Impairment in people with anterior cruciate ligament reconstruction in adjusting ground reaction force in running

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ABSTRACT

In healthy individuals, maximum vertical ground reaction force (MVGRF) remains close to constant during constant velocity running, despite variation in stiffness of the surface underfoot. Because the anterior cruciate ligament (ACL) possesses mechanoreceptors that influence recruitment of knee muscles, it may play a role in regulation of lower limb force output. This study was designed to examine the effect of recent ACL reconstruction on MVGRF in running. Seven patients who were 5–13 weeks post-ACL reconstruction and 7 healthy participants ran for 60 seconds in shoes modified with hard and with soft 1-cm outsoles. The MVGRF during running was measured for the ACL reconstructed and nonsurgical limbs of patients and limbs of healthy participants. The difference in MVGRF between running in hard and soft shoes was significantly greater in ACL reconstructed limbs than nonsurgical limbs (p = 0.003) and compared to limbs of healthy participants (p = 0.001). In contrast, a difference in MVGRF between shoes was not found between patients’ nonsurgical limbs and those of healthy participants. A lack of mechanoreceptive feedback from the ACL graft may be among the factors explaining the difference between the ACL reconstructed limbs and the other two limb conditions.

INTRODUCTION

Serious injury of the anterior cruciate ligament of the knee is a widespread problem, especially among participants in recreational or competitive sports. According to Dugan (2005), about 1 of every 3,000 individuals in the United States suffers an ACL injury at some point; with over 100,000 occurrences reported annually. For example, in 2002, at least 7,000 ACL injuries were reported in high school female basketball players, which accounted for 1 of every 65 participants. In the physically active population, the most common treatment in the United States is surgical reconstruction (Mirzayan, 2006) to restore the mechanical restraint that the ligament provides (i.e., to prevent excessive anterior glide and internal rotation of the tibia relative to the femur) (Zantop, Peterson, Sekiya, and Fu, 2006).

However, restoration of the ACL’s mechanical restraint does not guarantee return to preinjury levels of vigorous activity, such as sports participation. Mechanoreceptive input, for both proprioception and as stimulus for muscular reflexes, is both likely to be impaired after anterior cruciate ligament reconstruction (ACLR) and to be important in full recovery after surgery. Barrett (1991) has shown high correlation (r = 0.82) between function (as measured by Lysholm knee scale, a written questionnaire designed to measure disability due to knee problems) and knee joint proprioception (as measured by accuracy in reproducing passively generated knee joint angle) in patients who had undergone ACL reconstruction.

In support of the role that mechanoreception plays in full recovery from ACLR, Bonfim, Jansen-Paccolla, and Barela (2003) reported that deficits in postural stability, reflex responses, and detection of joint angle...
change were present in patients up to 30 months postoperatively. The influence of neural input from the ACL upon reflexive muscular recruitment was described by Dyhre-Poulsen and Krosgaard (2000), who showed that electrical stimulation of large diameter afferent axons from the ACL evoked motor responses of knee flexor muscles.

Researchers (De Witt, De Clercq, and Aerts, 2000; Dixon, Collop, and Batt, 2000; Ferris, Louie, and Farley, 1998; Nigg, Bahlsen, Luethi, and Stokes, 1987) have described the following phenomenon among healthy individuals. During running at a given velocity, the average maximum vertical ground reaction force (MVGRF) remains close to constant, despite the variation of stiffness of the surface underfoot. This finding implies adjustment of lower limb force output in response to differences in stiffness of the surface underfoot. This is contrary to what would be expected based exclusively on the laws of physical mechanics without benefit of adjustment of force due to neural input. If two inert objects of the same mass are dropped onto a compliant vs. a noncompliant surface from a given height, such that they strike the surface at an equal point of acceleration in both cases, the maximum vertical force will be lower with the more compliant surface. This result is because the vertical force of impact is a product of the mass of the object times the deceleration brought about by stopping it. If the rate of deceleration is slowed by the surface, then the force of impact will be lower. The researchers cited previously have described that this is not what occurs with a healthy human running over surfaces of different stiffness.

