Fall 2023 Research Summary
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Rationale:

Anterior cruciate ligament (ACL) rupture is a frequently occurring injury in the United States, affecting 68.6 persons per 100,000.[1] The use of electromyography (EMG) and motion capture allows for objective analysis of patients' movement. Muscle excitation from EMGs gives insight into muscle function, including the intensity of exercise.[2, 3] Motion capture enables clinicians to determine joint angles using inverse kinematics. After ACL reconstruction surgery, in the injured leg, we expect to see increases in normalized EMG muscle excitation in the quadriceps and decreases in hip flexion and knee flexion during gait.[2-5] These changes can be measured over time to track rehabilitation progression and may provide the clinician with valuable information, leading to improved patient outcomes.

In this project, we used Delsys EMG, Optitrack motion capture cameras, and Bertec force plates to compare lower limb joint kinematics and quadriceps excitation of the injured and contralateral leg at 3 months after ACL reconstruction surgery (n = 3). Hip, knee, and ankle angles and excitation of the vastus medialis (VM), rectus femoris (RF), and vastus lateralis (VL) muscles were calculated over the gait cycle. Muscle activation was also compared to muscle volumes obtained from magnetic resonance imaging (MRI). This project was conducted as a part of the collaborative study between The Musculoskeletal Biomechanics Lab at Carnegie Mellon and The University of Pittsburgh Medical Center.

Approach:

Before using subject data, we collected a practice motion capture dataset involving one subject performing overground walking and pre-processed it using Motive. The markers for motion capture were placed based on a modified version of the Rizzoli markerset (49 markers). Pre-processing involved digital labeling of the markers on the subject, removal of a marker if it fell off during the trial, and resolution of occlusion events (portions of the trial where the motion capture cameras cannot see the marker). After this, marker trajectories were filtered using a 2nd-order Butterworth filter (fc = 6Hz) and transformed for use in OpenSim. A generic model created by Rajagopal et. al was scaled in Opensim using the Scaling Tool.[6] The first step in the scaling process is to compare the distances between anatomical landmarks in the experimental markerset and the generic markerset during a static trial to determine scale factors between the experimental model and the generic model. These scale factors are then used to scale each of the bone segments of the model in the x-, y-, and z-dimensions. Next, the scaled model is placed in the optimal position by minimizing the distances between the experimental markers and the generic markers on the scaled bone segments, and then markers are moved on the model to most closely match the experimental markerset during the static trial. The model is considered appropriately scaled when the root mean square error and marker error (distance between experimental markers and model markers) are below 2 cm and 2-4 cm, respectively. Additionally, scaling can be evaluated visually on each segment by overlaying the experimental markerset data with the scaled model during a motion to verify that the markers overlap throughout the motion.

Once the model was properly scaled, the inverse kinematics toolbox in OpenSim was used to optimally position the model to align the markers on the model to each frame of the experimental markerset walking data. Once this process is complete, the joint angles between each segment of the model during the trial are calculated. This data was then imported into Matlab where it is segmented into gait cycles using force plate data to identify heel strike events. A threshold value of 20% of the participant's body weight was used to identify heel strike. 20 of the segmented gait cycles are averaged and the standard deviation is calculated for each participant. Across all three participants, the average joint angles and standard deviation are calculated and plotted for the ankle, knee, and hip of each leg.

For the EMGs, a similar process is used. First, the EMGs were synced with the motion capture data using the acceleration of the inertial measurement sensor in the RF EMG and the acceleration of the thigh cluster of markers during three jumps at the beginning of the trial. Next, they were filtered using several Butterworth filters to remove noise, rectified, and normalized by the highest peak of the maximum voluntary contraction and the dynamic activities during the study. Next, the EMGs of VM, RF, and VL for the injured and contralateral leg are segmented and plotted over the gait cycle (mean +/- std).

Lower body MRIs were collected and processed by Springbok Analytics to obtain muscle volumes. The ratios of contralateral to injured leg muscle volume were calculated as well as the ratios of contralateral to injured peak muscle excitation over the gait cycle for each VM, RF, and VL.

