Gait Adaptations In ACL-Injured Patients Before and After Operative Reconstruction

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Introduction
ACL-deficient and ACL-reconstructed patients have been shown to walk with altered joint moment patterns involving reduced extensor activity at the knee (DeVita et al., 1997, DeVita et al., 1998). However, the mechanisms involved in these adaptations are not completely understood. Each one of the prior studies has used a Newton-Euler inverse dynamics model to calculate net joint moments of the lower extremity and, while this certainly provides valuable information, a Lagrangian double pendulum model can be used to further break down these moments into interactions between the segments (Whittlesey and Hamill, 1996). Thus, any changes in net moment patterns can be attributed to interactive moments such as those due to thigh and hip accelerations and ground reaction forces. Therefore, the purposes of this study were: (1) to attribute specific interactions of the lower extremity to altered joint moment patterns found in ACL-injured patients and (2) to determine the time course involved in these adaptations. It was hypothesized that there would be a reduction in knee extensor activity in early stance in the injured limbs and that the major contributions to these adaptations would be the linear acceleration of the hip and the tangential acceleration of the thigh.

Methods
Six patients with isolated ACL tears (23.8 ± 5.7 yrs., 1.7 ± .05 m, and 67.8 ± 7.3 kg) were tested just before surgery, as well as 4 weeks and 6 months after surgery. Sex-matched controls (27 ± 2.6 yrs., 1.74 ± .05 m, and 66.9 ± 10.8 kg) were also tested at the same time intervals. Using synchronized ground reaction forces and 2D kinematics, joint moments were determined for both patient limbs (injured and uninjured) and the control limb corresponding to the injured limb of their matched patient. Moments were normalized to body weight * leg length. The Lagrangian equation for the net knee moment is as follows:

\[ M_{\text{knee}} = I_{lf} \alpha_l + m_{lf} d_{lf} \left[ L_f \alpha_r \cos(\theta_l - \theta_r) + L_r \omega_r^2 \sin(\theta_l - \theta_r) \right. \\
+ a_{x_r} \cos \theta_l + (a_{y_r} + g) \sin \theta_l \left] - GRF_y(Knee_y) - GRF_y(CP_y - Knee_y) \right. \]

The maximum knee angle as well as the values of each of the interactive moments during stance corresponding temporally to the three knee moment peaks (see Figure 1) were analyzed using a two factor (Visit x Limb) ANOVA.

Figure 1: Net knee moment showing the three peaks used for analysis.
Results and Discussion
Similar to the moments reported by DeVita et al. (DeVita et al., 1997; DeVita et al., 1998), the majority of differences found in the injured knees occurred at peak 1. The greatest of these were observed four weeks after surgery when the knee moment at peak 1 in the injured limb was reduced by approximately 63% compared to controls \( p < 0.01 \), see Figure 2. Although it improved significantly from four weeks, this peak remained 37% less than controls after six months \( p < 0.01 \). Another interesting finding at this visit was the exaggerated flexor peak in the uninjured limb at peak 2 \( p = 0.01 \). It is possible that this peak is part of a mechanism to compensate for the adaptations seen in the injured limb. In any case, it must be noted that the uninjured limb of ACL-injured patients should not be used as a control for the injured limb.

Among the interactive moments though, there were much greater differences. Figure 3 shows the values of each of the moment parameters in the injured limb at peak 1. The moments due to centripetal thigh and linear hip accelerations were essentially negligible. Although the tangential thigh acceleration had a greater relative input to the overall net knee moment, it only differed from controls in the uninjured limb at 4 weeks after surgery. This moment also tended to extend the knee more which conflicts with the trend in the patient limbs of reducing net extensor activity. Overall, these moments were relatively ineffective at controlling the lower extremity compared to the adaptations seen in the moments due to ground reaction forces (GRF).
With the greater magnitude of the ground reaction forces, greater changes in the net moment can be achieved while requiring smaller modifications in the forces. Thus, the GRF moments contribute most to the adaptations seen in the net knee moments. Prior to surgery, the moment due to vertical GRF at peak 1 tended to flex the injured knee more than controls, although this was not significant ($p = 0.06$). By four weeks after surgery though, this peak was significantly more flexor in both injured and uninjured limbs and after six months, the differences remained.

These differences observed in the vertical GRF knee moments can be broken down further by looking at the forces and moment arms involved. The moment arm is the anteroposterior (A/P) distance between the center of pressure and the knee joint center. Prior to surgery, when no differences were observed in the vertical GRF moment, there was a corresponding lack of differences in the force and moment arm. However at both post-surgery visits, a reduction in vertical force at peak 1 is more than offset by an increase in moment arm. This resulted in the increased flexor component in the injured limbs. At four weeks, the increase in moment arm seemed to be at least partly due to a less flexed knee ($p = 0.01$). By six months though, the knee angles did not differ, suggesting that the increases in moment arm may have been achieved by an anterior shift of the center of pressure. Whichever may be the case, the fact remains that significant differences still existed in vGRF moment, magnitude, and moment arm after six months.

As shown in Figure 3, the moment due to the vertical GRF at peak 1 did not change over time, but the net muscle moment did. The parameter responsible for this further reduction in knee extensor activity is the A/P ground reaction force. In the injured limb, a reduction in the braking phase of the anteroposterior GRF further reduced the weight acceptance knee moment peak. This peak was 30% less than it was before surgery ($p = 0.01$) and 40% less than the control peak ($p < 0.01$). Since the A/P GRF is responsible for the majority of shear forces at the knee and that actions resembling the braking phase in walking were the mechanisms for many of the patients’ injuries, this is not necessarily an unexpected adaptation. At this point, it can be surmised that the patients have adopted a cautious gait pattern.

In summary, the dominant factors affecting the moments about the injured knee are the ground reaction forces and the reduced motion of the knee joint. Through slight adjustments in knee angle, trunk position and vertical acceleration of the body center of mass, the patients were able to reduce the amount of extensor activity required by the knee and thus reduce the subsequent strain in the ACL. While this may indicate a lower risk of reinjury in the ligament, it has been suggested that this increased requirement of the knee flexors may lead to excessive joint contact forces amongst the stabilizing structures of the knee (Rudolph et al., 1998). Similar to studies by DeVita et al. (1998), the joint moments about the injured knees still differed from the healthy controls after rehabilitation. However, due to the existence of significant differences in interactive parameters after six months, these data indicate that the patients may be even further from their pre-injury state.

**References**


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