ABSTRACT

SIMENZ, C. J. The effect of anterior cruciate ligament reconstruction on ground reaction forces during locomotion. MS in Exercise and Sport Science - Human Performance, August 1999, 50pp. (M. K. Miller)

The purpose of this study was to examine the ground reaction forces of ACL reconstruction patients in both their affected and unaffected leg and to examine the ground reaction forces of nonpathological subjects in each leg. 28 adults aged 18-30 were placed into either the "healthy" or "ACL" group and then walked over a Bertec force platform approximately 20 times. Tekscan sensors were placed in the shoes of each participant as well. Peak force measurements in the vertical and anterior/posterior directions and temporal data were analyzed and tested for significance by a multi-factor ANOVA. Tekscan data were analyzed for center of pressure differences between legs and between groups. No statistical significance was found at alpha level $p > 0.05$ for the peak force variables or temporal variables, suggesting no difference in ground reaction forces of ACL subjects when compared to healthy subjects or when compared between legs. However, trends did develop in both force platform and Tekscan data that merit further research.

THE EFFECT OF ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION ON GROUND REACTION FORCES DURING LOCOMOTION

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INTRODUCTION

Background

The process of locomotion is not easily defined. This pattern of movement is extremely complex and is comprised of at least 20 components according to Murray (11), all of which lead to a smooth, sinusoidal translation of the center of gravity. These components include rotation and displacement patterns of all joints of the upper and lower extremity. The interrelation of these and other factors such as body condition and pathology makes studying the various components of the gait pattern a difficult task. When an injury occurs, especially to the lower extremity, the process of locomotion is impaired. The role of knee flexion in locomotion is crucial to normal gait (14) and impairment of this component has lasting effects.

Though the effects of knee injury and gait patterns have been studied extensively using kinematic factors (2,3,5,14) such as the analysis of angular displacements at joints and joint torques (2,3,16), relatively little research has been published on the ground reaction forces (GRF) of ACL reconstructed patients. While previous research (8,9) has shown symmetry in ground reaction forces of healthy subjects during locomotion, the kinetic and kinematic patterns of ACL reconstructed subjects have not been thoroughly examined. This study attempted to determine whether patients who had anterior cruciate ligament reconstruction exhibited different ground reaction forces in their repaired leg versus their healthy leg and if they differ from a healthy population.

ACL injury can lead to long term changes in locomotion, affecting not only the injured knee, but also the rest of the body (2,3,5,6,16). Surgery alone is not an assurance of return to preinjured gait patterns and function in daily activity (6). In the past 10 years, the rehabilitation protocol used for ACL reconstruction has changed considerably. New, more aggressive approaches have been taken, allowing for a quicker return to typical activity for the ACL reconstructed patient (4,12,15). While it has been theorized that patients will return to typical walking patterns after completion of rehabilitation, there has been little conclusive research to support it.

The research of ACL reconstruction has yielded a wide array of results regarding surgical procedure, graft placement, and rehabilitation; many of which are in conflict with each other (6,7,13,17). At this time there is little concrete evidence to support one specific standardized treatment protocol over another. This comes as little surprise, as not even surgical procedure for ACL reconstruction is agreed upon at present (17). There is a need for more research in this area to build a pool of knowledge upon which surgical procedure, rehabilitation, and future research can be based. This study attempted to determine the effects of ACL reconstruction on ground reaction force patterns during level walking. Peak forces at first maximum force, minimum force, and second maximum force in the vertical direction as well as maximum braking force and maximum propelling force in the anterior/posterior direction as described by Bates et al (1) will be analyzed.

The purpose of this study was to determine whether or not ACL reconstruction causes changes in ground reaction forces during locomotion in the adult population. It was also an attempt to identify any patterns in ground reaction force and plantar pressure in ACL reconstructed legs during locomotion that may differ from the unaffected leg or from the legs of a healthy population.

The primary hypothesis this study tested was: ACL reconstruction would cause changes in gait resulting in lower peak ground reaction forces on the reconstructed side than on the unaffected side. A secondary hypothesis was that ACL patients would exhibit dissimilar centers of pressure with their unaffected leg versus their ACL reconstructed leg, with the reconstructed leg exhibiting a lateral shift in center of pressure.

METHODS AND PROCEDURES

Subject Selection

A pool of 28 adult subjects volunteered to participate in this study, 14 of whom had undergone ACL reconstruction. Each subject was between 1 and 5 years postsurgery, as approximately 6 months were needed to return to normal activity (5,13). The ACL subjects had undergone reconstruction using either a patellar tendon or hamstring graft. Every subject exhibited a heel-toe walking pattern with her or his unaffected leg, determined if they exhibited a double peaked ground reaction force pattern as described by Bates et al. (1). Subjects were female and male, as specified in Tables 1 and 2. Subject characteristics were given in Table 3. Individuals were not selected to participate if they had posterior cruciate ligament repair. The healthy group consisted of 14 individuals with no known knee injuries or conditions, and no surgical repair or sprain of the lower extremity for at least 6 months. Each subject read and signed an informed consent form prior to participation in this study (see Appendix A). This research protocol was examined and approved by the University of Wisconsin-La Crosse Institutional Review Board for the Protection of Human Subjects prior to use.

Table 1. ACL reconstructed subject information

Subject	Age	Height	Weight	ACL/Healthy	Surgery (year)	Graft Type	Gender
A	24	67	215	Right ACL	1997	Patellar	M
B	24	73	205	Left ACL	1996	Hamstring	M
C	23	72	205	Right ACL	1995	Patellar	M
G	22	67	160	Left ACL	1995	Patellar	M
K	23	72	185	Right ACL	1998	Patellar	M
M	29	65	150	Left ACL	1997	Patellar	F
N	25	65	123	Right ACL	1996	Hamstring	F
O	22	71	195	Left ACL	1997	Patellar	M
P	22	67	130	Left ACL	1995	Patellar	F
R	21	61	120	Left ACL	1995	Patellar	F
S	21	66	130	Right ACL	1997	Patellar	F
V	22	69	135	Left ACL	1997	Patellar	F
W	20	66	142	Left ACL	1996	Patellar	F
X	21	64	114	Left ACL	1997	Hamstring	F

