\subsection\*{Contributions}

In this study, we introduce a bio-inspired biarticular configuration of an exoskeleton assisting hip and knee joints and delivering torque to the knee joint distally. We modeled the proposed exoskeleton through a musculoskeletal simulator framework and compared it to a typical monoarticular device that assists each joint directly by mounting an actuator to the joint. Modeling these exoskeletons through the musculoskeletal simulator framework enabled us to get insight into not only the moment and power profiles of exoskeletons and their power consumption, but also the effect of these assistive devices on the muscle activity, energy expenditure, joint reaction forces, and stiffness of simulated subjects.\\

At the first stage of this investigation, the simulation-based study was conducted with ideal exoskeletons in which devices had no mass, inertia, or any limiting factor, and the actuators of these devices had no constraints on delivering torque to the joints. This type of simulation was employed to verify the purposed exoskeleton modeling on the musculoskeletal simulator and gain insight into the performance of these devices, and their effect on the assisted musculoskeletal models under ideal conditions.\\

This phase of the study showed that despite the same effect of both devices on the metabolic rates of subjects, the devices have different profiles and power consumptions. These simulations showed that loading subjects with a heavy load mainly changes the profiles by magnitude and time-shifting, which depends on the toe-off time. It was also shown that the biarticular hip actuator has a more uniform work distribution in different load conditions, and it was less affected by loading subjects. Studying the muscular activities of the assisted subjects showed that the muscular activations of both assisted and unassisted degrees of freedom are also affected by augmenting ideal devices. Through the joint reaction loading analysis on this phase of simulations, it was observed that the assisted subjects have considerably different reaction loadings in all degrees of freedom of all joints of the lower extremity, and interestingly, despite the same effect of devices on muscles and metabolic rates of assisted subjects, they have a different impact on the reaction loadings of the knee and patellofemoral joint. Finally, the stiffness analysis showed that the assistive actuators have utterly different stiffness properties and impact on the stiffness of joints.\\

Although studying the exoskeletons without any restriction on their performance provides handy erudition about the exoskeletons, it is necessary to analyze and compare them in more realistic conditions that can be applied in real-time applications. Consequently, we organized another stage for this study in which we introduced a novel Pareto simulations framework to conduct reliable comparisons among different configurations of the exoskeletons in their optimal configurations by taking advantage of Pareto optimization and Pareto filtering methods and implementing them into the musculoskeletal framework. This designed method can provide average optimal trade-off cases for each configuration of the assistive devices based on the defined optimization objectives, which were defined as metabolic cost reduction and device power consumption as two optimization criteria in this study. The Pareto simulations and optimal devices resulting from this method can be employed to conduct fair comparisons, and it was used to study and compare the assistive devices with constraints on their maximum provideable torque to the joints.\\

Through this phase of the study, we showed that in both loading conditions, configurations exist for both devices that can provide the assistance of ideal exoskeletons with lower energy consumption. This finding implies that performing Pareto simulations to conduct fair comparisons of the performance of different devices is inevitable. The second phase of the simulations with constrained devices showed that although both devices have practically the same effect on the metabolic cost of subjects, the optimal configurations of these devices are considerably different and have different effects on the muscle activations of assisted subjects and the similarity between devices in ideal conditions does not hold for constrained devices. Notwithstanding the same total metabolic rate, further analyzing and comparing the metabolic rate of each muscle showed that the effect of these devices on the trends and rates of some muscles differed throughout the optimal trade-off curve.\\

Analyzing the power consumption of optimal devices resulting from the second stage revealed that the devices have remarkably different actuator arrangements for delivering the same amount of assistance to the subjects. In the subsequent analysis of optimal devices, it was shown that both devices have considerably different moment and power profiles in comparison with each other and their ideal configurations. Additionally, it was revealed that the monoarticular exoskeleton has high variations within optimal devices and between load conditions , which can considerably complicate the controlling of the device in different conditions. Studying the regeneratable power from the devices in different efficiency conditions showed the essence and importance of designing a regeneration mechanism and battery on the overall performance of the device. Through joint reaction force and stiffness analyses, it was shown that constraining devices affect the loading of the joints and their stiffness and that they do not resemble the loading and stiffnesses of joints assisted with ideal devices.\\

This study has been accomplished using OpenSim \cite{89}, which cannot simulate any dynamic variations without capturing their effect experimentally due to its neural control algorithm\cite{92}. To investigate the effect of the inertial properties of optimal devices, obtained from the Pareto simulations, on the metabolic rate of assisted subjects, we adopted the model developed by Browning et al. \cite{45} and performed offline simulations to investigate the effect of the inertial properties of optimal devices on the optimal trade-off curves of devices and assisted subjects’ effort.