Ferris, Louie, and Farley (1998) considered MVGRF in running to be a component of vertical lower limb stiffness. Based on empirical findings with healthy individuals during running, they concluded that the central nervous system (CNS) maintains vertical center of mass displacement that is almost invariant by adjusting lower limb stiffness in response to mechanoreceptive input regarding the surface underfoot. These authors measured the effect of changes in surface stiffness (vertical force/surface displacement) on lower limb stiffness (vertical force/vertical leg length change) and on vertical displacement of center of mass during running. They reported that the amount of the vertical displacement of the participants’ center of mass remained nearly constant, despite the variation in lower limb stiffness that occurred with changes in surface stiffness. In other words, the lower the stiffness of the surface, the greater the leg stiffness (i.e., there was an inversely proportional relationship between vertical leg length change and vertical surface deformation). Like the previously cited authors, Ferris, Louie, and Farley (1998) found MVGRF remained nearly constant (2.92 ± 0.04 times body weight), despite change in the stiffness of the surface underfoot.

In addition to the effects of surface stiffness on lower body kinetic and kinematic factors described by Ferris, Louie, and Farley (1998), there is empirical evidence that mechanoreceptive input regarding stiffness of the surface underfoot stimulates changes in muscular recruitment. Wakeling, von Tscharner, Nigg, and Stergiou (2001) found that by increasing shoe midsole stiffness by 13%, the participants (healthy recreational and competitive runners) showed an increase in the intensity of the electromyogram signal at foot-strike for tibialis anterior, soleus, gastrocnemius, biceps femoris, and vastus medialis by an average of 154%.

Additional evidence that the lack of mechanoreceptive input from the ACL may impair the muscular response to ground reaction force is described in articles published by Dyhre-Poulsen and Krosgaard (2000) and Tsuda, Ishibashi, Yoshihisa, and Toh (2003). Their findings imply that lack of neural input from the ACL may have a direct effect on knee flexors, because reflexive knee flexor muscle activity can be elicited by electrically stimulating the ACL.

Based on the evidence that stimulation of ACL mechanoreceptors influences lower limb force output, it is possible that patients who have undergone ACLR will show a degradation of the adjustment of MVGRF in response to change in stiffness underfoot. These patients have had the mechanical restraint of the ligament restored, but the mechanoreceptive nerve endings in the ACL generally have been lost through surgery. Researchers identified no mechanoreceptors in ACL grafts in a study with sheep until at least 12 weeks after surgical reconstruction (Denti, Monteleone, Berardi, and Panni, 1994). The effects of varying surface stiffness on MVGRF have been documented during running in healthy humans (De Witt, De Clercq, and Aerts, 2000; Dixon, Collop, and Batt, 2000; Ferris, Louie, and Farley, 1998; Nigg, 2001; Nigg, Bahlsen, Luethi, and Stokes, 1987). However, no studies have been conducted on the effect of variation of stiffness of supporting surface during running on people with any sort of medical impairment. Therefore, we hypothesized that patients who had undergone recent ACLR (between 5 and 13 weeks postsurgery) would have greater variability (defined as the difference in MVGRF between shoe conditions) during running on surfaces of different stiffness on their postsurgical limbs compared to their own nonsurgical limbs and compared to the limbs of healthy participants. We hypothesized also that there would be no difference in variability in MVGRF associated with difference in underfoot surface stiffness, between the nonsurgical limbs of patients and those of healthy participants.


