

The Effect of Gastrocnemius Avoidance Gait on Knee Loads

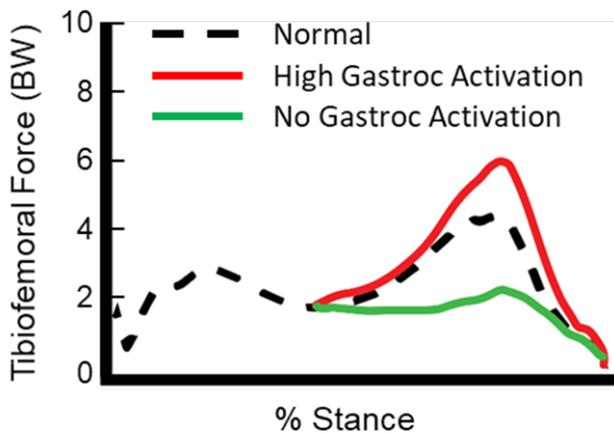
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Introduction

Background

Osteoarthritis affects roughly 10% of people over the age of 60 and is the second leading cause of years lost to disability in America. While joint replacement is effective at improving quality of life for individuals with end-stage osteoarthritis, conservative interventions to delay the need for this invasive procedure are desirable. Increased contact forces in the joint are thought to accelerate the structural progression of osteoarthritis, so many conservative interventions seek to reduce these forces.



Demers et al. 2014, modified

Project Goal

Demers et al. (2014) used simulations to investigate the sensitivity of joint loading to systematic changes in the coordination strategy used to achieve experimentally measured walking kinematics. The results of this study suggest that recruiting the soleus more heavily than the gastrocnemius could reduce knee loads (adapted image below). Kinematics were fixed in this study, so it is not clear what compensatory kinematic and coordination changes would occur when adopting a gastrocnemius avoidance gait. The purpose of our study was to experimentally reproduce Demers' simulation results by testing if walking with gastrocnemius avoidance gait does reduce late stance knee loads.

Simulation Approach

Reducing knee joint moments in all three planes is the objective of many kinematic gait retraining

paradigms; however, since coordination retraining aims to reduce the muscular contribution to joint loading, inverse dynamics alone is insufficient to assess its effectiveness in reducing knee loads. Additionally, solutions to the muscle redundancy problem that are not informed by EMG are insufficient to characterize changes in knee loads that result from coordination changes. EMG has been incorporated into simulations in a variety of ways including EMG-informed CMC (Hamner et al., 2013, Demerican et al., 2015), EMG-driven forward dynamics (Arnold et al., 2013), calibrated, EMG-driven forward simulations (Lloyd and Besier, 2003), and EMG-driven forward simulation-static optimization hybrids (Sartori et al., 2014). The benefit of EMG-driven forward simulations is that the joint moments produced by the muscles from the simulation can be validated against inverse dynamics joint moments. In fact, the calibration step in calibrated EMG-driven simulations involves parameter optimization to match muscle joint moments to inverse dynamics joint moments for a variety of activities. We chose to use an EMG-driven forward simulation similar to Arnold et al. (2013) with the understanding that our forward dynamics and inverse dynamics joint moments would not match exactly, but that we would be able to tune them by hand or use a parameter optimization package such as CEINMS (Pizzolato et al., 2015) if necessary.

Methods

Experiment

We collected marker, force, and electromyography (EMG) data on a single subject walking on an instrumented treadmill. We collected marker data sufficient to estimate the kinematics of both legs and the torso. We collected EMG from medial and lateral gastrocnemius, soleus, tibialis anterior, vastus lateralis, vastus medialis, rectus femoris, biceps femoris, semitendinosus, gracilis, gluteus medius, and gluteus maximus. The subject performed a static calibration trial that was later used to scale the model as well as maximum contraction trials for EMG normalization. During the walking trial, the subject was given real-time feedback of the EMG signals from his medial gastrocnemius and soleus muscles and instructed to change his gait to modulate the activation ratio of the two muscles.

