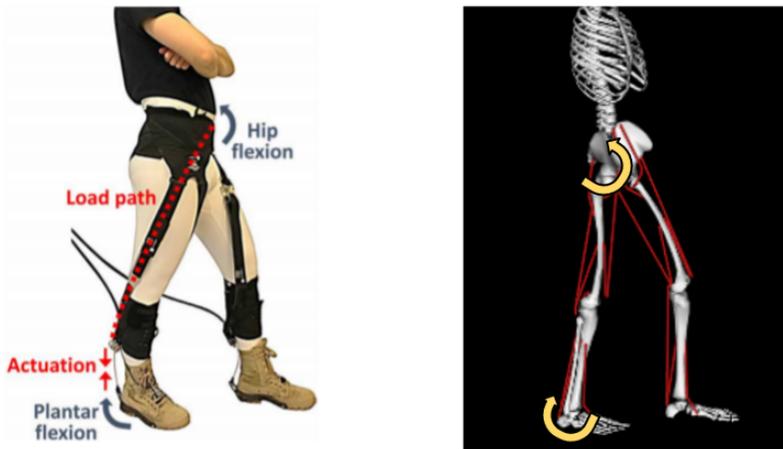


Simulation-based soft exosuit design



Team Members

- Nick Bianco
- Rachel Troutman

Project Summary

GitHub Repository

<https://github.com/nickbianco/soft-exosuit-design>

Motivation

Recent developments in the design of soft exosuits have been shown to effectively assist human locomotion. A variety of joint configurations (ankle + hip, Quinlivan et al 2017; hip, Ding et al 2016; etc) and assistive strategies have reduced the metabolic cost of walking by 5 to 20 percent. However, the human-device interaction across the design space has yet to be fully characterized, given time and cost limitations in the laboratory setting. Computational modeling and simulation approaches mitigate many of these limitations, and thus there exists an opportunity to improve the efficacy of device configurations and assistive strategies prior to experimental implementation. Using a pre-existing musculoskeletal model and example data used for human gait, our goal is to simulate a current exosuit design ([Harvard Biodesign Lab](#)) and optimize the device configuration and assistive strategy across the hip and ankle.

Research Questions

- Can an OpenSim model reproduce experimental observed metabolic cost reductions in Quinlivan et al. 2017?
- Is the linear in metabolic cost reduction valid beyond peak assistive forces in Quinlivan et al. 2017?
- What is the optimal joint torque distribution for a given exosuit configuration?
 - Now, reintroduce geometry. Can we find similar trends in metabolic cost reduction for lower device powers through optimization of exosuit configuration?
- How does device mass affect the results?

Study Approach

- Approach #1: implement ideal force actuators at ankle and hip
 - Problem #1 (simulate Quinlivan study)
 - Minimize: activations squared (possibly metabolic cost)
 - Constraints:
 - Track model kinematics, muscle parameter trajectories, joint torques
 - Bounds on excitations
 - Prescribe: biologically inspired device control at 4 assistance levels
 - Solve for: muscle activations, metabolic cost
 - Problem #2 (optimize exosuit moment arms)
 - Problem #3 (optimize exosuit moment arms + device control)
- [Optional, if time allows] Approach #2 : model device as Path Actuator
 - Problem #4 (redo Problem #1 and compare)
 - Problem #5 (solve for passive properties of device)
 - Problem #6 (solve for passive properties of device + introduce device mass)
- Implementation
 - Use DeGroot muscle redundancy solver MATLAB framework

- Compute metabolic cost using Umberger model
- Plot trajectories of muscle activations, fiber lengths/velocities, tendon lengths