**MATERIALS AND METHODS**

**Participants**

Seven patients (four females and three males) participated in this study. Their ages ranged from 15 to 30 years (mean = 20.5 ± 5.3), and their body mass ranged from 42 to 94 kg (mean = 65.8 ± 17.6). The patients had all undergone ACLR (and in one case minor meniscectomy) for unilateral anterior cruciate ligament rupture. In four cases, the graft source was patellar tendon allograft, in two cases it was patellar tendon autograft, and in one case, it was hamstring tendon autograft. By the time of experimental trial, all patients had met the criteria to return to running as part of their rehabilitation. These criteria included being free of pain; having full knee extension and full or close to full knee flexion range of motion; and having manual muscle test grade of the postsurgical limb for knee extension, which was categorically the same as that of the nonsurgical limb (e.g., 4+/5), and likewise for knee flexion. The range of time between surgery date and experimental trial was 35–98 days (mean = 67 days ± 22). In each case, data collection took place within 2 weeks of adding running to the patient’s rehabilitation program.

The healthy participants (four females and three males) were volunteers who were matched for gender and approximate age to the patients. Their ages ranged from 16 to 28 years (mean = 19.3 ± 4.0), and body mass ranged from 50 to 86 kg (mean = 70.5 kg ± 12.8). They were without history of knee surgery or significant knee injury. All participants, patients and healthy individuals, completed the informed consent process as directed by the Human Subjects Committee of the Institutional Review Board of the University of Minnesota.

**Instrumentation**

**PEDAR insole foot vertical force measurement system**

The PEDAR insole vertical force measurement system (Novel Electronics, Munich, Germany) was used to measure MVGRF. The data-gathering components of this instrument are insoles that are shaped like the insole of a shoe and are constructed of a matrix of 99 sensors, each with an effective sensor area of approximately 1.5 cm². Each sensor consists of two electromagnetic surfaces separated by a 2-mm foam core between the superior and inferior surfaces. When the insole is compressed, the current between the sensors decreases in proportion to the decrease in distance between them, thereby converting capacitance into pressure data. Because of the mathematical relationship between force and pressure, force = pressure X area, a researcher is easily able to analyze data either as pressure or as vertical force.

The sampling rate of the PEDAR used in this study was 50 Hz (once every 20 milliseconds). Researchers have previously demonstrated that the shapes of vertical force profiles generated by the PEDAR at 50 Hz correlated closely to those generated by force plates sampling at 99 Hz, r = 0.99 (Barnett, Cunningham, and West, 2000) and at 1,000 Hz, r = 0.95 (Cordero, Koopman, and van der Helm, 2004).

As described in the article by Clarke, Frederick, and Cooper (1983), the propulsive peak ground reaction force reaches maximum at about 85 milliseconds. The propulsive peak is the parameter of interest in this study because it reflects (among other factors) the output of muscular tensions. Therefore, in our study (using the sampling rate of 50 Hz), we expected that the foot contact force profile would be sampled at least 4 times before it reached the propulsive peak vertical force and at least 10 times for even the shortest foot contact period (Figure 1).

**Modified shoes**

Shoes (“Athletic Works, Major,” Wal-Mart, Bentonville, AR, USA) were modified by gluing an outsole attachment to the factory outsole (Figure 2). The outsole attachment was a 1-cm-thick pad of ethylvinyl-acetate, glued onto each of six sets of identical shoes of successive sizes (American men’s sizes 6 1/2, 7 1/2, 8 1/2, 9 1/2, 10 1/2, and 12). There were two pairs of shoes in each size, identical to each other except for the stiffness of the outsole. The soft outsole had a durometer rating of 45 Shore, and the hard outsole a rating of 55 Shore. The modified shoes had a semicurved last to accommodate the maximum number of participants.
Procedure

This study used a crossover design, with the order of treatment (shoe condition) alternated within the participants for both patient and healthy groups. The participants read and signed the consent form and then were weighed for body mass. Data collection with the postsurgical participants took place at the end of their rehabilitation sessions.