Results:

Overall, the results were consistent with expectations. At the start of the gait cycle, the injured leg's hip flexion angle is 4.7 degrees lower than the contralateral leg (Figure 1). The decrease in hip flexion at the start of the gait cycle means that the leg will not be extended out as far in front of the body, which leads to decreased stride length. A reduced stride length is previously reported in individuals after ACL reconstruction.[4,5] The knee flexion angle at mid-stance (10-30% of the gait cycle) is also 8.0 degrees lower in the injured leg (Figure 1). Lower knee flexion angles in the injured leg are likely due to knee stiffness or the body's preference to avoid loading the injured knee and knee extensors in greater amounts of knee flexion. At heel strike, the injured leg's ankle is slightly plantarflexed (angle < 0 — toe pointed downwards) instead of dorsiflexed (angle > 0 — toe pointed upwards) which is consistent with decreased hip and knee flexion angles and reduced stride length (Figure 2).

Concerning the EMG results, the injured leg has higher amounts of relative VM, RF, and VM activation (Figures 3 and 4). These results are consistent with previous research.[2] Because the injured leg is weaker, walking is more intense and requires greater muscle activation to produce a similar amount of force as the contralateral leg.

Except for one outlier (P3 VM), all the contralateral to injured leg muscle excitation ratios are less than one and the contralateral to injured leg muscle volume ratios are greater than one (Figure 5). This is expected because muscle volume is correlated with muscle strength and therefore a higher relative activation would typically be required for a smaller muscle to produce a similar amount of force.[7] The limitations of the relative muscle activation because of the normalization process will be further discussed in the technical setbacks and limitations section.

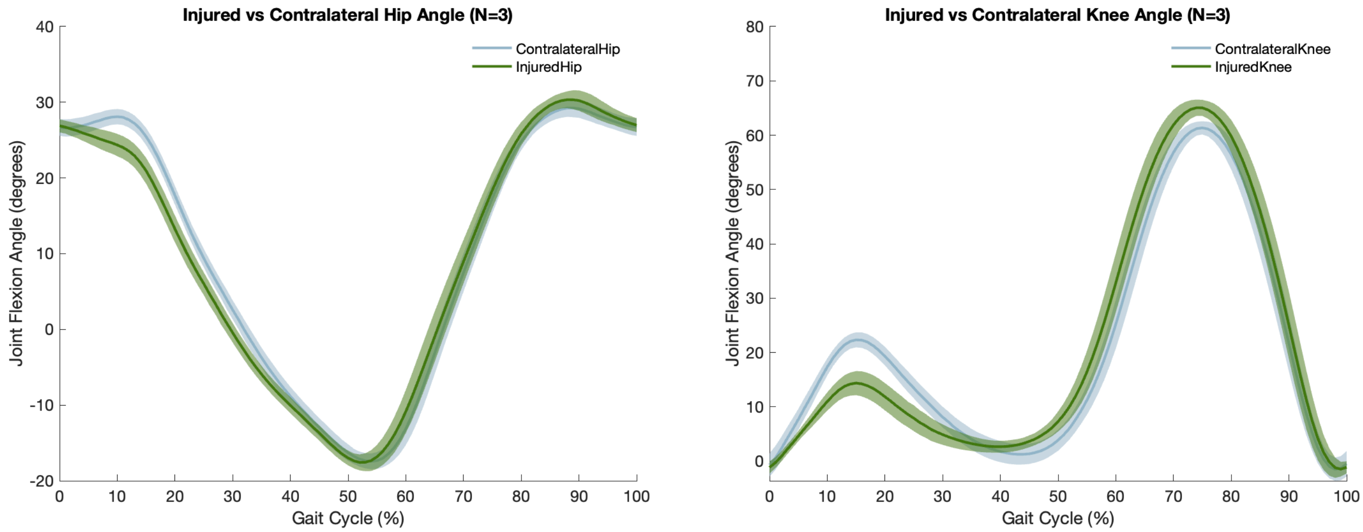

Figure 1: Injured vs Contralateral Hip and Knee Angle During Walking (n=3)

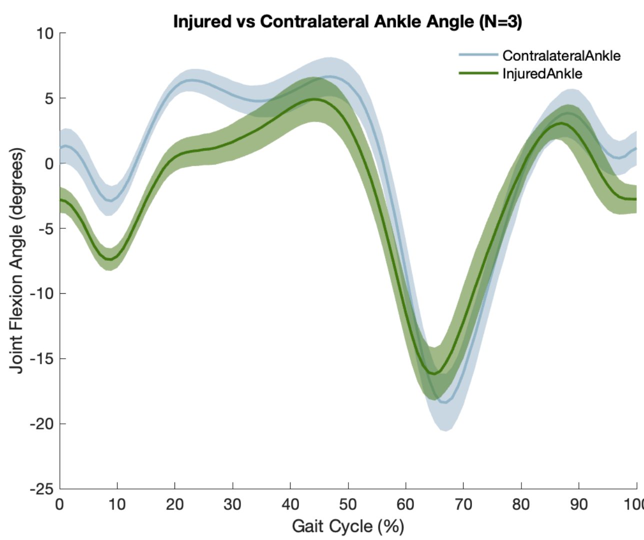

Figure 2: Injured vs Contralateral Ankle Angle During Walking (n=3)

