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Long Term Gait Deviations in Anterior Cruciate Ligament Reconstructed Females

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Abstract

Purpose—Little is known of the potential long term gait alterations that occur after an anterior cruciate ligament (ACL) reconstruction. In particular, variables such as impact loading which have been previously associated with joint deterioration have not been studied in walking and running after an ACL reconstruction. The purpose of this study was to define the alterations in impact forces, loading rates, and the accompanying sagittal plane kinematic and kinetic mechanics at the time of impact between the ACL reconstructed group and a healthy control group.

Methods—40 females (20 ACL reconstruction, 20 controls) participated in the study. An instrumented gait analysis was performed on all subjects. Between group and limb comparisons were made for initial vertical impact force, loading rate, sagittal plane knee and hip angles as well as moments.

Results—During walking and running the ACL cohort had significantly greater initial vertical impact force ($p=0.002$ and $p=0.001$), and loading rates ($p=0.03$ and $p=0.01$), as well as a smaller knee extensor moment and hip angle during walking ($p=0.000$ and $p=0.01$). There was a trend towards a smaller knee moment and hip angle during running ($p=0.08$ and $p=0.06$) as well as a larger hip extensor moment during walking ($p=0.06$) in the ACL group. No differences were found for hip extensor moment during running, knee angles between groups during walking or running. Lastly, no between limb differences were found for any variable.

Conclusion—Gait deviations such as elevated impact loading and loading rates do not resolve long term after the individual has resumed previous activity levels and may contribute to the greater risk of early joint degeneration in this population.

Keywords

running; knee injury; impact force; loading rates; walking

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Introduction

An estimated 250,000 Anterior cruciate ligament (ACL) tears occur annually(10). This injury also disproportionately affects females participating in sports involving jumping and pivoting who are at a 4–6 times greater risk of suffering an ACL tear (3, 13). The post surgical management of these patients is associated with a long and extensive period of rehabilitation (1). Even though good clinical results are reported, there are well documented deficits in quadriceps strength and functional performance during tasks such as jumping which persist even after rehabilitation has completed (14, 23, 24, 34). Comparatively though, little is still known of the long term alterations in gait (12). For instance, early after surgical reconstruction reductions in the knee extensor moment during early stance in walking and jogging have been reported (12, 17). Such a gait pattern has been speculated to lead to greater initial impact forces and loading rates in the post ACL reconstructed patient (17). These forces are thought to be one of the contributing factors to the deterioration of the tibiofemoral joint after an ACL reconstruction (2). In fact, elevated impact forces and loading rates have been previously associated with cartilage degeneration, subchondral bone stiffening, and have been observed in individuals with established tibiofemoral osteoarthritis (8, 16, 20). Evidence of long term increases in the initial impact forces in the post ACL reconstructed patient has to date been limited to one small study(5). While jump landing kinematics and muscular control during jump tasks has been studied extensively, very little is known about impact forces and the associated kinematics and kinetics of walking after ACL injury. This is surprising, as these are the earliest and most common activities that ACL injured patients return to after surgery. Given that an average adult takes between 5,000 and 7,000 steps per day even slight adaptations in these variables could have a significant cumulative effect on a already vulnerable knee (39).

Healthy cohorts demonstrate a more extended knee angle and a smaller knee extension moment in early stance during loading response is associated with greater vertical impact peak loading (6, 27). This is concerning, as reduced knee extensor moments are common in individuals who have undergone ACL reconstruction (12, 33). Further investigation into these findings, particularly with emphasis on the sagittal plane mechanics at the hip and knee during early stance could provide important insights into the control of initial impact loading during gait after an ACL reconstruction. Identification of the potential modifiable kinematic and kinetic alterations that occur at initial impact could provide a focus for developing new treatments to teach patients to reduce impact loading during gait after an ACL reconstruction.

The identification of potential long term gait adaptations after an ACL reconstruction requires the assessment of both walking and running. For instance, the differing muscular demands and gait mechanics between walking and running may make it easier to maintain compensations during walking but not running (22). Such a finding could suggest that long term adaptations are task dependent. Conversely, if compensations are found in both walking and running then this could suggest a global reorganization of neuromuscular control of gait that is not task dependent. Lastly, while differences in loading rates have been reported in other tasks such as drop vertical jumps, there been no comparisons of the

differences in initial impact loading and associated sagittal plane adaptations in walking and running after an ACL reconstruction (24).