Tasks

- Set up GitHub repository (organization of data files, source code, etc) [Nick]
- Download and configure GPOPS-II, DeGroot muscle redundancy solver [Rachel]
- Choose and modify OpenSim model [Nick, Rachel]
- Review relevant optimal control theory and numerical methods (i.e. direct collocation) [Nick, Rachel]
- Problem #1 - Reproduce Quinlivan 2017 study with Gait2354 default data
 - Digitize assistive curves from paper [Rachel]
 - Create generic exosuit assistive torque subroutine [Nick]
 - Run preliminary simulation [Rachel, Nick]
- Problem #2 - Reproduce Quinlivan 2017 with shifted hip assistive torque to align with biological peak [Rachel]
- Problem #3 - Examine tradeoff in hip flexion and ankle plantarflexion moment arms [Nick]
- Problem #4 - Examine tradeoff in hip extension and hip abduction moment arms [Nick]
- Problem #5 - Examine tradeoff in third case (need to define) [Rachel (probably)]
- [Optional] Compare estimations using metabolic cost solutions [Nick]
- [Optional] Compare estimations using device mass cost solutions [Rachel]
- Organize results and produce figures
 - Find a way to condense current results
- Draft final presentation
- Draft demo video
- Draft final report

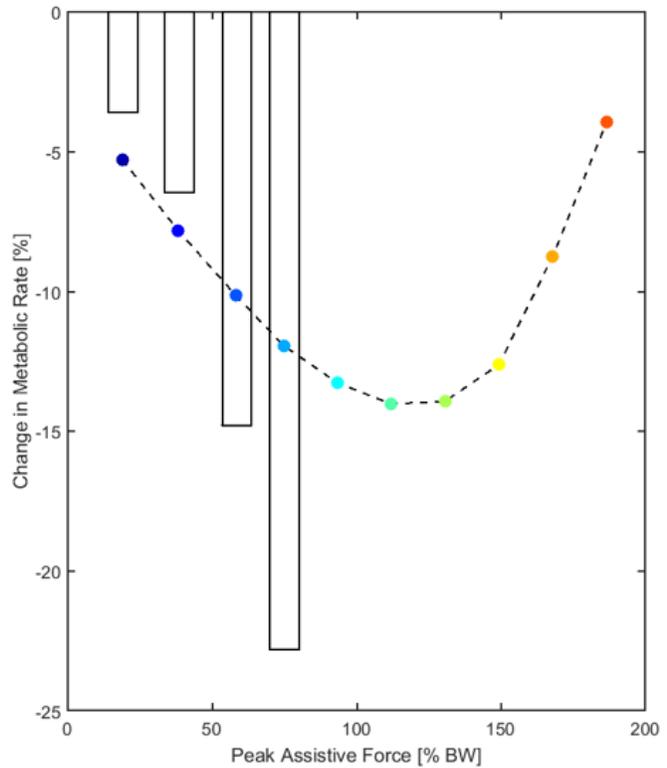
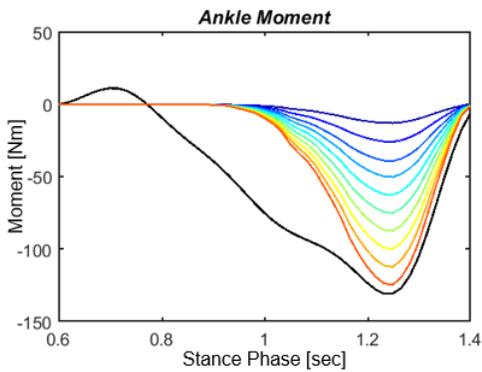
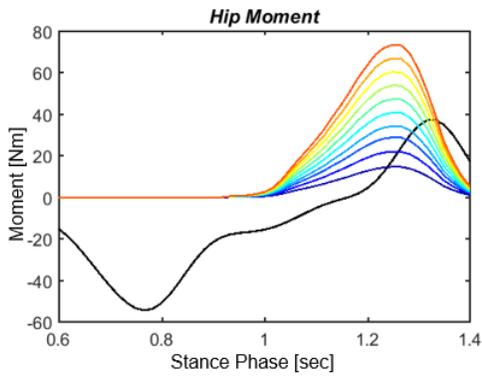
Methods

Our general optimization approach was solving a muscle redundancy problem using the Gait2354 OpenSim default model and associated data. The problem formulation is based on the direct collocation optimal control framework for solving the muscle redundancy published by Friedl De Groot.