Additionally, we introduced a modified augmentation factor as a modification of the augmentation factor developed by Mooney et al. \cite{41} as a general exoskeleton performance metric framework. The augmentation factor was modified by combining the effect of inertias along with masses on the performance of devices, and it was used to study the efficiency of selected optimal devices obtained by Pareto simulations under consideration of their inertial properties.\\

The last stage of this study stepped forward toward providing more realistic results from the simulations. This phase of the study revealed that the performance of devices is profoundly affected by their inertial properties; the effect was more severe for the monoarticular device due to its kinematic design. Analysis of optimal devices under their inertial properties effect using developed model, and the modified augmentation factor showed that most of the optimal biarticular exoskeletons were able to preserve their optimal performance under their mass and inertia effect. This eases designing an optimal device. Unlike the biarticular exoskeleton, the developed model, along with the modified augmentation factor, showed that the optimality of monoarticular devices was profoundly altered by reflecting the effect of their inertial properties. To improve the performance of the monoarticular exoskeleton, we suggested two monoarticular designs different from its conventional design that can mitigate the effect of a device’s inertial properties. Finally, the performance of the devices was analyzed under regeneration, revealing the promising effect of regeneration on the optimality of devices.\\

The paper is organized as follows: the Kinematic Modeling section presents the kinematics of the proposed biarticular and the monoarticular exoskeletons mechanism and explicates the relationship between these two assistive devices. The Musculoskeletal Simulation section begins by discussing the musculoskeletal model used for performing simulations in the Opensim framework, then explains the procedures of the simulations and analyses performed to obtain the studied criteria before discussing the modeling and simulation of assisted subjects. The Pareto simulation subsection is dedicated to explaining this method's workflow on the Opensim framework and the investigated objective on this phase of the study, followed by the Assistive Devices Inertial Properties Effect subsection, organized to explain the developed metabolic model of adding mass and inertia and modified augmentation factor. Lastly, we discuss the validation of performed simulations, defined performance metrics, and present models and methods of statistical analyses, which concludes the Musculoskeletal Simulation section.\\

The results and their discussion have been separated into three main subsections in the Results and Discussion section, including Ideal Exoskeleton Results, Pareto Simulation Results, and Optimal Devices Inertial Properties Effect subsections. The Ideal Exoskeleton Results subsection presents and discusses the results of simulations performed based on exoskeletons in ideal conditions without any constraints on their performance. Next, the Pareto Simulation Results discusses the optimal trade-off curves or Pareto front of simulated devices and conducts some comparisons between ideal and constrained devices along, within, and between constrained device comparisons. The Optimal Devices Inertial Properties Effect subsection discusses the effect of devices’ inertial properties and regeneration on optimal trade-off curves. The Results and Discussion section concludes by discussing the general shortcomings of simulation-based studies and the specific limitations of the study we conducted. Finally, the Conclusions and Future Work section concludes the paper

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\section\*{Kinematic Modeling}

The biarticular exoskeleton was designed to assist hip and knee joints. The exoskeleton was inspired from the biarticular muscles and their functionality and the aim of the design was keeping the large portion of the device weight around proximal joint (Hip) while delivering the required power to distal joint (Knee). The parallelogram mechanism has been purposed to accomplish this goal and take advantage of biarticular muscles biological features in the exoskeleton. The purposed assistive device is shown in figure \ref{Fig\_Exos\_Kinematics\_Model}\subref{Fig\_Biarticular\_Exo\_Mechanism}.\\

\begin{figure\*}[h!]