Force and marker data were low-pass filtered at 15 Hz. EMG data was demeaned, bandpass-filtered (30Hz-500Hz), rectified, and lowpass-filtered (4Hz). EMG during the walking trial was normalized by maximum, normalized EMG signal for each muscle during the maximum contraction trial. From the walking trial, two gait cycles were chosen for analysis: a cycle with low gastrocnemius activity ("gastroc avoidance") and high gastrocnemius activity ("more gastroc") relative to soleus activity.

Simulation

Model

We used a modified version of the musculoskeletal model presented in Rajagopal et al. (2016). We lumped the mass of the arms into the torso mass, deleted muscles on the swing leg, and deleted some muscles that crossed only the hip and ankle that we did not measure.

Scaling and Inverse Kinematics

We scaled our model according to the recommendations made in the [April 27, 2017 Opensim webinar](#). The feet in the scaled model look large in the vertical direction as they represent both the feet and the shoes. Once we had scaled our model, we ran inverse kinematics for both the gastroc avoidance and more gastroc gait cycles.



Inverse Dynamics

Experimental ground reaction forces and inverse kinematics were input into the Inverse Dynamics tool to determine joint moments to compare to the forward dynamics joint moments for validation.

Forward Dynamics

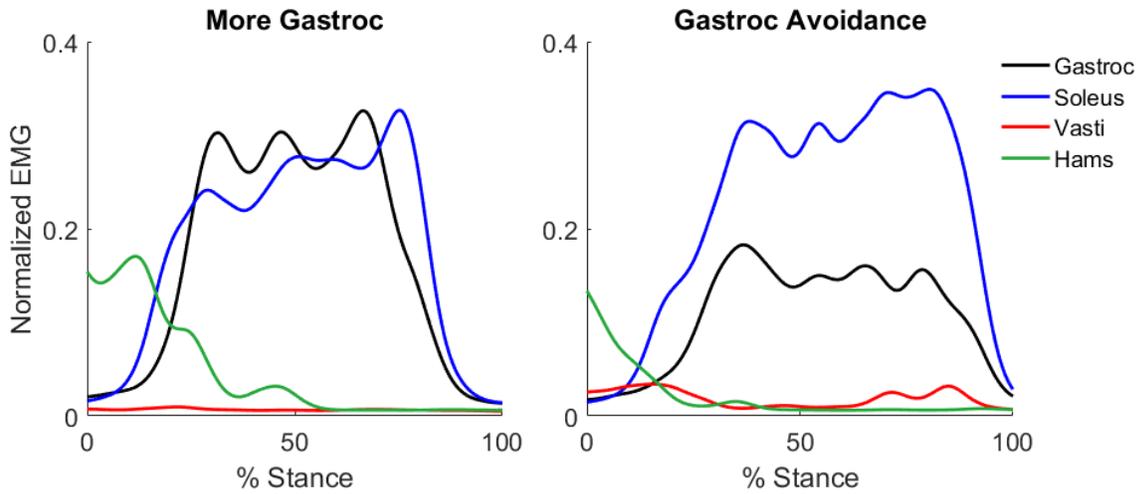
Prescribed motion from the inverse kinematics results were incorporated into the scaled model for both gait cycles. Normalized EMG was added as an excitation control. For muscles that remained in the model from which we did not collect EMG, we made assumptions from surrounding muscles with similar function. For example, vastus lateralis EMG was used as excitation for the vastus lateralis and vastus intermedius and the tibialis posterior was assigned the same excitation as the soleus.