A cost function was minimized subject to a set of constraints including tracking the joint moments calculated from inverse dynamics. The joint moments are made up of contributions from the muscles, reserve actuators, and assistive moments from the exosuit. Depending on the problem being solved, the assistive torques of the device were either prescribed or solved for as part of the optimization.

Quinlivan 2017

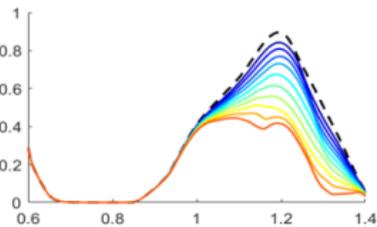
In this first problem, the goal was to simulate the experiments done with the soft exosuit in a paper by Quinlivan et al. We wanted to see if we could recreate their metabolic cost savings and predict what would happen above the maximum condition they tested. The cost function for this problem was a sum of excitations and activations of the muscles used to create the desired walking motion. For the first four conditions (blue curves), the exosuit moments were pulled directly from the paper. The next six conditions are scaled versions of these curves. The applied moments are shown below as a function of time of the stance phase. The associated change in metabolic rate for each condition is shown on the right. The simulation results are shown as colored dots and the experimental results from the paper are shown as bars.



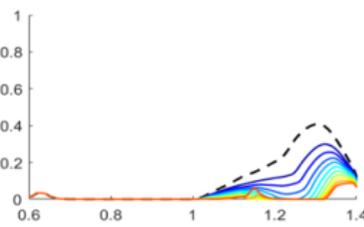
The same general trend of decreasing metabolic cost as peak assistive force increases is seen in both the simulation and experimental results. The simulation results however, underestimate the metabolic savings at the higher force levels. All metabolic costs were calculated using the Umberger metabolics model. The simulations suggest that there is some opportunity to further decrease metabolic cost with with device by increasing the assistive force above the maximum experimental trial, but the linear trend seen in the paper doesn't continue indefinitely. A bowl shape curve was found for metabolic savings over the 10 force levels tested.

The initial decrease in metabolic cost can be seen because of muscles such as the medial gastrocnemius and iliacus which are assisted by the device. Their activations go down as the assistive force from the exosuit goes up. Other muscles however, such as the tibialis anterior, are fighting against the device so their activation goes up with assistance level increasing metabolic cost.

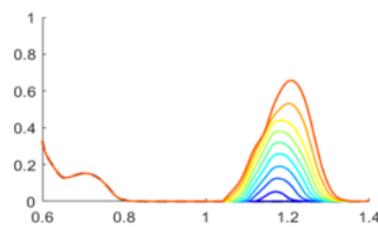
Medial Gastrocnemius



Iliacus



Tibialis Anterior



Exosuit Joint Tradeoff

The next question we investigated is if we are given a single actuator, how should we allocate assistive moments across the exosuit to cause the greatest reduction in metabolic cost? We introduced a tradeoff parameter, alpha, to allow us to weight the assistance between multiple joints.

Hip and Ankle

The first case we investigated is similar to the Quinlivan study in that we only assist at the ankle and hip. In this case, positive values of alpha means the hip flexion moment would be assisted more than the ankle plantar-flexion moment and vice-versa for negative values of alpha. In this problem we are optimizing for alpha and the exosuit activation over a range of maximum actuator forces.