Table 2. Healthy subject information

Subject	Age	Height	Weight	ACL/Healthy	Surgery (year)	Graft Type	Gender
D	27	67	140	Healthy	----	----------	M
E	23	71	163	Healthy	----	----------	M
F	22	64	152	Healthy	----	----------	F
H	22	68	150	Healthy	----	----------	F
I	24	68	147	Healthy	----	----------	M
J	23	69	153	Healthy	----	----------	F
L	24	67	133	Healthy	----	----------	F
Q	21	71	190	Healthy	----	----------	M
T	24	66	130	Healthy	----	----------	F
U	23	66	140	Healthy	----	----------	F
Y	23	66.5	138	Healthy	----	----------	F
Z	22	69	165	Healthy	----	----------	M
AA	20	67.5	140	Healthy	----	----------	F
BB	23	77	230	Healthy	----	----------	M

Table 3. Subject characteristics

ACL SUBJECTS	N	Mean	Std. Deviation
Age	14	22.7	2.3
Height (in)	14	67.5	3.5
Weight (lb.)	14	157.8	35.9

HEALTHY SUBJECTS	N	Mean	Std. Deviation
Age	14	22.9	1.6
Height (in)	14	68.4	3.1
Weight (lb.)	14	155.1	26.6

Experimental Protocol

Each subject was asked to perform 20 walking trials over a 25-foot walkway and over a Bertec © force platform system interfaced with a computer with the Ariel Performance Analysis System (APAS) © software package. The force platform sampled at 500 Hz and was set into the ground flush with the walking surface. Subjects wore one of two types of running shoe provided to them. Three walking trials of each leg were recorded and analyzed. The subjects performed 20 trials so a level of comfort was developed prior to the recorded trials, with the last three trials of each leg recorded.

Walking pace was set at 112 steps/min ± 5 steps by use of an auditory metronome. Each subject also wore Tekscan © shoe inserts to measure the plantar pressures during overground walking. These inserts initially contained 960 sensors before timing. Each sensor was trimmed to fit into one of the provided running shoes, decreasing the number of sensors. The sensors were used to measure the pressure distribution under the foot during locomotion. Sensor boxes were fixed to the ankles of each subject and tethered to a computer by a 20-foot cord. The Tekscan sensors were then calibrated to each subject's body weight prior to data collection. The computer sampled at a rate of 300 Hz which allowed for approximately seven steps (3 steps from one foot, 3 from the other foot) to be collected for each of three recorded walking trials.

Outcome Measures/Statistics

Data were collected and the ground reaction forces were compared between the control and ACL repair populations. Data were averaged from three trials for each subject and normalized by subject body weight by dividing each force magnitude for a subject by his or her body weight and multiplying by 100. The results were then presented as a percentage of subject body weight. The vertical (Fz) and anterior/posterior (Fy) force platform data were analyzed. Peak force magnitudes were recorded for first maximum force (Fz1), minimum force (Fz2), and second maximum force (Fz3) as well as maximum braking force (Fy1) and maximum propulsion force (Fy2) for the last three walking trials with each leg. Time in stance was calculated for each subject for his or her affected and unaffected legs. ACL reconstructed subjects data were grouped by unaffected leg and reconstructed leg. Healthy subjects were matched to each ACL subject so that the healthy unaffected and affected groups had the same number of right and left legs as the ACL affected and unaffected groups. Data were then analyzed using a multi-factor ANOVA test which compared each factor between affected and unaffected legs as well as between groups to check for statistical significance at an alpha level of $p < 0.05$. A qualitative approach was used to analyze the Tekscan data whereby the center of pressure patterns for each subject was examined from the three trial data collected and then compared for differences between legs and versus the other group. Six steps (three right, three left) were examined for each subject from the trial data. Three sets of two consecutive steps were taken from the recorded data. These were studied primarily to detect any medial/lateral differences in center of pressure exhibited by the ACL subjects (see Figures 1 and 2). A center of pressure pattern exhibited in two of three trials was considered normal for that subject and noted. It was expected that ACL reconstructed subjects would exhibit a lateral shift in center or pressure in the reconstructed leg.

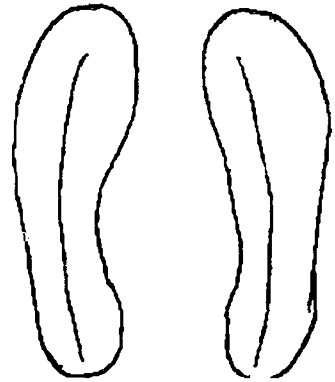

Figure 1. Example of typical center of pressure patterns during locomotion.

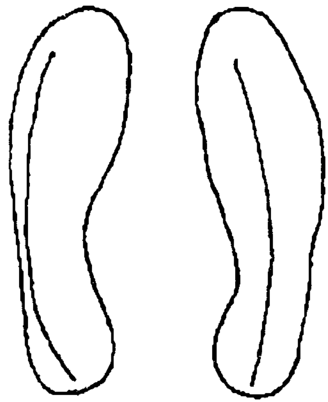

Figure 2. Example of a lateral shift in center of pressure in left foot during locomotion.

RESULTS

Force Platform Data

The first vertical peak force magnitude, Fz1, yielded no statistically significant differences between legs or between groups when examined by a multi-factor ANOVA within alpha level $p < 0.05$. However, Fz1 examined between groups yielded a p-value of 0.113 (see Table 4), which approached significance. Figure 3 shows that ACL subjects exhibited a slightly lower relative peak force at Fz1 in both legs (117.6% and 117.1%) when compared to the healthy subjects (121.6% and 123.2%).

Table 4. Between subjects effects - ANOVA statistics (GRFs)

Source	Variable	Sum of Squares	df	F	p-value
ACL	Fz1	0.0360	1	2.593	0.113
vs.	Fz2	0.0133	1	1.698	0.198
Healthy	Fz3	0.0151	1	2.888	0.095
Subjects	Fy1	0.0015	1	0.567	0.455
	Fy2	0.00003	1	0.017	0.896
Affected	Fz1	0.0006	1	0.042	0.839
vs.	Fz2	0.0101	1	1.284	0.262
Unaffected	Fz3	0.0026	1	0.504	0.481
Leg	Fy1	0.0012	1	0.451	0.505
	Fy2	0.0003	1	0.185	0.669
Group	Fz1	0.0016	1	0.115	0.736
vs.	Fz2	0.00002	1	0.003	0.958
Leg	Fz3	0.0042	1	0.814	0.371
	Fy1	0.00002	1	0.006	0.936
	Fy2	0.0003	1	0.142	0.708