Therefore, the purpose of this study was two fold: first to compare the initial impact forces and vertical loading rates, between those who have had an ACL reconstruction to uninjured individuals. A secondary purpose was to identify the alterations in the sagittal plane hip and knee angles as well as moments that occur during initial impact loading. We hypothesized that during walking and running the operative limb would have a greater impact peak force and loading rate as compared to their **non operative limb and a healthy control group**. We also hypothesized there would be a smaller hip flexion and knee flexion angles, a smaller knee extensor moment, and a larger hip extensor moment at the time of initial impact peak in the involved limb when compared their **non operative limb and a healthy control group**.

Material and Methods

Two groups of twenty females between the ages of 18–45 were recruited for this study. All participants in the ACL group must have had a unilateral ACL reconstruction, completed rehabilitation, and returned to playing sports. Only females were recruited as previous research has shown gender differences in gait within healthy cohorts and in other tasks post operatively such as jumping (9, 25, 26). Subjects were not excluded on the basis of a bone patellar bone graft, hamstring graft, allograft, meniscal repair, or partial meniscectomy at the time of ACL reconstruction. Eligible subjects were involved in sports at the time of injury and reported that they had returned to competitive sports and jogging at least two times a week on the Tegner activity scale (38). The control group had to be free of any injuries within the past six months; individuals with a previous ACL reconstruction, meniscectomy, or other knee surgery were excluded. The control group was matched to the ACL subjects by age, height, weight, and activity level. In addition, all participants had to be heel-toe runners. The study was approved by the internal review board and prior to enrollment, all participants gave their written consent for the study.

All participants completed an instrumented gait analysis. First, retroreflective markers were placed on the skin at the L4–5 junction, bilateral iliac crests, anterior superior iliac spines, greater trochanters, medial and lateral femoral condyles, tibial plateaus, malleoli, as well as the first and fifth metatarsal heads. Rigid shells with clusters of 4 markers were placed bilaterally on the thigh, and shank. Additional tracking markers were placed on the posterior aspect of the shoes. All subjects wore New Balance WR662 running sneakers (New Balance, Brighton, MA, USA). A standing calibration was then collected, followed by a hip motion trial to establish the hip joint center (35). The participant was then asked to walk at a self selected pace for a 5 minute warm-up to familiarize herself with the instrumented treadmill (Bertec Corp, Columbus, Ohio). After the warm-up, the individual walked at 1.5 m/s and then ran at 2.8 m/s during which the marker trajectories and force data were recorded. The marker trajectories were recorded at 200Hz with a 15 camera motion analysis system (Motion Analysis Corp, Santa Rosa, USA). Force data was collected at 1200 Hz and heel strike and toe off were determined when the vertical ground reaction force was greater or less than 30N.

The data were then post-processed with Visual 3D software (C-motion, Germantown, MD, USA) to filter the data, identify the functional hip joint center as well as calculate joint angles and joint moments. Marker trajectory data was filtered at 8Hz and force data at 35Hz using a 4th order Butterworth low pass filter. Custom Labview code (National instruments, Austin, Texas) was used to extract the sagittal plane joint angles, moments, loading rates and initial impact forces. The average loading rate was determined between 20–80% of the period of time between foot strike and the initial impact peak in running or at 15% of stance in walking (Figure 1). The period of time for calculation of the average loading rate, and the initial impact forces was chosen to be consistent with other previously published reports(19, 20). Forces were expressed as (%BW) and loading rates as (%BW/s). Joint angles were determined using an x-y-z Cardan angle sequence, referencing the distal segment to the proximal. Joint moments were calculated from the proximal end of the distal segment and were normalized to body mass and height. The injured limb in the ACL subject was compared to the same limb in the matched control participant. Means and standard deviations were then calculated for each group. Because of the differences in walking and running gait we stratified the analysis based upon gait condition and differences between groups were determined using an independent two samples t-test. To determine between limb differences, the value of the injured limb was subtracted from the non-injured limb, this was repeated in the same limb order for the matched control subject. Means and differences between groups were then determined as described above. Statistical analysis was performed with SPSS version 18.0 (IBM SPSS, Chicago, Illinois). Differences were considered significant if $p < 0.05$, trends were operationally defined as p -values between $p = 0.05$ to $p = 0.10$. Lastly, each participants kinematic and kinetic data series were time normalized and averaged across the 5 trials and then across each group to generate ensemble average graphs. Because of this the discrete data points taken from individual trials may not be reflected in the time normalized and averaged graphs.

Results

The groups were equally balanced for age, activity level, height and weight (Table 1). The ACL group was on average 5 ± 3 years following completion of rehabilitation with 8 individuals having a hamstring graft, 7 with a patellar tendon graft and 5 with an autograft. In addition 5 ACL subjects had a meniscus repair and 4 had a meniscectomy.