The PEDAR insoles of appropriate size were placed inside the shoes by the primary investigator before the participants donned them. The data cables from the insoles to the mobile data-gathering unit were secured to the participants’ ankles and proximal shins with elastic bands. The mobile data-gathering unit was cable connected to a computer. The insole was unloaded on each side by the subject lifting each of his or her feet. This allowed the PEDAR to be calibrated for each pair of insoles before each trial (session of running in one of the shoe conditions) by establishing a “zero vertical force” on each side.

During each trial, each participant had a 60-second period of treadmill walking for warm-up prior to running and then was asked to self-select the trial speed by advancing the treadmill speed according to his or her comfort in running. Once they had reached their self-selected speed, he or she signaled the investigator to begin data collection, which continued for 60 seconds. Each participant was required to use the same running speed for both the hard and soft shoe conditions.

Figure 1 portrays an example of the output of the PEDAR from a single foot contact. In the vertical force profile generated, the passage of time is represented on the x-axis, and force underfoot is represented on the y-axis. For each force profile, the 60 consecutive milliseconds with the highest force for each foot contact (representing the interval enclosed by four data points per each foot contact, as portrayed on Figure 1) was averaged to generate a peak force value for that foot contact and then all resulting force values from every foot contact within the 60-second trial were averaged to generate a mean MVGRF value for that limb for that shoe condition. This ranged from 60 to 100 foot contacts per limb, depending on the cadence of the participant.

During the running trials, the mobile data-gathering unit was supported by the investigator standing beside the participant. The vertical ground reaction force data collection took place in real time via cable connection from the mobile data-gathering unit to a computer. The procedure was performed twice with each participant, one time for each shoe condition. The data were normalized by dividing resultant Newtons of force by kilograms of body mass.

Statistical analysis

We calculated a “difference score” (the difference in mean MVGRF per kg body mass generated by each limb between the running trials in the hard vs. the soft shoes), which was used as the dependent variable. The data for 7 ACL reconstructed limbs, 7 nonsurgical limbs of patients, and 14 limbs of healthy individuals were compared by using a one-way ANOVA. The difference scores for the right and left lower limbs of healthy subjects were averaged, so that there were seven difference scores contributing to the mean for each limb condition. The difference scores between 1) the ACL reconstructed limbs and nonsurgical limbs of the patients, 2) the ACL reconstructed limbs and those of healthy participants, and 3) the nonsurgical limbs of patients and those of healthy participants were analyzed with planned pairwise comparisons. The difference scores between the left and right limbs of the healthy participants were analyzed with a paired t-test to ensure that limb differences could not be ascribed to limb dominance. Statistical analysis was done by using the SPSS Graduate Package of Statistical Software for Social Sciences, Version 10.

RESULTS

Participant means and standard deviations in MVGRF during running for each limb in both shoe conditions, along with difference scores for each limb, are displayed in Tables 1 and 2. The mean difference scores for each limb group were 1.81 ± 1.47 N/kg body mass for the ACL reconstructed limbs, 0.04 ± 0.93 N/kg body mass for the nonsurgical limbs of postsurgical
The one-way ANOVA of difference scores between limb conditions was significant (F = 8.30; df = 27; p = 0.002). Planned pairwise comparisons showed that the mean difference score for the ACL reconstructed limbs was significantly greater than that of the nonsurgical limbs of patient participants (df = 1; p = 0.003) and than that of healthy participants (df = 1; p = 0.001). This finding supported our study’s first hypothesis (i.e., patients who had undergone recent ACLR had greater variability in MVGRF during running on surfaces of different stiffness on their postsurgical limbs than their own nonsurgical limbs and those of healthy participants).