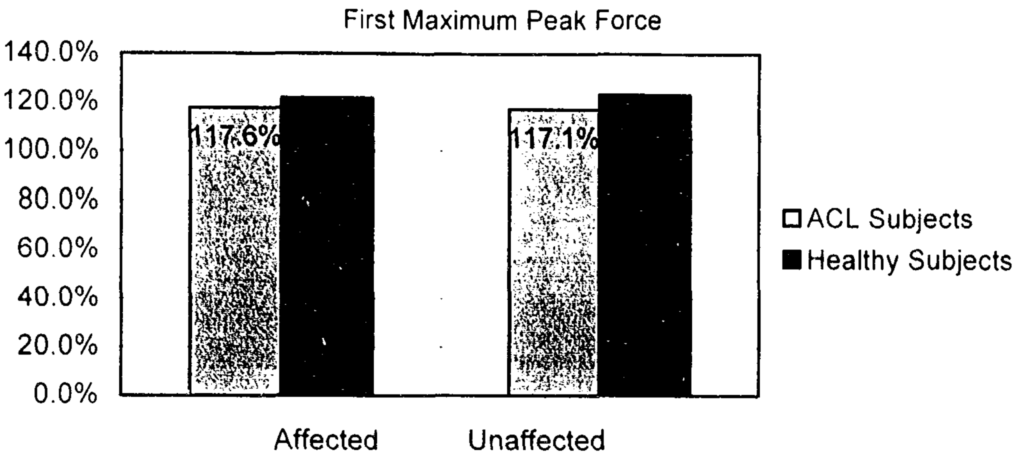

Figure 3. First maximum peak force for ACL subjects and healthy subjects.

There were no statistically significant differences found between or within groups for the Fz2 variable. Again, the measure of force between groups, p = 0.198, was nearer to significance than any other the minimum force value (see Table 4). Figure 4 showed that ACL subjects exhibited relatively larger peak forces at the Fz2 in both legs (70.2% and 67.6%) than the healthy subjects (66.4% and 65.2%).

No statistically significant differences were found between or within groups for the Fz3 when examined by the multi-factor ANOVA. This variable yielded p = 0.095 for the Fz3 between groups measure (see Table 4). This measure was the closest to any statistical significance in the study. ACL subjects again exhibited greater peak forces for Fz3 in both legs (118.7% and 118.3%) than did the healthy subjects (114.9% and 115.6%) (see Figure 5).

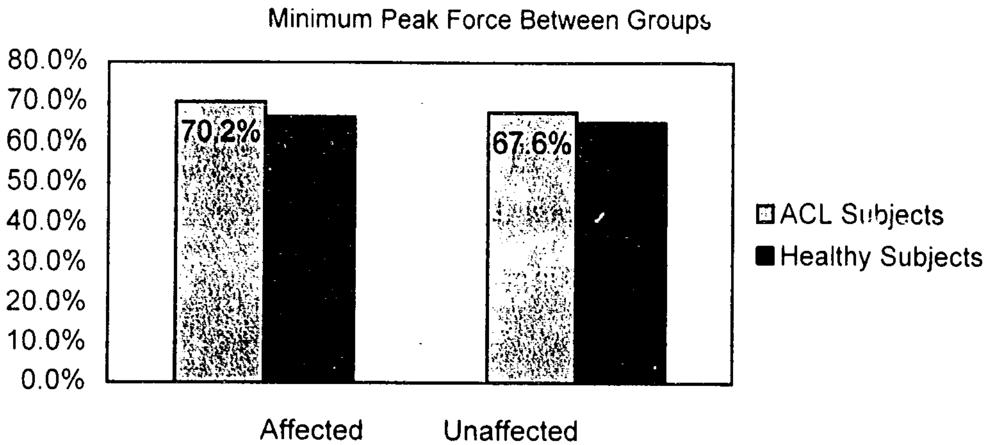

Figure 4. Minimum peak force for ACL subjects and healthy subjects.

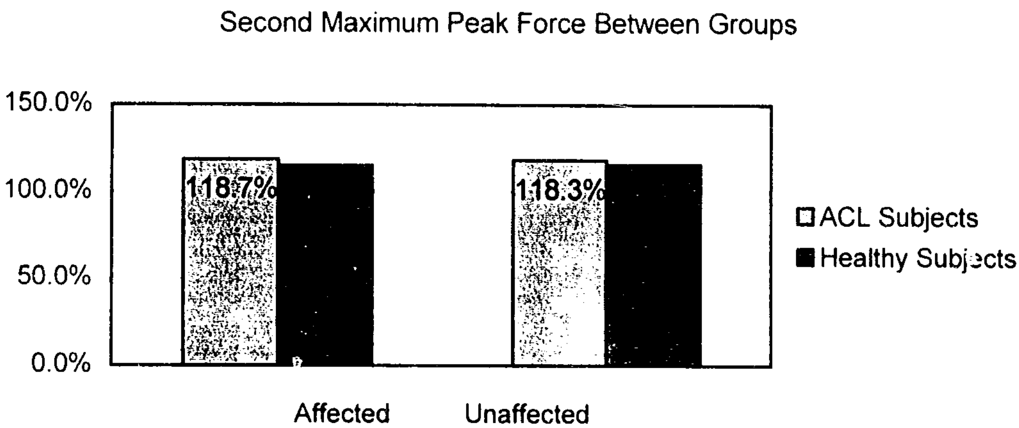

Figure 5. Second maximum peak force for ACL subjects and healthy subjects.

Peak braking forces (Fy1) were measured as the subjects walked over the force plate and were examined. No differences approaching any statistical significance were found either between or within groups for this variable. The measure with the lowest p-value was 0.455, the measure between groups.

Peak propulsive forces (Fy2) were examined in the same fashion as the other variables, but yielded nothing approaching a significant level either between ($p = 0.669$) or within groups ($p = 0.896$). Relative peak values were very similar between groups for this factor.

Temporal data were examined and tested by ANOVA. No statistical significance was found when time in stance and time to first maximum force were compared between legs or between groups (see Table 5) . The lowest p-values found were that of time in stance compared between legs ($p = 0.758$) and time to Fz1 calculated between groups ($p = 0.497$). Descriptive force and temporal statistics were calculated for healthy and ACL reconstructed subjects (see Tables 6 and 7).