Joint Reaction and Muscle Analyses

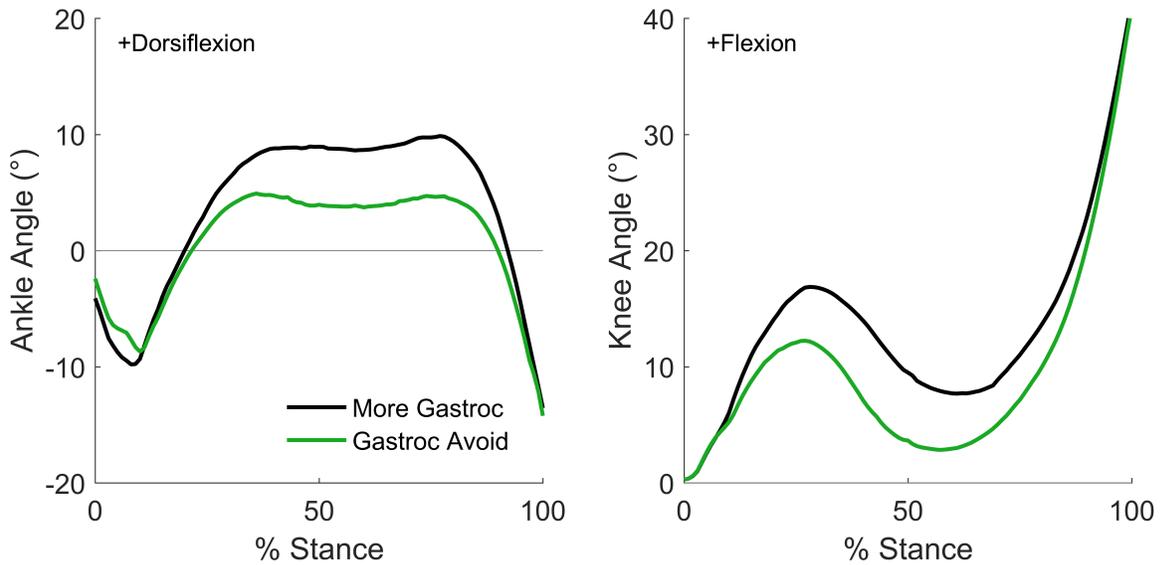
Following the forward dynamics simulation, both the joint reaction and muscle analysis tools were used to compute knee loads and joint moments, respectively. Knee loads are expressed in the axial direction of the tibial reference frame. The muscle analysis tool outputs the moment of each muscle for each coordinate, so the knee moment was calculated as the sum of the moment contribution of every muscle to knee flexion/extension.

Results

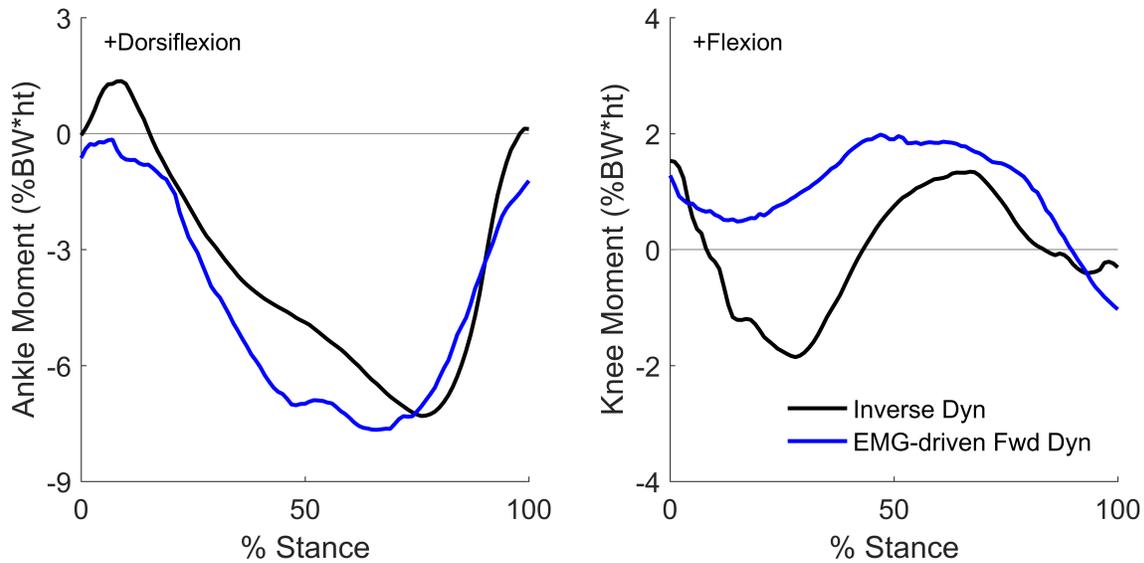
The normalized EMG for both gait cycles demonstrates relatively equal activation between the gastrocnemius (average between medial and lateral) and soleus activations in the "more gastroc" case. In the gastroc avoidance case, gastrocnemius activation is much lower than soleus. Vasti activation (average between medial and lateral) is quite small for both cases, but especially the more gastroc case.