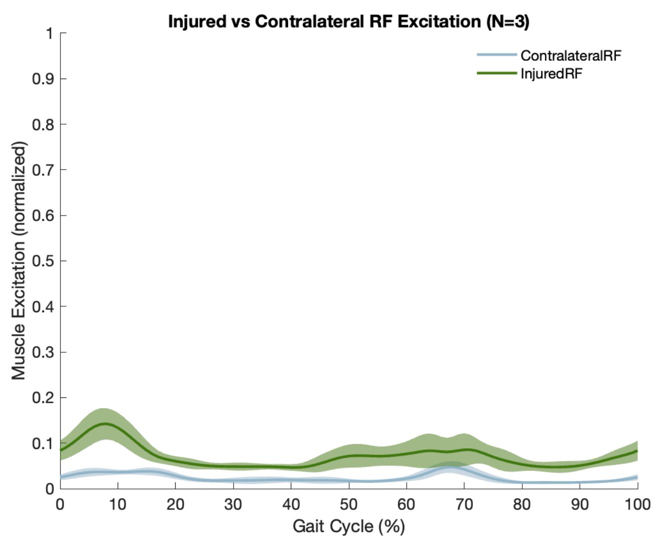

Figure 3: Injured vs Contralateral Rectus Femoris Excitation During Walking (n=3)

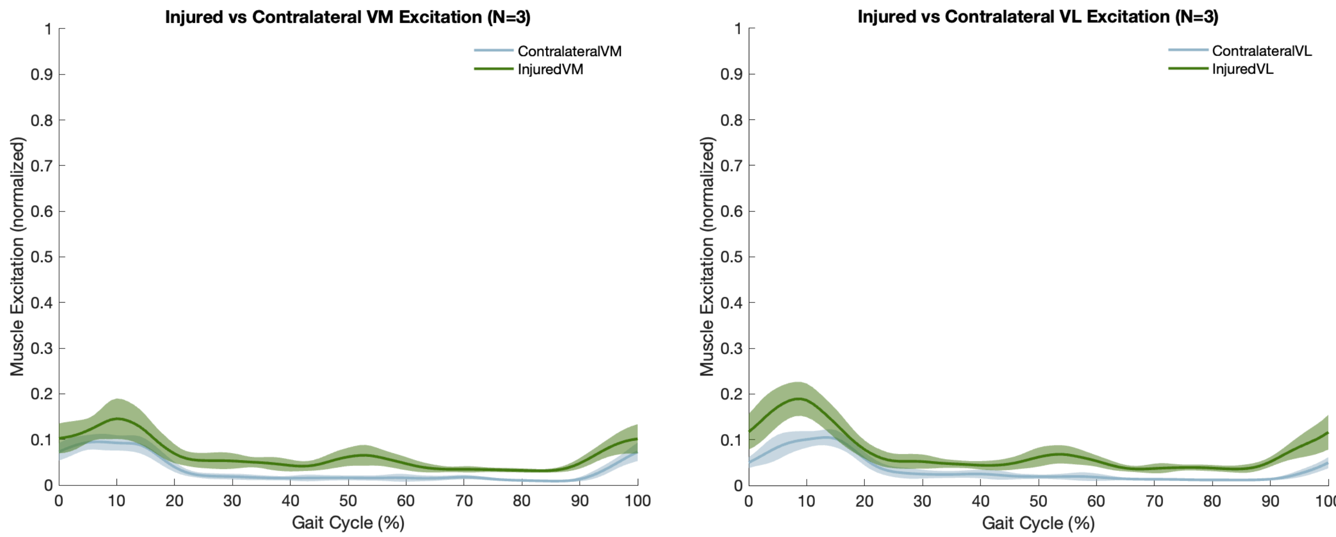

Figure 4: Injured vs Contralateral Vastus Medialis and Vastus Lateralis Excitation During Walking (n=3)