Table 5. Between subjects effects - ANOVA statistics (temporal data)

Source	Variable	Sum of Squares	df	F	p-value
ACL	Time in Stance	0.00009	1	0.081	0.777
vs.					
Healthy	Time to Fz1	0.00031	1	0.468	0.497
Subjects					
Affected	Time in Stance	0.0001	1	0.096	0.758
vs.					
Unaffected	Time to Fz1	0.00001	1	0.021	0.885
Leg					
Group	Time in Stance	0.0003	1	0.303	0.584
vs.					
Leg	Time to Fz1	0.00008	1	0.117	0.734

Table 6. Descriptive statistics normalized force data - ACL subjects

Reconstructed Leg

	N	Mean (%)	Std. Error	Std. Deviation
Fz1	14	117.6	.033	.123
Fz2	14	70.2	.024	.090
Fz3	14	118.7	.013	.047
Fy1	14	21.6	.018	.067
Fy2	14	-28.7	.009	.034

Unaffected Leg

	N	Mean (%)	Std. Error	Std. Deviation
Fz1	14	117.1	.025	.093
Fz2	14	67.6	.021	.079
Fz3	14	118.3	.016	.060
Fy1	14	22.6	.013	.048
Fy2	14	-28.8	.010	.038

*Mean values are normalized to body weight and represent the percentage of total body weight.

Table 7. Descriptive statistics normalized force data - healthy subjects

Affected Leg- Matched to ACL Reconstructed Leg

	N	Mean (%)	Std. Error	Std. Deviation
Fz1	14	121.6	.035	.132
Fz2	14	66.4	.025	.094
Fz3	14	114.9	.025	.093
Fy1	14	23.9	.013	.047
Fy2	14	-28.1	.013	.050

Unaffected Leg- Matched to ACL Unaffected Leg

	N	Mean (%)	Std. Error	Std. Deviation
Fz1	14	123.2	.032	.120
Fz2	14	65.2	.025	.092
Fz3	14	115.6	.022	.083
Fy1	14	22.3	.011	.042
Fy2	14	-29.1	.012	.047

*Mean values are normalized to body weight and represent the percentage of total body weight.

Tekscan Data

The Tekscan data was examined qualitatively to determine differences in center of pressure patterns between feet and between groups. Three trials were examined for each subject, with 6 total steps examined. A pattern exhibited in 2 of the 3 trials was considered normal for that subject and noted. It was expected that ACL subjects would exhibit a lateral shift in their center of pressure in the reconstructed leg (see Figures 1 and 2). No differences were found between feet in the normal group, with only small variations in center of pressure between trials. Within the ACL group, 6 subjects exhibited different center of pressure patterns between feet. Five of these subjects exhibited a lateral shift in center of pressure in their ACL reconstructed leg while maintaining a more typical pattern with their healthy leg (see Figure 2). The sixth subject was found to have an extreme lateral pattern in the healthy leg and a typical pattern in the ACL reconstructed leg. Although the Bertec force platform also calculates center of pressure during locomotion, it could not capture consecutive footfalls of the right and left leg. Zhu et al described the advantage of examining both interstep and intertrial variations (18). For this reason, the Tekscan data were used to examine center of pressure patterns.

DISCUSSION

Force Platform Data

The results of this study were not expected, but some trends merit further investigation. The force platform data collected did not support the hypothesis that ACL reconstruction subjects would exhibit lower peak ground reaction forces on their reconstructed leg versus their unaffected leg and versus a healthy population. No differences in the temporal data existed between or within groups.

The measurement of Fz1 between groups was not found to be significantly different. Though not statistically significant, this measure showed a slight difference in Fz1 between groups. ACL subjects tended to land with less relative force at Fz1, suggesting that the subjects may have entered the stance phase of gait with greater knee flexion. This flexed knee approach would seem to match the findings of studies examining ACL deficient and injured subjects (2,3) rather than ACL reconstructed subjects, since injured subjects have been shown to exhibit this gait pattern while reconstructed subjects have not (3,5). This would be a direct affect of the most important determinant of normal gait, knee flexion in stance (14). According to Saunders et al, this change in flexion in stance would lead to other changes in the gait pattern, which would make gait less efficient (14).

Fz2 measured between groups was again found to have no significance. While the numbers were not statistically significant, the finding, combined with the Fz1 measurement, showed an emerging trend of different walking patterns between groups.

The increased peak force at Fz2 exhibited by the ACL subjects could be the result of co-activation of the hamstrings and quadriceps during mid-stance (1,4,15). This co-contraction generally occurs to stabilize the knee during the stance phase of gait (1,4,15). The result may have been that the reconstructed knee resisted flexion, decreasing downward acceleration and increasing the Fz2 when compared to healthy subjects. Flexion of the knee in stance has been shown to affect the power and work done by both the hip and knee, with the hip increasing and the knee decreasing output (1,4). These changes specifically demonstrate an agreement with what Saunders et al hypothesized (14).

The Fz3 measured between groups was found to be the nearest to statistical significance (see Table 3). The lack of statistical significance would seem to concur with the findings of Bulgheroni et al, who reported that vertical ground reaction force values returned to normal in tested ACL reconstruction subjects (3). No literature found commented on greater Fz3 exhibited by ACL reconstructed patients.

The temporal values were also found to be statistically insignificant. These values have not been reported in previous ACL gait literature and therefore cannot be compared with the present data. The metronome cadence could have affected this outcome measure, since subjects were asked to keep pace with the auditory signal when walking. By testing subjects at self-selected pace, this measure could be better examined.

These findings were not statistically significant, which would seem to agree with prior research (2,5,16) in which subjects exhibited patterns that appeared to return to presurgery levels after ACL reconstruction. They may also lend support to the idea that gait is symmetrical, however, the between groups measurements merit further investigation with larger subject pools to test the trends found in this study. The power values found during the statistical analysis were very low, with the highest value at p = 0.443. The limited subject pool in this study is a common problem in the ACL gait research available at present, with most studies having between 5 and 15 ACL reconstructed subjects (2,3,5,6,16).

A trend, though not statistically significant, did develop that seemed to lend partial support to the primary hypothesis. The Fz1, Fz2, and Fz3 values for examined between groups demonstrated values that approached significance. This suggests that some differences may exist between ACL subjects and healthy subjects during the stance phase of locomotion. Since prior research has shown that some combination kinematic/kinetic factors such as work at the knee and hip do change after ACL reconstruction, more research is warranted (2,3,5,6,15). These differences may be due to protective mechanisms employed by the ACL subjects after surgery. Some researchers have found neuromuscular responses such as changes in muscle activation patterns to be the cause of changes in the gait pattern of ACL injured and ACL reconstructed subjects (3,5), and this would seem a reasonable conclusion. By changing the gait pattern to lessen the stress on the affected leg postsurgery, subjects may develop habits that remain long after the leg has returned to near presurgery levels.