There was a significantly higher initial impact force ($p = 0.001$) and average loading rate ($p = 0.01$) in the ACL group as compared to the control group during running (Table 2). In addition, during walking the ACL group had a significantly greater initial impact force ($p = 0.002$) and average loading rate ($p = 0.03$) (Table 2).

Sagittal plane gait mechanics in early stance were different between the ACL cohort and matched control group (Table 2). The ACL cohort had significantly less hip flexion ($p = 0.01$) as compared to the control group during walking (Figure 2). However, this difference did not reach significance in running (0.06). There was also a trend towards a greater hip extensor moment in the ACL cohort in walking ($p = 0.06$) that was not seen in running ($p = 0.2$) (Figure 3). Knee flexion angle was not different between groups (Table 2), however, the ACL group demonstrated significantly lower knee extensor moments during walking

($p=0.000$) and a trend toward lower moments during running when compared to the control group ($p=0.08$). Lastly, we found no differences between limbs for any of the sagittal plane angles or moments (Table 3).

Discussion

The purpose of this paper was to determine whether females who have had an ACL reconstruction exhibit greater initial impact forces and loading rates during walking and running when compared to a healthy control group. Additionally, we sought to determine if there were significant differences in sagittal plane hip and knee joint angles and moments at the same time that initial vertical impact force was determined. We found significantly higher initial impact forces and loading rates during walking and running in ACL reconstructed athletes compared to controls. We also found differences in hip flexion angles as well as hip and knee moments between groups. Contrary to our expectations we found no between limb differences for any variable. These findings suggest there are long term gait adaptations that occur during both walking and running after an ACL.

Greater impact forces during gait have been previously associated with alterations in both bone and cartilage that may contribute to the development of OA (15, 16). In the current study we found significantly higher impact forces during both walking and running in the ACL cohort. Since walking and running are highly repetitive even small increases in impact forces could contribute to joint deterioration. For instance, modest increases in impact forces result in subchondral bone stiffening, as well as the initiation and propagation of surface fissure in cartilage, which can lead to long term joint damage (8, 16, 20, 30, 31). The larger forces observed in both walking and running suggest that the post ACL reconstructed individual experiences these greater forces during daily activities and not just during sporting activities that incorporate running. Individuals who have had an ACL reconstruction typically also have other injuries such as bone bruises and meniscus tears, which may reduce the capacity of the tibiofemoral joint to modulate such transient forces, resulting in further joint damage (18, 29). However, this premise is speculative at this point but it does highlight the need for additional longitudinal studies combining biomechanics and markers (radiographic and serum) of joint degeneration. Such studies would provide a more complete understanding of the interaction of altered mechanics on joint health after an ACL reconstruction from which new therapies could be developed.

The greater loading rates found in the ACL cohort and not the healthy control group is consistent with a previous smaller study (5). High loading rates have been associated with a number of overuse injuries such as tibial stress fractures and plantar fasciitis (19, 28). The elevated loading rates observed in the current study could be one of the contributing factors to the higher risk for developing an overuse injury after an ACL reconstruction (40). The application of rapidly applied loads to the body has also been shown to diminish the viscoelastic response of cartilage, affecting its ability to dampen forces without being damaged (30). Indeed, loading rates have been reported to be 50% higher in those with tibiofemoral OA than healthy individuals (20). Adequate quadriceps strength has been shown to play a critical role in modulating these forces (15), and several studies have reported quadriceps strength reductions after an ACL reconstruction (14, 34, 36, 42).

Although quadriceps strength and activation was not measured in the current study, the reduced knee extensor moment demonstrated by the ACL cohort may indicate either a reduced activation of the quadriceps or greater activation of the knee flexors during the initial contact of running and walking.

While we found a significantly lower knee extensor moment, the knee flexion angle was not statistically different between groups. These results suggest that over time knee kinematics may become similar to matched control subjects, but alterations in the knee extensor moment persist (figure 2, 3). While we measured the knee extensor moment at initial vertical impact force, our results are similar to others who have reported reductions in the knee extensor moment at other time points during stance (12, 33). It appears, that such a gait pattern persists even after rehabilitation has completed and the individual returns to full sports participation. This gait pattern could be a learned response that persists after the initial injury to minimize anterior translation of the tibia, or it could be a carry-over from a period of limited knee mobility and strength after surgery. As discussed previously the persistence of a smaller knee extensor moment diminishes the dampening of impact forces by the **surrounding musculature** resulting in a greater portion of the forces being translated to the cartilage and osseous structures of the lower extremity (17).