There was no significant difference in MVGRF between shoe conditions for nonsurgical limbs compared to those of control participants (df = 1; p = 0.986). This finding supported the second hypothesis (i.e., there

### TABLE 1 Patient limb per shoe condition means, standard deviations, and difference scores

<table>
<thead>
<tr>
<th>Subject no.</th>
<th>Mean (SD) MVGRF surgical limbs (N/kg body mass)</th>
<th>Mean (SD) MVGRF nonsurgical limbs (N/kg body mass)</th>
<th>Difference score surgical</th>
<th>Difference score nonsurgical</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Hard soft</td>
<td>Hard soft</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>15.21(2.41)</td>
<td>15.53(2.45)</td>
<td>1.17</td>
<td>-0.02</td>
</tr>
<tr>
<td>2</td>
<td>19.72(3.14)</td>
<td>14.34(2.28)</td>
<td>4.96</td>
<td>-1.02</td>
</tr>
<tr>
<td>3</td>
<td>18.37(2.91)</td>
<td>20.26(3.13)</td>
<td>1.11</td>
<td>0.67</td>
</tr>
<tr>
<td>4</td>
<td>16.98(2.35)</td>
<td>20.32(3.36)</td>
<td>2.08</td>
<td>1.58</td>
</tr>
<tr>
<td>5</td>
<td>13.39(2.25)</td>
<td>14.13(2.39)</td>
<td>0.80</td>
<td>-0.36</td>
</tr>
<tr>
<td>6</td>
<td>11.38(2.57)</td>
<td>9.67(2.23)</td>
<td>1.20</td>
<td>-0.82</td>
</tr>
<tr>
<td>7</td>
<td>15.65(2.52)</td>
<td>16.58(2.66)</td>
<td>0.79</td>
<td>0.26</td>
</tr>
</tbody>
</table>

### TABLE 2 Healthy participant limb per shoe condition means, standard deviations, and difference scores

<table>
<thead>
<tr>
<th>Subject no.</th>
<th>Mean (SD) MVGRF left (N/kg body mass)</th>
<th>Mean (SD) MVGRF right (N/kg body mass)</th>
<th>Difference score left</th>
<th>Difference score right</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Hard soft</td>
<td>Hard soft</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>19.31(3.32)</td>
<td>19.62(3.34)</td>
<td>0.29</td>
<td>0.84</td>
</tr>
<tr>
<td>2</td>
<td>22.05(3.32)</td>
<td>22.17(3.42)</td>
<td>0.06</td>
<td>-0.41</td>
</tr>
<tr>
<td>3</td>
<td>19.29(2.03)</td>
<td>19.68(3.18)</td>
<td>0.56</td>
<td>0.99</td>
</tr>
<tr>
<td>4</td>
<td>15.98(1.08)</td>
<td>14.29(2.24)</td>
<td>0.27</td>
<td>0.96</td>
</tr>
<tr>
<td>5</td>
<td>21.83(3.46)</td>
<td>22.60(3.61)</td>
<td>-1.58</td>
<td>-1.53</td>
</tr>
<tr>
<td>6</td>
<td>16.47(2.60)</td>
<td>16.40(2.58)</td>
<td>-0.48</td>
<td>0.30</td>
</tr>
<tr>
<td>7</td>
<td>18.11(2.87)</td>
<td>18.26(2.90)</td>
<td>0.16</td>
<td>-0.23</td>
</tr>
</tbody>
</table>

An image of Figure 3 showing mean difference scores in MVGRF between limb conditions.
was not a significant difference in MVGRF variability associated with surfaces of different stiffness between those two limb groups).

A paired t-test revealed no significant difference between left and right limbs in the control group ($d = 6; p = 0.27$). Levene’s statistic generated by the sample of all within-subject difference scores was 0.71, yielding a $p = 0.50$, signifying adequate homogeneity of variance at $\alpha \leq 0.05$.

**DISCUSSION**

The data imply that MVGRF response to the stiffness of the surface underfoot in running is affected by postsurgical status. The MVGRF for the soft shoe condition was significantly less than it was for the hard shoe condition with the ACL reconstructed limbs compared to the patients’ nonsurgical limbs and to that in limbs of healthy participants. This finding supported our study’s first hypothesis. There was not a significant difference in MVGRF between shoe conditions in running for the nonsurgical limbs of patients and that measured in the limbs of healthy participants; a finding that supported our study’s second hypothesis. There are multiple feasible explanations for this finding related to postsurgical status, including impoverishment of neural input from the insensate ACL graft or a deficit in the speed in which the lower extremity muscles could be recruited.