\centering

\subfloat[\small{biarticular device}]{\includegraphics[width=2in]{Cartoons/Biarticular\_Exo\_Mechanism.pdf}

\label{Fig\_Biarticular\_Exo\_Mechanism}}

\hfil

\subfloat[\small{monoarticular device}]{\includegraphics[width=2in]{Cartoons/Monoarticular\_Exo\_Mechanism.pdf}

\label{Fig\_Monoarticular\_Exo\_Mechanism}}

\vspace{1mm}

\caption{\small{\textbf{Assistive devices kinematics model.} The parallelogram mechanism has been used to model the biarticular exoskeleton and the monoarticular exoskeleton modeled by two link serial manipulator.}}

\label{Fig\_Exos\_Kinematics\_Model}

\end{figure\*}

The monoarticular exoskeleton can be modeled as a two-link serial manipulator as shown in figure \ref{Fig\_Exos\_Kinematics\_Model}\subref{Fig\_Monoarticular\_Exo\_Mechanism} where each joint was assisted by the directly joint actuator. The kinematics modeling of the monoarticular and biarticular exoskeletons in both configuration and motion level has been represented in \nameref{S1\_Appendix}.\\

As it can be interpreted from the kinematics of exoskeletons represented in \nameref{S1\_Appendix}, a linear mapping between monoarticular and biarticular exoskeletons can be established to relate these two device through a linear jacobian as it is represented in Eqn \eqref{Eqn\_Mono\_Bi\_Jacobian}.

\begin{equation}\label{Eqn\_Mono\_Bi\_Jacobian}

\begin{aligned}

\omega\_{2\times 1, \mathrm{monoarticular}} &= J\_{2\times 2}\omega\_{2\times 1, \mathrm{biarticular}}\\

\left\lbrack \begin{array}{c}

{}^{\mathrm{torso}} {\omega\_{\mathrm{mono}} }^{\mathrm{femur}} \\

{}^{\mathrm{femur}} {\omega\_{\mathrm{mono}} }^{\mathrm{tibia}}

\end{array}\right\rbrack &=\left\lbrack \begin{array}{cc}

1 & 0\\

-1 & 1

\end{array}\right\rbrack \left\lbrack \begin{array}{c}

{}^{\mathrm{torso}} {\omega\_{\mathrm{bi}} }^{\mathrm{femur}} \\

{}^{\mathrm{torso}} {\omega\_{\mathrm{bi}} }^{\mathrm{tibia}}

\end{array}\right\rbrack

\end{aligned}

\end{equation}

Using Eqn.\eqref{Eqn\_Mono\_Bi\_Jacobian} which is a mapping between the angular velocities of the exoskeletons, we can derive the mapping between the provided torque by exoskeletons as shown in Eqn. \eqref{Mono\_Bi\_Torque\_Mapping}.

\begin{equation}\label{Mono\_Bi\_Torque\_Mapping}

\begin{aligned}

\tau\_{2\times 1,\;\mathrm{biarticular}} &=J^T \tau\_{2\times 1,\;\mathrm{monoarticular}}\\

\left\lbrack \begin{array}{c}

{\tau^{\mathrm{torso}/\mathrm{femur}} }\_{\mathrm{bi}} \\

{\tau^{\mathrm{torso}/\mathrm{tibia}} }\_{\mathrm{bi}}

\end{array}\right\rbrack &={\left\lbrack \begin{array}{cc}

1 & 0\\

-1 & 1

\end{array}\right\rbrack }^T \left\lbrack \begin{array}{c}

{\tau^{\mathrm{torso}/\mathrm{femur}} }\_{\mathrm{mono}} \\

{\tau^{\mathrm{femur}/\mathrm{tibia}} }\_{\mathrm{mono}}

\end{array}\right\rbrack\\

\end{aligned}

\end{equation}

This relation between two exoskeleton has been used to verify the modeling of the exoskeleton through musculoskeletal simulation framework.

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\section\*{Musculoskeletal Simulation}

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\subsection\*{Musculoskeletal Model}

The exoskeletons have been studied through musculoskeletal simulations by conducting the simulations of the seven subjects walking normally and while carrying a 38kg load on the torso at their chosen speed. The data that has been used in this study was experimentally collected and processed by Dembia et.al. \cite{93} and their experimental protocol was approved by the Stanford University Institutional Review Board \cite{93}.\\

The musculoskeletal model used in the simulations, which was the same with the model used by dembia et al. \cite{93}, was a three-dimensional model developed by Rajagopal et al. \cite{130} with 39 degrees of freedom where the lower limbs were actuated using 80 massless musculotendon actuators, and the upper limb actuated by 17 torque actuators\cite{130}. \\

This three-dimensional musculoskeletal model was adapted by locking some unnecessary degrees of freedom for both normal walking and walking with a heavy load scenarios and modeling the extra load on the torso of the musculoskeletal model for the walking with heavy load condition \cite{93}.\\

Since this research was built upon the study performed by Dembia et al., we will follow the similar terminologies in most of the cases to avoid any confusion for the readers. Therefore, the \textit{loaded} condition will refer to the subjects walking while carrying the 38Kg load on their torso while the \textit{noload} condition will reference the subjects walking without any extra load at their self chosen speed.