Several interesting alterations in the knee moment values and curves between walking and running were observed. For example, we noted that the ACL group shifted from a small knee extensor moment during loading response in walking to a small knee flexor moment during running (table 2, figure 1). The differences are likely do to initial impact peak occurring earlier in running than walking. For example, in running the initial impact peak occurs very early and this period of time occurs as the knee moment transitions from a flexor to extensor moment (figure 1a). By contrast the linear region of the first vertical slope of the vertical ground reaction force during the walking is longer. This period of time is typical associated with the end of loading response, and is associated with a larger knee extensor moment (Figure 1). Our findings of the overall pattern of the curves and magnitudes are similar to the other published work that has examined both walking and running after an ACL reconstruction (17). We can also observe from figure 1 that the ACL cohort appears to reverse from an early flexor to extensor moment later than the control group during running but not in walking. This could be indicative of a decreased ability to rapidly generate sufficient quadriceps force during early stance during the more demanding task of running. Future studies using electromyography are needed to better elucidate this observation. During both walking and running the ACL group uses a smaller knee moment throughout stance, which is similar to a previous report that focused on peak knee extensor moment later in stance (17).

We found that individuals who had an ACL reconstruction utilized a significantly smaller hip flexion angle with a trend towards a larger hip extensor moment during walking. The same trend persisted during running, although it did not achieve statistical significance. The smaller hip flexion angle would result in a more vertically oriented position of the limb at the time of initial impact. The reduction in hip flexion may have been due to a greater reliance on the hip extensor muscles as seen by the larger hip extensor moment. The hip extensor muscles in early stance help limit the forward flexion of the thigh which assists the

quadriceps in the control of knee flexion (41). It is possible that the reduced knee extensor moment was offset by a larger hip extensor moment to maintain stability of the knee. Such a strategy though comes at the expense of greater impact loading and loading rates during early stance.

Between limb differences in impact forces and loading rates have been reported in individuals after an ACL reconstruction as they perform other tasks such as jump landings(7, 24). In the current study we did not find any such differences. One potential reason could be due to task specificity. During a drop vertical jump landing where both legs make contact at relatively the same period of time and the individual can bias the landing to the non-injured leg. By contrast gait is a reciprocal motion involving a period of single limb support for which it maybe more difficult to decrease the rate of loading on one limb as compared to the other. Bilateral accommodations to the post operative ACL gait pattern at the hip and knee have been observed at other time points of the gait cycle (9, 12). The findings of the current study provide additional evidence that comparisons to the uninvolved limb during gait may reduce the power to observe potential adaptations and could be indicative of a increased risk of injury bilaterally.

Besides a reduction in the neuromuscular control of the knee there are other potential mechanisms that could contribute to greater loading rates and impact forces in the ACL cohort after surgery. For instance, poor knee proprioception is thought to be related to greater impact forces (32). A recent observational study of healthy females did find a relationship between proprioception and loading rates (30). However, Co et al., did not find a link between loading rates and knee proprioception in those who have had an ACL reconstruction(5). The ambiguity between studies could be due to differing methodologies of quantifying knee proprioception and patient populations. If there are alterations in the ability of the central nervous system to use sensory input from soft tissue and ligamentous structures within the knee following injury, it may rely more on the acquisition of input other peripheral tissues such as osseous structures. This in turn could result in the individual landing harder during gait to generate sufficient sensory input. While this point is speculative a recent review does highlight the need for additional studies to better assess the role of the various inputs into the somatosensory system on function after an ACL reconstruction(11).

These data add to growing evidence on persistent adaptations in the gait patterns of those who have had an ACL reconstruction. Previous authors have found that other variables such as the knee adduction moment, as well as tibial internal rotation to be greater after surgery (4, 37). Computational modeling studies have also shown that alterations in gait mechanics after an ACL reconstruction may result in greater stress on areas of the joint that do not commonly absorb such forces(2). With repetitive impact this cartilage may begin to deteriorate. Increased impact forces could be one of the factors that exacerbate stress on previously unloaded parts of the tibiofemoral joint. It is still unknown, whether these adaptations occur because of an effect of the surgical reconstruction of the ACL or are the product of unaddressed neuromuscular adaptations that occur during rehabilitation. Also, we cannot rule out that this gait pattern may have existed prior to the injury. Previous

prospective research has shown altered movement patterns exist in females prior ACL tears and other injuries such as iliotibial band syndrome (13, 21).