Tekscan Data

Of the 14 ACL reconstruction subjects who took part in the study, 6 showed different center of pressure patterns between feet. All 6 exhibited a lateral shift in center of pressure, with 5 subjects doing so on the reconstructed side. This finding partially supports the second hypothesis and paves the way for future study regarding center of pressure measurement on ACL subjects. Reasons for the relatively small group exhibiting differences may include type of rehabilitation done, effort put into rehabilitation, and gait patterns prior to injury. While these qualitative data are not enough to base a conclusion on, it would appear that a trend of changing center of pressure patterns on the reconstructed side may occur in ACL subjects. If this is a protective mechanism to prevent tibial internal rotation and forward translation, this could suggest knee instability (3,6,10,16). When or why this occurs is not certain, and merits further study. It is possible that this pattern of lateral center of pressure was adopted by the ACL subjects after injury to lessen the stress on the medial knee during locomotion as well as to limit the internal rotation of the lower leg at the knee (5). Determining the cause and effects of this pattern of lateral center of pressure is important because it may change the total gait pattern and have lasting effects on the body. Bulgheroni et al. found that both vertical ground reaction force and kinematic walking patterns returned to normal after surgery and rehabilitation (3).

CONCLUSIONS

This study was designed to investigate the effects of ACL reconstruction on ground reaction forces during locomotion. The results of this study found that no significant differences exist between ACL individuals and healthy subjects or between legs for any of the variables examined. Trends did emerge that suggest there may be some small differences between ACL and healthy subjects in vertical ground reaction force patterns. Center of pressure patterns of ACL subjects were found to exhibit a lateral shift in six cases. Five subjects exhibited lateral shift in the ACL leg, and one exhibited a lateral shift in the unaffected leg. These findings seem to support the few studies on ACL reconstruction and gait which state that gait patterns appear to return to values near those of healthy subjects.

Future Considerations

In the future, by using a larger subject pool, entire subject groups with ACL reconstruction on the same leg, same surgical procedure, and the same rehabilitation protocol, these trends can be examined more accurately. Also, the impulse in stance phase could be examined to determine whether or not ACL reconstructed individuals differed from nonpathological subjects. When preparing future studies, many factors not controlled in this study such as those mentioned above, need to be taken into account. ACL reconstruction introduces many variables into a research study, and only by careful control of these variables can truly meaningful data be produced.

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APPENDIX A

INFORMED CONSENT FORM

Informed Consent

The Effect of Anterior Cruciate Ligament Reconstruction on Ground Reaction Forces During Locomotion
Principal Investigator: Christopher J. Simenz

I______________________________, have volunteered to participate in this study to compare healthy adults and those with anterior cruciate ligament reconstruction surgery. I have been informed that this study will require me to walk over a force platform approximately 20 times at a pre-determined cadence (112 steps/minute).

Testing

I have been informed that testing will involve walking over a force plate at a pre-determined cadence (112 steps/minute) for 20 trials. I have also been informed that I will have to wear sensor inserts in my shoes when walking. I will also need to fill out a knee injury questionnaire prior to participation. The total amount of time for the testing will be 60 minutes.

Risks

As this study involves only walking at a moderate pace, risks associated with this particular research are minimal.

Supervision

Time will be allowed for me to warm up properly prior to testing. I have the right to ask the tester any questions at any time. I may also refuse to participate in any test or exercise I choose without any risk or recourse to myself.

Confidentiality

I have been informed that all personal information and data collected will remain confidential and that I will be issued a subject number. If this study is to be published, data will be pooled and I will be referred to by number only. Any questions regarding the protection of human subjects may be referred to Dr. Garth Tymeson, Chair, University of Wisconsin- La Crosse Institutional Review Board for the Protection of Human Subjects at (608) 785-8155.

I acknowledge that I have read the above statement and that it has been explained to me to my satisfaction. I have been informed what is expected of me, and realize that I may drop out of this project at any time and for any reason without any negative consequences. If I have any questions at any time regarding the study or my role in it, I may contact Christopher Simenz at (608) 784-6749 or Dr. Marilyn K. Miller at (608) 785-6527. By signing this statement, I give full consent to participate in this study.

Signed:______________________________ Date:_____________

Witness:_____________________________ Date:_____________

APPENDIX B

KNEE INJURY QUESTIONNAIRE

KNEE INJURY QUESTIONNAIRE Name____________________

1. Have you ever injured your knee? Yes____ No____

2. Have you had ACL reconstruction? Yes____ No____
   Patellar Tendon Graft_____ When______
   Hamstring Graft_________ When______
   Donor (Cadaver) Graft____ When______

3. If you answered yes to #1 above, have you injured…
   MCL ____ When____ Leg: R L Surgery: Yes____ No____
   PCL_____ When____ Leg: R L Surgery: Yes____ No____
   Meniscus____ When____ Leg: R L Surgery: Yes____ No____

4. Have you had any other injury to your lower extremity? Yes____ No____
   If yes, when?__________
   ____
   Did you have surgery? Yes____ No____
   When?_________
   ____

5. Have you sprained your knee/ankle in the last 6 months? Yes____ No____

6. Height___________ Weight_____________

APPENDIX C

REVIEW OF RELATED LITERATURE

REVIEW OF RELATED LITERATURE

Anterior Cruciate Ligament

The anterior cruciate ligament (ACL) is a structure that runs from the superior proximal tibia, through the intracondylar notch in the femur, and attaches on the distal femur (7,22,27,30). This ligament functions to prevent anterior displacement of the tibia in relation to the femur (7,22,23,27,30). It is one of a series of ligaments that stabilize the knee joint. It works in conjunction with the posterior cruciate ligament (PCL), which prevents posterior displacement of the tibia, and the medial and lateral collateral ligaments (MCL and LCL), which prevent medial and lateral displacement of the tibia and fibula relative to the femur. When working in unison, these ligaments provide a stable knee in joint extension. Flexion at the knee reduces tension in the anterior cruciate and collateral ligaments, allowing for greater range of motion, but severely compromising the stability of the joint (17). Since the ACL prevents forward displacement of the tibia, it is the key structure in stabilizing the knee during forward motion (29,39). Of the four ligaments that stabilize the knee, the ACL is the most critical to normal knee function (29,30).

Anterior Cruciate Ligament Injury

Injury to the ACL is more common with growing participation in athletics and physical activity. According to Woo et al (39), sports participation accounts for up to 90% of ACL injuries in young adults. Many researchers have also found that women are at a higher risk than men for ACL injury (2,15). There is also some data to support the that artificial playing surfaces may influence the occurrence of ACL injury (30). Some of the more common activities for ACL injury are snow skiing, football, basketball, and gymnastics (29).