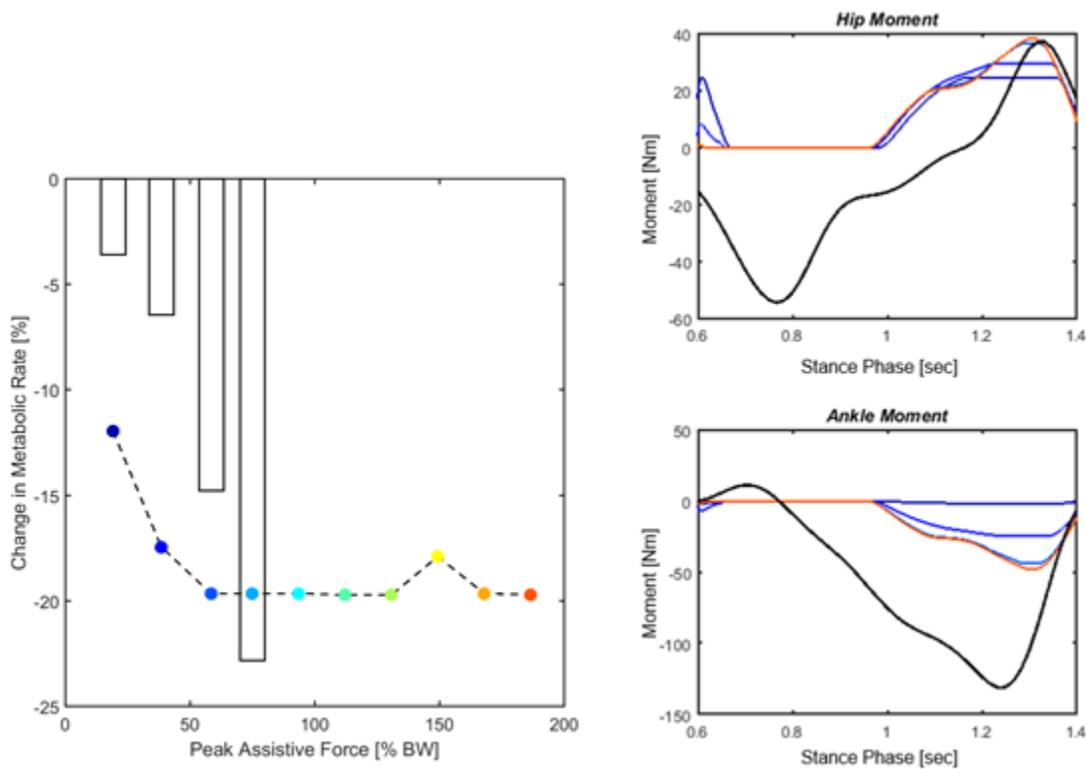
$$J = \int_{t_0}^{t_f} (e^2 + a^2) dt$$

$$T_{exo} = \begin{cases} T^{hipFlex} = a_{exo} \hat{T}_{max} (1 + \alpha) \\ T^{anklePF} = a_{exo} \hat{T}_{max} (1 - \alpha) \\ 0, \text{ other DOFs} \end{cases}$$

$$-1 \leq \alpha \leq 1$$

We found that instead of getting a bowl shaped curve for change in metabolic rate, the values plateau at larger peak assistive forces. This can be seen in the exosuit moment plots because all of the curves collapse on top of each other at the higher forces.

Interestingly, we find that the optimized assistive hip moment exceeds the inverse dynamics moment. This means there were more benefits to assisting the ankle with this profile than there were costs associated with exceeding the hip moment.