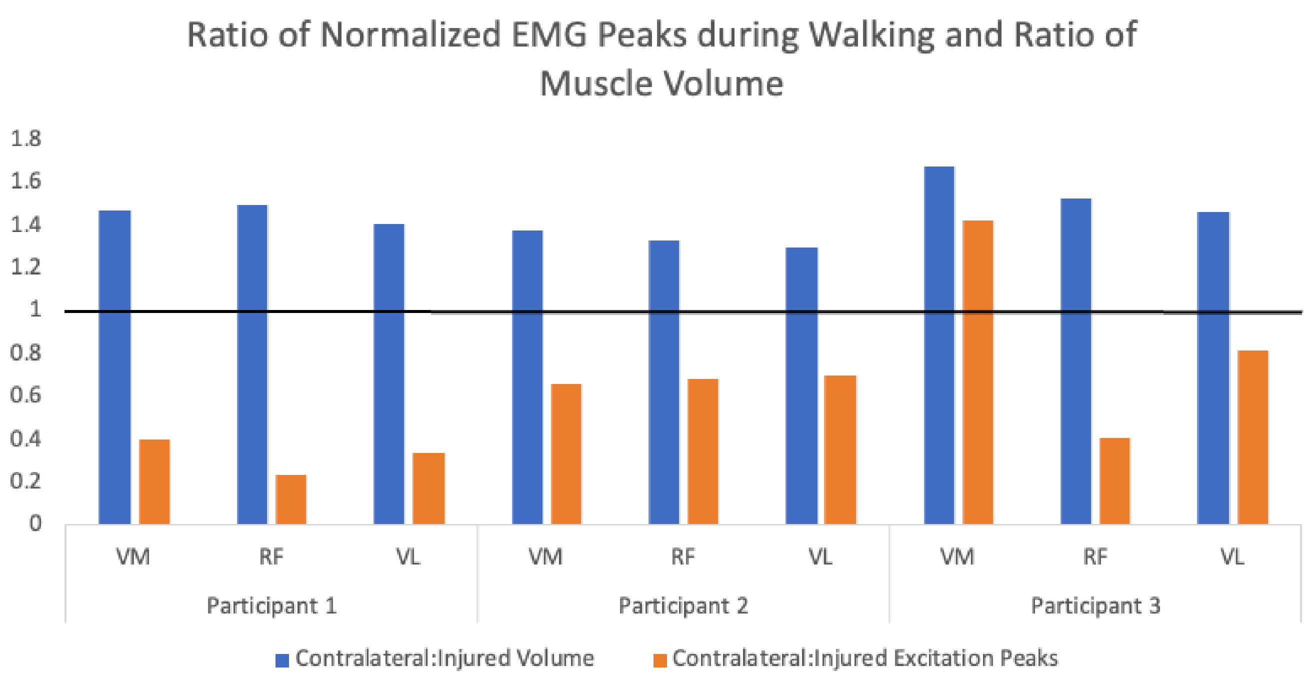

Figure 5: Ratio of Normalized EMG Peaks during Walking and Ratio of Muscle Volume

Technical Setbacks and Limitations:

The only technical setback encountered was scaling of the foot using OpenSim. Because of differences in participant footwear and shoe size, placing markers consistently is difficult. Additionally, all of the markers on the foot are close to each other because the foot is much smaller than other bone segments. Additionally, the generic model treats the foot as two rigid bones which also leads to difficulty scaling. However, through experimentation with many different marker pairings, the foot can still be accurately scaled.

The normalization process for EMGs is limited because the majority of subjects are unable to produce a high maximum voluntary isometric contraction (MVC), especially in their injured leg.[8] Sometimes during dynamic movements, the subjects will produce higher excitations than their MVC trials. As a result, we used the highest value for each muscle across all trials to normalize the EMG data.

Another limitation of this project is the limited sample size. Statistical analyses were not performed because of the small sample size. More participants from this study must be analyzed before making any conclusions.

Conclusion

The results of this project showed numerous differences in joint kinematics of the injured leg at 3 months after ACL reconstruction surgery, including less hip flexion and ankle dorsiflexion at the start of the gait cycle and less knee flexion during stance. Additionally, more EMG activity in the injured leg's quadriceps musculature was also successfully identified. These differences were then compared to one another using the ratio of the contralateral to the injured leg for EMG peaks and muscle volume, and these ratios were consistent with expectations. To establish greater statistical significance, this analysis will be carried out on a larger sample size of participants from this study at multiple time points in the future. The preliminary results from this project are promising and establish a greater understanding of how patients' muscle and joint function are altered after ACL reconstruction.

References:

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