A series of events that commonly occur in both sporting and everyday activities generally precede ACL injury. These events include deceleration of the body, coupled with extreme valgus (outward bending) or varus (inward bending) stress on the tibia, internal rotation of the tibia, and hyperextension that force the ACL against the edge of the intracondylar notch of the femur (17,29). Examples of actions which could produce such forces include but are not limited to: cutting activities in court or field sports, landing from dismounts in gymnastics, or even leaning and rotating the body to pick grocery bags from the trunk of a car. Once ruptured, an ACL cannot repair itself. Due to limited blood flow, the ACL has almost no healing ability (28,39), and once stretched, the ACL often never returns to original shape and length. Surgical reconstruction is the only way to effectively treat a completely torn ACL.

Anterior Cruciate Ligament Reconstruction

There are 3 commonly utilized methods used to replace a torn ACL. All 3 methods involve the use of a tissue graft. The original tissue is removed from the site and the graft is placed as close as possible to the original ligament site (17,29,39). Generally, small tunnels are drilled into the tibia and femur to reproduce the normal insertion sites of the ligament (17,29,39). The graft is then placed so that both ends are anchored in the tunnels and the middle portion is left to act as the new ACL. Over time, bone grows and bonds with the graft sites and securely fixes them in place. The tissue of the graft is then remodeled and revascularized by the body over time. It is important, however, to note that no graft, no matter how well placed, functions as effectively as the original ligament (17).

Infrapatellar Tendon Graft

Repair of the ACL by infrapatellar tendon graft involves taking a portion of the infrapatellar tendon of the patient from just below the kneecap and using it as a graft (17,29,39). This graft is harvested along with a piece of bone from each end, allowing for a bone-on-bone attachment. Screws are then placed to fix the graft to the femur and tibia, allowing for a more immediate and aggressive treatment to take place. Drawbacks to this method include a weakening of the patellar tendon of the involved knee, excessive scarring within the joint, and possible erosion of the underside surface of the patella.

Semitendinosus/Gracilis Tendon Graft

This ACL graft, collected from two of the five tendons of the medial thigh musculature, the semitendinosus and the gracilis, is an excellent substitute for the original ligament (17,29). These tendons are more elastic than the patellar tendon, making this procedure favorable for athletes wishing to return to sports requiring aggressive running and kicking (29). The main drawback to the semitendinosus/gracilis graft is that fixation to bone is a more difficult process since the relatively soft tissue of the tendons must be sewn to bone. This substantially increases the nonweight bearing time for each patient after surgery, however, following the initial nonweight bearing recovery of approximately 4 weeks, a more aggressive approach to active rehabilitation of the knee is usually taken (29). Another factor in favor of the semitendinosus/gracilis graft is the tendency of patients to have less anterior knee pain after surgery compared to the patellar tendon graft (17,29).

Cadaver Patellar Tendon Graft (Allograft)

Cadaver tendons are commonly used to replace the torn ACL in patients (39). The primary advantage of this process is that no tissue is taken from the patient to use as a graft. Since foreign tissue is being introduced to the body of the patient, it takes much longer for the body to remodel and revascularize the tissue (29,39). This graft is a common choice for patients with advanced degeneration of the knee due to the minimal disturbance of existing knee structures prior to surgery (29).

Rehabilitation

After reconstructive ACL surgery, patients are left with the difficult task of rehabilitation. Normal gait is affected by injury to the ACL (6,13,31) and returning to near-normal patterns is difficult. To compensate for instability of the knee due to an ACL tear, individuals exhibit inefficient transfer of weight (6). This is primarily due to different protective mechanisms the body uses to prevent anterior displacement and internal rotation of the tibia. It is uncertain whether or not these patterns are ever fully restored (6,13,14).

Less than 15 years ago, some physicians were casting ACL reconstruction patients after surgery. In the last decade, however, a more aggressive approach has been taken towards rehabilitation of ACL reconstruction patients (12,26,34). This progression involves an accelerated program of rehabilitation designed to return the patient to full activity after only 4 to 6 months post surgery (14). Despite the obvious benefit of shortened recovery time, several researchers have found problems with accelerated rehabilitation (13,14,21,36). Devita et al found that gait biomechanics were atypical following accelerated rehabilitation (14). A recent study by Bulgheroni et al found that the vertical component of the ground reaction forces of ACL reconstruction patients approached near normal values (6). This would suggest that in some cases gait kinetics may return to normal following the new standard of accelerated rehabilitation.

The Lower Extremity and Human Gait

The process of locomotion in humans has been defined previously as the translation of the body from one point to another by means of a bipedal gait (31). The kinetics and kinematics of gait in humans have been thoroughly studied at various speeds (1,3,8,9,10). Murray has described three components essential to locomotion: 1) the ability to support the upright body; 2) the ability to maintain balance in the upright position; and 3) the ability to execute the stepping movement (24). The process of locomotion involves flexion and extension at the ankles, knees, and hips. According to Saunders et al there are six determinants of efficient gait in humans (31): 1) pelvic rotation, 2) pelvic tilt, 3) knee flexion in stance phase, 4) foot mechanisms, 5) knee mechanisms, and 6) lateral displacement of pelvis.

These six components work in unison to translate the center of gravity smoothly in the forward direction, while moving it as little as possible in the vertical direction. The result is a smooth, sinusoidal pattern of vertical displacement (24,31). Murray reported vertical center of gravity oscillations of 4.8 to 6 centimeters depending on walking pace (24). This minimizes the energy required to perform the locomotive process (31).

Saunders et al described knee flexion in stance as the most important determinant of gait (31). Knee flexion in stance has the most importance because impairment leads to changes in both the hip and ankle mechanics (5,6,13,14,21,36). ACL deficient and ACL reconstructed patients exhibit decreased knee flexion when the knee is at an angle less than 40 degrees (5). Patients keep the affected knee flexed throughout the stance phase of gait to insure that it will not move through the 40 degrees of motion just prior to full extension (5). Avoidance of this near-full extension of the knee appears as a crouched, flexed knee gait pattern that increases the load on the hamstrings (13). This results in a shorter stride length in subjects with knee pathology (1). Other determinants of gait are affected in a number of ways. Since flexion of the knee is altered, the rotation and tilt of the pelvis are also changed to bring the leg through the swing phase. It has also been determined that the instability of the knee leads to an inefficient weight transfer during the stance phase where weight is shifted laterally to the outside of the foot, changing foot and knee mechanisms (6,14,21,36).

Knee Flexion in Stance

During the stance phase of locomotion, the weight of the body is shifted over the lower extremity while the knee joint is flexed (24,31). Saunders, et al. described this part of the stance phase as the "double lock cycle", referring to the full extension of the knee at heel strike, flexion during stance, and full extension at toe off (31). This phase of stance accounts for approximately 40 % of the gait cycle. Injury to the knee often results in the adoption of a guarding mechanism in which the last few degrees of extension of the knee are avoided (5). This may be a neuromuscular adaptation to the injury that prevents contraction of the quadriceps muscle, an activity which has been found to cause pain to those with ACL injury (5,6). Bulgheroni et al. found that this pattern was not evident in ACL reconstructed patients (6).