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\subsection\*{Simulation Procedure}

%TODO: The professor asked me to have more detailed explaination on metabolic model.

The first step for conducting the simulation of the specific subject is scaling the generic dynamic model to acquire a musculoskeletal model matching with the anthropometry of the subject which was performed using OpenSim Scale Tool and according to the mass and height of the subject, the maximum isometric forces of the muscles were scaled. After obtaining the specific model for the subject, inverse kinematics of the subject were computed using OpenSim Inverse Kinematics Tool and the motion capture data collected experimentally.\\

On the next stage of the simulation workflow, the scaled model, inverse kinematics and ground reaction forces were employed to run the RRA algorithm\cite{103}. The RRA algorithm reduces the incompatibility of experimental data including ground reaction forces and trace data and musculoskeletal model by slightly adjusting inertial properties and kinematics. Then adjusted model and kinematics generated by RRA were employed to perform muscle driven simulations using Computed muscle control algorithm in OpenSim\cite{104}.\\

Computed Muscle Control (CMC) algorithm simulates the muscle recruitment of the subject by resolving muscle redundancy problem using static optimization to find the required muscle excitations to track the provided kinematics. The CMC simulations output were then used to run the analysis tool of OpenSim to compute subjects metabolic power consumption, and muscles moment.\\

\begin{figure\*}[ht]

\includegraphics[width=\linewidth]{Cartoons/OpenSim.pdf}

\vspace{-5mm}

\caption{\small{\textbf{Opensim simulation procedure block diagram.} The workflow of the simulation in OpenSim has been shown briefly in which green blocks stands for output, blue blocks are OpenSim simulations or analyses, purple blocks are models that have been used for simulations and analyses, and finally, red blocks represent processed experimental data.}}

\label{Fig\_OpenSim\_Sim\_Procedure}

\end{figure\*}

The OpenSim solves the muscle redundancy problem to track experimentally measured motion and uses effort-based objective, as Eqn.\eqref{Eqn\_CMC\_Objective}, to solve for a set of muscle excitations to track measured motions and forces within a specified tolerance using static optimization at each time step during the motion of interest\cite{92}. Therefore, the kinematics and dynamics of the subject will remain consistent during the simulations and any additional mass and inertia on the subject that has not been captured by experiments will cause a systematic error on the results.\\

\begin{equation}\label{Eqn\_CMC\_Objective}

J = \sum\_{i\in nMuscles} a\_{i}^{2} + \sum\_{i \in nReserves} (\frac{\tau\_{r,i}}{w\_{r,i}})^2

\end{equation}

With the knowledge of the OpenSim neural control algorithm, we used the adjusted model and kinematics provided by Dembia et al.\cite{93} instead of reproducing all data from the beginning of the simulation procedure which also helped us to ease the verification of the simulations procedure thanks to \cite{93} for verified simulations data.\\

\paragraph\*{Metabolic Model.} To calculate the estimation of the instantaneous metabolic power of subjects, Umberger \cite{105} muscle energetic model which was modified by Uchida et al. \cite{106} were employed in which average power consumption of a muscle during a gait cycle was calculated using Eq.\eqref{Eqn\_avg\_muscle\_power} \cite{106}.\\

\begin{equation}\label{Eqn\_avg\_muscle\_power}

P\_{avg} = \frac{m}{t\_1 - t\_0}\int\_{t\_0}^{t\_1} \dot{E(t)} dt

\end{equation}

Where m is muscle mass, and $\dot{E(t)}$ is the normalized metabolic power consumed. This model generates metabolic power of all muscles and then whole body metabolic power was calculated by summing all muscles metabolic power \cite{106}. For computing the gross metabolic energy consumption of subjects, we integrated the metabolic power over the gait cycle and then divided by the mass of subjects.\\

As it is mentioned in \cite{93}, due to experimental data insufficiency, some subjects and trials simulation were not a complete gait cycle, therefore, the metabolic energy were calculated for a half of a gait cycle for these subjects and trials which is a verified method for computing the energy according to \cite{93}. \\