In support of the explanation that an insensate ACL graft impairs the muscular response to change in stiffness of the underfoot surface, an argument may be constructed based partially on a review article written by Winter and Eng (1995). In it, they described that intensity of muscular recruitment around a joint during gait is poorly predictable. According to these authors, what is predictable is the net moment of support (which results in vertical ground reaction force) and is the result of the net joint moments. To maintain constant net support moment despite the variation of the stiffness underfoot, moments crossing all lower body joints must be coordinated.

The vertical ground reaction force is the result of the runner’s force of gravity plus vertical components of all muscle and soft tissue forces acting across the lower body joints, minus the externally applied forces that must be counteracted by the internal muscle and connective tissue forces. One possible explanation for the findings of the present study is that, with impoverished knee proprioception due to an insensate ACL graft, the CNS may have difficulty coordinating the vertical knee forces in response to changing support conditions and therefore difficulty adjusting the total vertical support force. Contraction of the quadriceps causes anterior translation of the tibia, and one of the functions of the ACL as a sensory organ is to facilitate hamstring muscle activation to constrain this translation (Tsuda, Ishibashi, Yoshihisa, and Toh, 2003). It is possible that the impairment in adjustment of force output is a result of diminished input to the hamstrings as a result of lack of kinesthetic input from the ACL.

The findings described by Ferris, Louie, and Farley (1998), by Moritz, Greene, and Farley (2004), and by Moritz and Farley (2005) provide a potential explanation for a neuromuscular control mechanism that may be impaired by ACL rupture, subacutely after surgical reconstruction. The authors demonstrated that lower limb stiffness is adjusted in response to surface stiffness to maintain nearly constant vertical excursion of the center of mass. It is possible that in patients who have undergone ACL reconstruction, the impairment in adjusting MVGRF response to change in underfoot stiffness is the inability to adjust lower limb stiffness in response to mechanoreceptive input. Simultaneous kinematic data are necessary to test that hypothesis.

An explanation for our finding of lower MVGRF with the ACL reconstructed limbs in running with soft shoes, alternative to that of diminished knee joint proprioception, may involve a postsurgical impairment in muscular recruitment rate. LaStayo et al (2003) described that the speed of generating force after lengthening contraction of lower limbs is, in part, a function of training. As one practices exercise that involves a stretch shortened cycle, such as running, then that individual will increase the rate at which he or she is able to generate force in recoil of the limbs after compression. However, lack of this type of training is bilateral in these patients, and because lower force with the soft shoe condition was seen only on the surgical side, the training deficit does not explain the effect. It may be, however, that a decrease in the rate of muscular recruitment with a less stiff surface underfoot is an effect of any surgical disruption of connective tissue or resulting immobilization. A follow-up study intended to investigate that possibility is also proposed.

The subacute status of the surgery was integral to the design of the study because the intention was to attempt to capture MVGRF behavior before significant reinnervation of the graft had likely started (Denti, Monteleone, Berardi, and Panni, 1994). It is reasonable to conclude that the effect of limb status on MVGRF difference score between shoe conditions was not merely due to the novelty of running. The novelty of running was obviously equal for both nonsurgical and the ACL reconstructed limb, but the mean difference scores between those two limb groups were significantly different: 1.81 N/kg body mass between shoe...
conditions on the surgical side vs. 0.04 N/kg body mass on the nonsurgical side.

Potential limitations to the conclusions of this study include 1) the degree of intersubject variability in MVGRF during running; 2) the possibility that output of extraneous muscular actions of the subjects’ feet, unrelated to ground reaction force, was included in the raw data; and 3) the question about whether the PEDAR’s 50 Hz sampling rate was adequate to capture the data relevant to the experimental questions.