Pelvic Displacement

During locomotion, the pelvis performs a series of movements in three planes to accommodate the smooth translation of the center of gravity (24,31,38). Pelvic transverse rotation serves to limit the vertical oscillation of the center of gravity (31). Pelvic tilt occurs to allow the leg to clear the floor during the swing phase of stance (24,31). Lateral sway of the pelvis allows the center of gravity to displace itself over the support leg during locomotion (31).

Foot Mechanisms

Just prior to heel strike, the foot is held in dorsiflexion (24,31). At heel strike, the foot plantar flexes rapidly to make maximal contact with the ground. During the mid-stance portion of the stance phase, the center of gravity is shifted over the lower extremity, and after it passes over the foot, plantar flexion occurs. After toe off, the foot is again dorsiflexed to allow for ground clearance during the swing phase (24,31).

Normal Gait

Normal walking gait is characterized by two phases, stance and swing (24,31). The stance phase includes the time in which the foot is in contact with the ground. It accounts for approximately 40% of the total time spent in one walking cycle (24). It is during this phase the ground reaction forces are applied. The heel strike, mid-stance, and toe off occur during this phase of the walking cycle (5,31). The swing phase is the time in which the foot is not in contact with the ground. The swing phase accounts for approximately 60% of the gait cycle.

The goal of "normal" locomotion is the smooth translation of the center of gravity. When the body is functioning normally, gait is performed in a manner so that energy is conserved and locomotion is most efficient (31). When the body is injured or debilitated in some fashion, the body has to use more energy to compensate for the change in the normal pattern of motion (16,31). The change in position of the center of gravity follows a sinusoidal pattern (24,31) in typical gait, but can change drastically when one or more of the determinants is compromised (24,31). An injury to the knee changes not only the patterns of movement about the ankle and the hip, but also the muscular activation and torque about each joint (5,13,36). The reduced motion at the knee results in increased joint torque at the hip (5,13,36).

ACL Injury/Reconstruction and Gait

There have been a handful of studies in the last decade that examined gait patterns of ACL injured and ACL reconstructed patients. Berchuck et al studied the gait patterns of 16 ACL deficient subjects and compared them to 10 healthy subjects. Subjects walked and jogged down a 10-meter walkway, and climbed stairs. Walking speed and cadence were not discussed in the paper. Data were captured by a two-camera system and a force platform. Force plate data were collected in the middle stride of several strides. Seventy-five percent of the ACL deficient subjects were found to walk with an adapted gait pattern, while 25% had a pattern similar to the healthy subjects. ACL subjects in this study exhibited increased hip flexion moment (11.9% body weight * height) and decreased knee flexion moment (-1.3% body weight * height) when compared to healthy subjects (8.6 and 2.9). Co-contraction of the hamstrings was noted during the mid-stance phase and was coined the "quadriceps avoidance pattern" (5).

Bulgheroni et al examined the gait patterns of 15 ACL reconstructed subjects, 10 ACL injured subjects, and 5 healthy subjects. There was no discussion of walking speed or cadence used. Data were captured using an Elite three-dimensional optoelectric system, a Kistler force platform, and a Telemg electromyograph system. The ACL reconstructed subjects were found to exhibit no kinematic differences in walking compared to healthy subjects. ACL reconstructed subjects exhibited joint torque and power patterns that were different from both ACL injured and healthy subjects. These values were found to approach those of healthy subjects post-rehabilitation time increased. EMG analysis found that subjects had less activity in the rectus femoris during loading and increased activity in the vastus lateralis. This was explained as a protective mechanism to prevent "pivoting" at the knee. Greater hamstring activity throughout the movement was noted in ACL subjects. This was also the only study that discussed ground reaction forces. ACL reconstructed subjects were found to show no significant differences in vertical ground reaction forces compared to healthy subjects (6).

DeVita et al studied 9 ACL reconstructed subjects and 10 healthy subjects walking in 1997. A cadence of approximately 120 steps/minute was used by all subjects. The ACL subjects were tested 2 weeks after injury and 3 and 5 weeks after reconstruction. Data were collected using an AMTI force platform and a Sony video camera. It was determined that work done at the knee by ACL subjects 5 weeks after surgery was five times less than the work done by healthy subjects. ACL subjects exhibited more knee flexion at all phases of stance than normal subjects. Torques at the knee and hip were graphed but no numeric data were offered. Gait kinematics for ACL subjects were found to return to levels approaching those of healthy subjects at 5 weeks. Again, more work was done at the hip and less was done at the knee by ACL reconstruction subjects (9.8 J and 0.2 J) when compared to healthy subjects (8.3 J and 3.4 J). The time in which power was generated, the power phase at the hip (seconds), was also found to be one and a half times longer for ACL reconstruction subjects than for healthy subjects (13).

In 1998, DeVita et al examined the gait patterns of 8 ACL subjects who had undergone accelerated rehabilitation, and compared them to 22 healthy subjects. Data were collected using an AMTI force platform and a Sony video camera. No kinematic differences were found between patients who completed the rehabilitation and healthy subjects. ACL reconstructed subjects exhibited 77% more work performed by the hip than healthy subjects. ACL reconstructed subjects also did only 53% of the work at the knee that healthy subjects did. Two percent more work was performed at the ankle by ACL reconstruction subjects than healthy subjects did. Angular impulse was also found to differ at the hip and knee. Impulse at the knee was found to be less (5.75 Nms) in ACL subjects than in healthy subjects (9.5 Nms). Impulse at the hip was greater (12 Nms) in ACL subjects than in healthy subjects (8 Nms). Joint torque curves were again graphed, and while it was noted that joint torques at the hip and knee of ACL subjects were larger and smaller than those of healthy subjects respectively, no specific numeric data were offered (14).

Timoney et al examined the return to normal gait patterns after ACL reconstruction (36). Ten ACL reconstructed subjects were on average 10 months postsurgery and were compared to 10 healthy subjects. Data were captured using an AMTI force plate and a VICON motion analysis system. It was theorized that ACL subjects would exhibit a variation of the "quadriceps avoidance pattern" described by Berchuck et al (5), since they exhibited a smaller external knee flexion moment at mid-stance. Timoney et al. did not consider this a "true" quadriceps avoidance pattern, as there must have been an internal extension moment produced by the quadriceps to offset the external flexion moment and stabilize the knee. ACL reconstruction subjects were found to exhibit a lower loading rate (44.3 body weight/second) than healthy subjects (66.6 body weight/second). External knee flexion moment (% body weight * height) were examined for both affected and unaffected leg and for healthy subjects. ACL subjects were found to have a lower moment in the affected leg (2.02%) compared to the unaffected leg (3.10%) and control subjects (3.74%) (36).