Hip, Knee and Ankle

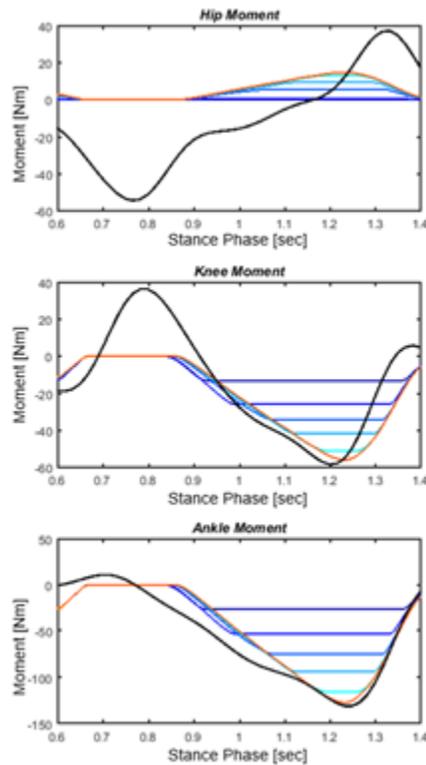
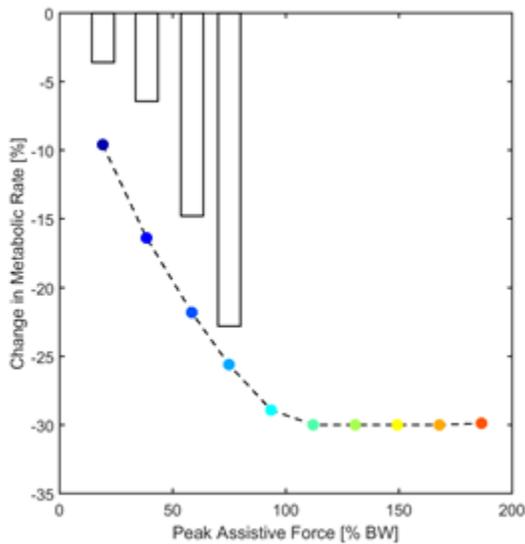
The next case we explored included adding an assistive moment at the knee. In the current device, the load path passes through the knee center. If we shifted the attachment point of the device at the ankle to increase the moment arm and assist more in ankle plantar-flexion, the load path would pass behind the knee and we would assist in knee flexion. If we shifted the moment arm of the device at the hip to assist in more hip flexion the load path would pass in front of the knee center and we would assist in knee extension. This tradeoff is again captured with alpha.

$$J = \int_{t_0}^{t_f} (e^2 + a^2) dt$$

$$T_{exo} = \begin{cases} T^{hipFlex} = a_{exo} \hat{T}_{max} (1 + \alpha) \\ T^{knee} = a_{exo} \hat{T}_{max} \alpha \\ T^{anklePF} = a_{exo} \hat{T}_{max} (1 - \alpha) \\ 0, \text{ other DOFs} \end{cases}$$

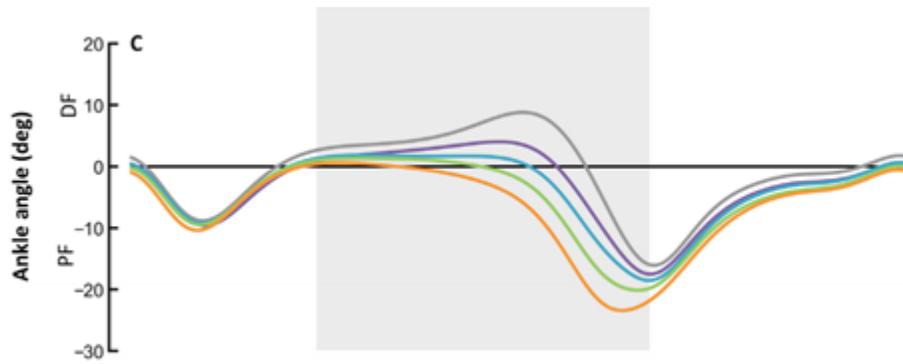
$$-1 \leq \alpha \leq 1$$

We again see the metabolic reductions plateau at higher force values. The changes are overall larger because we are now assisting an extra joint. Looking at the assistive moments across the joints we now find that the optimizer is favoring the ankle over the hip. The knee and ankle moment are close to each other in timing so great metabolic savings could be achieved even though the profile no longer aligns well with the hip. This shows the importance of the timing between the assistive moments and joint moments from inverse dynamics.



Limitations

One limitation is that the kinematics data was fixed for this study. The Quinlivan paper showed that ankle angle changed significantly with assistance level and this is not something we captured. The change in ankle angle observed in the study is shown below. Grey is the powered-off condition and purple, blue, green, and orange are the low, medium, high, and maximum conditions respectively.