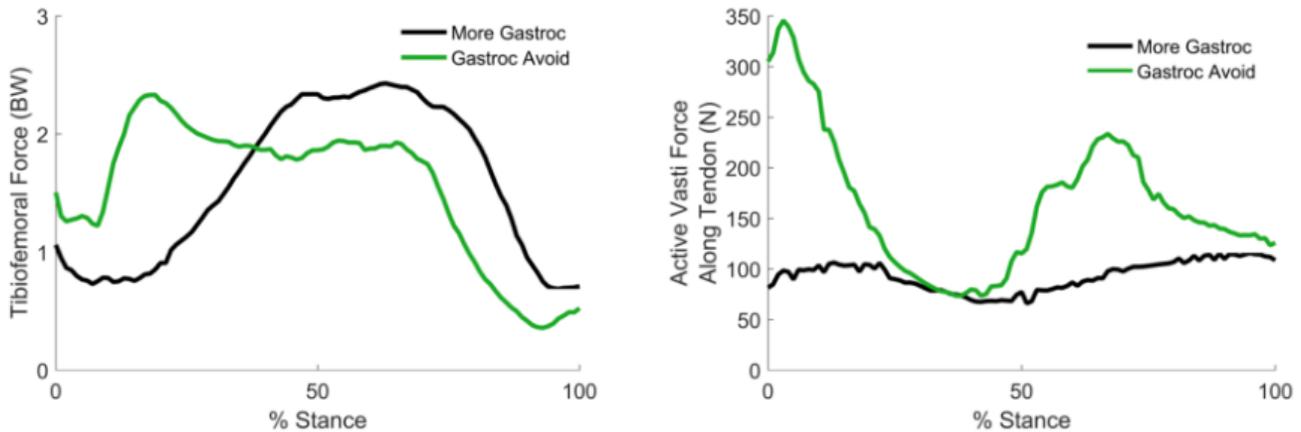
The joint angle results from inverse kinematics match the general shape that we would expect. During the gastroc avoidance gait cycle, the subject walked with a more plantarflexed ankle and a more extended knee than during the more gastroc case.



The plot below compares ankle and knee joint moments for the more gastroc case calculated from inverse dynamics and the EMG-driven forward dynamics simulation. The ankle moments match relatively well in both shape and magnitude. The forward dynamics knee moment is generally shifted in the flexion direction. In fact, there is no extension moment at all during early stance. The knee moment matches better during late stance which is where we will be analyzing knee loads. These results are without any tuning of parameters converting EMG to control inputs or model parameters.



The knee loads (tibiofemoral force) during late stance (where we are more confident with our model validity) are lower during the gastroc avoidance gait cycle than the more gastroc gait cycle. The magnitude and shape of the gastroc avoidance profile matches the simulation results from Demers et al. (2014). The early stance knee loads from the more gastroc case are less than one body weight, which along with the lack of early stance extension moment (plot above) indicate an under representation of the quadriceps muscles. This is confirmed in the plot of active fiber force along the tendon of the quadriceps muscles below.



Future Work

Further work is necessary to improve the agreement of the inverse dynamics joint moments and forward dynamics joint moments. Hand tuning parameters associated with converting EMG to control inputs (delays, scale factors) is a place to start, along with hand tuning optimal fiber force, optimal fiber length, and tendon slack length to make the joint moments match more closely. For example, shortening the optimal fiber length of the vasti could cause them to generate more passive force during early stance, solving one of our identified problems with the current results. For a more comprehensive tuning of parameters, we could use CEINMS to tune parameters using other motion trials before running the forward simulations of the motions we care about.

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