo\_Energy\_BoxPlot}, the assisted subjects in the \textit{loaded} condition have a comparable metabolic rate with unassisted subjects who do not carry any load, meaning that the best both ideal exoskeletons can do is to partially compensate for the metabolic power demanded by the \textit{loaded} subject to a/the subject without any load. Nonetheless, the metabolic demand of the assisted \textit{loaded} subjects have a statistically significant difference.\\

The power consumption of the monoarticular and biarticular devices showed significant pairwise differences between the hip and knee actuators in the {\it noload} condition, indicating their different power demands for delivering assistance to the subjects. On the other hand, we did not observe a significant pairwise difference between the actuators of devices in the {\it loaded} condition, which shows the high variation in power consumption of {\it loaded} subjects.

Additionally, examining the power consumption of devices within load conditions showed that the hip actuator of the monoarticular exoskeleton was considerably affected by loading subjects, and the hip actuator consumed significantly more power than the knee actuator when subjects were loaded. On the other hand, the mechanical work on the biarticular exoskeleton was distributed uniformly between the actuators, and the load condition difference did not show any significant differences.\\

**\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 was already discussed, there is a linear mapping between the monoarticular and biarticular exoskeletons, and if the modeling of the devices is correct in the musculoskeletal simulator, this kinematics relation must be held.\\

Figure \ref{Fig\_IdealExo\_Speed} represents the velocity profiles of the biarticular and monoarticular exoskeletons in both load conditions. From Eqn. \eqref{Eqn\_Mono\_Bi\_Jacobian} we expected to have the same angular velocity profiles at the hip actuator, and since the hip actuators 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 shows/exhibits a 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 walking with a 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 the hip joint 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 from the knee joint, in which a major portion of antagonism occurred during the stance phase, the hip got into muscle and actuator torque contraction during the swing phase. \\

The analysis of the torque profiles of a device in different load conditions, represented in Figure \ref{Fig\_IdealExo\_Torque}, 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 the \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.