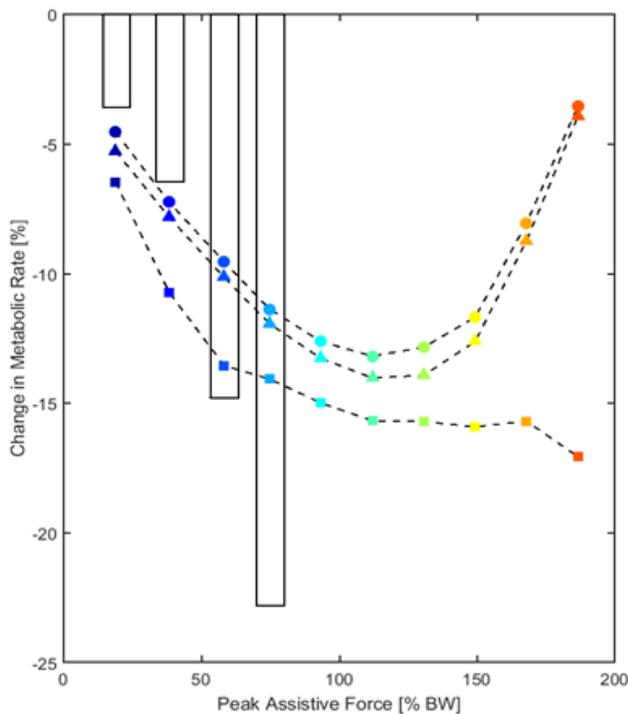


The timing and magnitude of the joint moments also had some discrepancies between the paper and data used for the simulations. For example, the timing of the peak ankle and hip moments aligned more closely and the magnitude of the hip moment was much higher in the data from the paper. Another limitation of this work is that we only looked at one subject so we could not look at the sensitivity of the device performance to subject variability.

In terms of calculating the metabolic rate of walking we had a few additional limitations. One, is that we only used the muscles in the legs to calculate the metabolic rate instead of a measure of the whole body. We only looked at the stance phase of the gait cycle and didn't take the mass of the exosuit into account.

Future Work

For future work, we want to explore other joint combinations, such as the tradeoff between assisting in hip extension versus hip abduction, and run the optimization on multiple subjects. We also want to look at the impact of different cost function formulations on our results by adding in terms for metabolics and mass penalties. Below is an example of the same exosuit torques, from Quinlivan et al., applied with different cost functions. The circles are from the sum of excitations and activations, the squares use the Minetti and Alexander model for metabolics, and the triangles combines these two methods together in the cost function of the optimization.



The solution using the Minetti and Alexander model for metabolic cost gave a greater reduction in the metabolic rate at the tested conditions but it gave a sparse solution. Many of the leg muscles were not used at all during the gait cycle. Future work could be done to tune the combined cost function, the triangles, to provide physiological solutions that reduce metabolic rates beyond the results presented here.

References

Browning, Raymond C., et al. "The effects of adding mass to the legs on the energetics and biomechanics of walking." *Medicine and science in sports and exercise* 39.3 (2007): 515.

De Groot, Friedl, et al. "Evaluation of direct collocation optimal control problem formulations for solving the muscle redundancy problem." *Annals of biomedical engineering* 44.10 (2016): 2922-2936.

Quinlivan, B. T., et al. "Assistance magnitude versus metabolic cost reductions for a tethered multiarticular soft exosuit." *Science Robotics* 2.2 (2017): eaah4416.

Minetti, A. E., and R. McN Alexander. "A theory of metabolic costs for bipedal gaits." *Journal of Theoretical Biology* 186.4 (1997): 467-476.

Mooney, Luke M., Elliott J. Rouse, and Hugh M. Herr. "Autonomous exoskeleton reduces metabolic cost of human walking during load carriage." *Journal of neuroengineering and rehabilitation* 11.1 (2014): 80.

Umberger, Brian R., Karin GM Gerritsen, and Philip E. Martin. "A model of human muscle energy expenditure." *Computer methods in biomechanics and biomedical engineering* 6.2 (2003): 99-111.

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