There are several limitations to the current study. For instance, impact forces and loading rates are not direct measurements of tibiofemoral joint contact forces. However, they are reflective of the load the entire extremity experiences and give an indication of the forces the leg is attempting to modulate. Also, this study was done only in women as previous research has shown gender differences in movement patterns after surgery (25, 26). As such these results cannot be extrapolated to men. However, the results do offer insights into potential long term compensatory strategies and future studies assessing if potential gender differences exist in impact loading during gait after surgery are warranted. Finally, while elevated impact forces and loading rates are associated with joint deterioration, we did not quantify this within the study population. These results do highlight that future studies should consider these variables in prospective trials of ACL outcomes on future OA development.

Conclusion

We found significantly higher average loading rates and initial impact peaks in both walking and running in females who have had an ACL reconstruction when compared to healthy control participants. The ACL cohort also had a smaller hip flexion angle, greater hip extension moment and a smaller knee extensor moment at the time of initial impact peak. These results suggest that there continues to be unresolved gait adaptations that lead to greater impact loading even after rehabilitation has been completed. Such high impact loading may contribute to the elevated risk of future joint injury in the ACL population.

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Figure 1.

Vertical Ground reaction force (%BW), knee joint moments (N m/kg m) for the ACL group (large dashes) and the control group (small dashes) during a)walking and b)running. A vertical line was superimposed on the time point approximating the initial impact peak during running and 15% of stance in walking. The region over which loading rates were measured occurred sooner in running than walking. This also corresponded to the period of time when individuals transitioned from a small flexor to an extensor moment. Knee extensor moment is positive. The transition was delayed in the ACL cohort during running

and explains the small knee flexor moment observed at initial impact peak. Lastly, the orange brackets indicate the region over which the average loading rate was determined.

Figure 2.

Kinematic ensemble curves for the ACL (dashed line) and control group (solid line) during walking and running for a) knee flexion, b) hip flexion and during running for c) knee flexion as well as d) hip flexion.

Figure 3.

Joint moment ensemble curves for the ACL (dashed line) and control group (solid line) during walking and running for a) knee flexion, b) hip flexion and during running for c) knee flexion as well as d) hip flexion.

Table 1

Subject Demographics mean (standard deviation) for the ACL and control group.

	ACL	Control
Age (yrs)	25 (6.2)	26 (5.1)
Height (m)	1.6 (0.06)	1.6 (0.06)
Weight (kg)	64 (8.2)	61 (5.1)
Activity (Tegner)	6.4 (1.4)	6.5 (1.1)
Time since Surgery (yrs)	5.2 (3.2)	

Variables of interest between the ACL group and the control group expressed as mean, standard deviation, and p values. Hip flexion angle and the knee extensor moment was positive, while hip extensor angle, knee flexion angle, and the hip extensor moment were all negative.

	Walk		Run	
	ACL	Control	ACL	Control
Impact force (BW)	1.0 (0.08)	0.9 (0.08)	1.6 (0.24)	1.4 (0.18)
Average Loading Rate (BW/s)	17.0 (6.0)	13.3 (4.6)	82.8 (27.2)	65.8 (15.2)
Hip Flexion Angle (Deg)	20.0 (7.4)	26.1 (7.4)	26.0 (7.3)	30.7 (7.9)
Hip Extensor Moment (N ^{*m} /Kg ^{*m})	-0.3 (0.4)	-0.1 (0.4)	-0.7 (0.7)	-0.5 (0.6)
Knee Flexion Angle (Deg)	-18.5 (5.0)	-20.1 (5.1)	-25.7 (6.2)	-27.2 (4.0)
Knee Extensor moment (N ^{*m} /Kg ^{*m})	0.04 (0.2)	0.23 (0.1)	-0.07 (0.4)	0.11 (0.1)

Table 3

The between limb differences for the ACL and control group expressed as mean, standard deviation and p values.

	Walk			Run		
	ACL	Control	p	ACL	Control	p
Impact force (BW)	0.03(0.25)	-0.02 (0.08)	0.85	0.06 (0.15)	-0.007 (0.14)	0.11
Average Loading Rate (BW/s)	-1.0 (6.7)	-0.19(2.3)	0.60	1.7 (12.81)	-0.76 (10.6)	0.51
Hip Flexion Angle (Deg)	-1.04 (10.3)	0.19 (4.7)	0.62	-0.56 (3.18)	-0.26 (3.0)	0.75
Hip Extensor Moment (N*m/Kg*m)	-0.15 (0.73)	0.15 (0.61)	0.16	-0.21(1.17)	0.11 (0.96)	0.33
Knee Flexion Angle (Deg)	1.38 (9.3)	0.09 (4.9)	0.59	1.07 (5.02)	0.20 (3.34)	0.51
Knee Extensor moment (N*m/Kg*m)	-0.05 (0.42)	0.006 (0.09)	0.59	-0.034 (0.13)	-0.02 (0.102)	0.70