Force Plate/Ground Reaction Forces

Ground reaction forces are measured in many movements, particularly walking and running. During locomotion, forces such as motive, resistive, and braking forces are exerted by the body on the force platform. The motive forces are then countered by reaction forces, which act to propel the body and control equilibrium (25). These forces are measured by a force platform interfaced to a computer. The force platform measures three components of the force: the vertical ground reaction force (Fz), the anterior-posterior ground reaction force (Fy), and the medial-to-lateral ground reaction force (Fx). This allows for a three-dimensional analysis of the force produced by the subject when he or she is in contact with the ground.

Vertical ground reaction forces (VGRF) in normal, heel-toe walking typically show two peaks of force with a trough between them (18,25,31,32,33). This represents the mechanical process of Fz1, Fz2, and Fz3(4). The reason for the depression during Fz2 is the fact that weight is transferred from heel to toe on the planted foot during this phase of the gait cycle (11) coupled with a downward acceleration. The acceleration then moves upward as the foot propels the body upward and forward culminating at Fz3. A braking force is applied as the heel strikes in the anterior direction, and the foot propels the body forward with a force applied in the posterior direction (25). Force is directed laterally during the heel strike, moves medially in mid-stance, and laterally in toe off (25).

Gait Symmetry

There is much debate in the area of gait research regarding symmetry between limbs. Previous studies have both supported (18,19) and refuted (11,20,35) the idea that nonpathological populations exhibit symmetry between limbs when walking. Hamill et al reported that 10 healthy individuals showed no significant differences in ground reaction forces between extremities during locomotion (18). Subjects walked and ran over a Kistler force platform and 11 vertical, five anterior-posterior, and four medio-lateral variables were examined. Upon statistical evaluation, no significant differences were found between extremities. These findings mirrored those of Hannah et al (19), who found symmetry between extremities during normal walking. This idea is not completely accepted, as evidence to the contrary has been presented repeatedly (11,35). Crowe et al examined 107 male and female subjects walking over a 12-meter walkway. Each subject completed 20 trials at a speed of approximately 1.5 m/sec. The research showed that while patterns in walking were somewhat similar, they did not match the definition of symmetrical as the body does not have perfect left-right anatomical and physiological symmetry (11). Also, the researchers determined that the gait patterns of all subjects were not consistent due to continual adjustments by the body (11). The work of Singh also showed asymmetrical patterns between legs during walking, but noted that subjects generally used one leg more than the other during walking, opening the door for subsequent leg dominance studies. Singh tested 94 medical students, recording pressure exerted by the foot, wear patterns on shoes, and which leg subjects led with when asked to walk (35). Herzog et al found that although normal gait appears to be symmetrical, it was not according to the true definition of symmetry, which requires exact correspondence between variables compared. Sixty-two subjects performed repeated walking trials at a speed of approximately one and a half meters/second along a 16-meter walkway. Data for each leg were compared using a symmetry index, a mathematical equation which directly compared the data from one leg to another, giving the results as percentages. Upper and lower limbs were found to vary from 4 to 13,000% between extremities (20).

When injury occurs in one leg, small differences may become more pronounced and can drastically affect gait patterns (1,5,13,31). Due to the interrelation of all of the determinants of gait, this may lead to long term injury and debilitation of not only the deficient joint or limb, but the rest of the body (31). One of the major concerns regarding injury to the lower extremity is meniscal damage and resultant osteoarthritis of the knee joint (17,31).

Walking Speed vs. Cadence

One of the most difficult decisions a researcher must make when implementing a research protocol for a gait study is whether or not to set speed or cadence constraints on subjects, since data promoting and refuting both exist (37,38). A 1995 study by Zhu examined eight healthy men walking and determined that cadence can greatly affect such factors as pressure-time impulses, floor-to-floor contact time, and peak-pressures. Normal cadence for subjects was found to be 109 steps/minute. When cadence was changed to 78 steps/minute and 120 steps/minute, significant differences were found in the aforementioned factors. As cadence increased, plantar pressures during walking increased and floor-to-floor contact time decreased (40).

In 1997, White et al studied 24 subjects walking both overground and on an instrumented treadmill. Subjects walked back and forth on a 12-meter runway during overground trials at a self-selected pace. That pace was then matched by use of an audio metronome so all trials were the same. Cadence recorded at self-selected normal walking pace was between 108.4 and 112.5 steps/minute. Cadence and walking speed were found to be higher on treadmill than on overground walking trials. Average vertical ground reaction forces were similar when treadmill and overground walking trials were compared. Significant differences in peak force measurements during walking were found to be speed-dependent (37).

Winter, while doing a study on the biomechanical and motor patterns in normal walking, found that hip and knee angle patterns were almost identical for three chosen cadences; slow, 84.7 steps/minute; normal, 105 steps/minute; and fast, 121.6 steps/minute (38). It was also determined that joint moments at the hip and knee were variable at all speeds. These data were compiled over the course of 6 years from several treadmill and overground walking assessments. EMG analyses also revealed that muscle activity showed consistent timing for all speeds, but increased in amplitude as walking speed increased (38).

Only one study of ACL reconstructed subjects performing locomotion reported any speed or cadence numbers. DeVita et al reported an average cadence of 120 steps/minute for 19 subjects utilizing self-selected pace (13). Timoney et al allowed 20 subjects to set a self-selected pace when walking in another ACL reconstruction and gait study (36).

<u>Summary</u>

The ACL is the most important stabilizer of the knee. Because knee flexion in stance is the most important determinant of normal gait, injury to the ACL brings with it many negative changes to the process of locomotion. Many active people choose to have ACL reconstruction in an attempt to regain some knee stability. Though the process of ACL reconstruction and subsequent rehabilitation have improved over the last decade, it cannot be determined with certainty whether gait patterns truly return to preinjured levels. If the knee has not truly recovered, the other parts of the lower extremitiy may be required to change their roles. This may affect the entire body. It is vital that further research is done to determine whether or not gait patterns do return to preinjured levels, or which factors, if any, do not. After making such discoveries, new surgical and rehabilitation procedures can be employed to improve the function and stability of the knee after reconstruction.

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