For the first potential limitation, that of intersubject variability, researchers have described that with healthy individuals, the MVGRF of running varies linearly with velocity (Arampatsis, Knicker, Metzler, and Bruggeman, 2000; Nigg, Bahlsen, Luethi, and Stokes, 1987). Based upon those findings, the self-selection of running velocity for the trials by participants would therefore be expected to introduce the covariate of velocity in the dependent variable of MVGRF per kilogram body mass. However, when the group means (mean MVGRF scores for each level of limb group at both levels of shoe condition) of this sample were analyzed through ANCOVA, such treatment of the data did not provide adequate statistical power. This implies that the differences between individuals with this sample were likely more attributable to random error than to the covariate of velocity. Therefore, one-way ANOVA of difference scores was chosen as the method to minimize the effect of intersubject variability due to random error.

For the second potential issue, the method used to calculate a mean MVGRF for each trial most likely generated a mean value that did not include extraneous variability. This conclusion is based on the reliability of the coefficients of variation (standard deviation/mean) for limb group for each shoe condition. Partitioning each limb group by each shoe condition resulted in six possible combinations of levels, and in each combination, the coefficient of variation was between 15% and 17%, demonstrating a high degree of reliability in the size of the standard deviation relative to the mean. This reliability of the coefficient of variation would not likely have been seen in the presence of random, extraneous forces.

The issue of sampling rate presents a potential limitation in the interpretation of the data in this study. Although empirical data show that ground reaction force sampled at 50 Hz is strongly correlated to data sampled at higher frequencies (Barnett, Cunningham, and West, 2000; Cordero, Koopman, and van der Helm, 2004), theoretically it can be argued that this sampling rate may not be adequate. If the force profile is considered as the top phase of a sine wave, the highest frequency of such a wave generated by our data was 2.5 Hz. The 50-Hz sampling rate samples this wave every 18 degrees; therefore, the greatest deviation from the actual peak may be sine of 72 degrees, the value of which is 0.951. Therefore, the greatest amount of potential error from the actual peak with the shortest duration force profile is almost 0.05 (Qian and Chen, 1996), implying that the actual peak force may not have been sampled in several of the force profiles of the data.

Directions for further study

The findings described in our study suggest that recently ACL reconstructed limbs are impaired in the ability to maintain an MVGRF that is close to constant, despite change in the stiffness of the surface underfoot (as manipulated through shoe hardness), compared to the patients’ nonsurgical limbs and to those of healthy participants. We did not find a difference in MVGRF, during running on surfaces of different stiffness, between the patients’ nonsurgical limbs and those of healthy participants. Because this study is novel in investigating the phenomenon of reliability of MVGRF in the presence of any sort of musculoskeletal impairment, what we cannot validly conclude is whether the impairment is truly the result of diminishment of neural input as a result of an insensate ACL graft as proposed or if a similar impairment would result from any surgical disruption of knee connective tissues.

Therefore, a possible follow-up study is a similar design with postmeniscectomy patients. This would help to illuminate whether surgical intervention without resection of an innervated structure might have a similar effect on MVGRF behavior, because partial meniscectomy usually involves resection only of the interior rim of the meniscus, which is not an innervated tissue (Gray, 1999). If the findings for the behavior of MVGRF in partial meniscectomy patients are similar to those with ACL reconstruction patients, it would make the explanation of neural impoverishment less likely.

Another direction for follow-up study regards which biomechanical variable is being controlled by the CNS in response to stiffness of the surface underfoot. Ferris, Louie, and Farley (1998), Moritz and Farley (2005), and Moritz, Greene, and Farley (2004) describe the effects of surface on lower limb vertical stiffness. Vertical force, which was the dependent variable in our study, is only one of the biomechanical variables necessary to determine limb stiffness; the other variable is vertical displacement. It is possible that the finding of the present study of an impaired MVGRF response to change in underfoot stiffness is the result of impairment in the ability to adjust lower body stiffness in response to mechanoreceptive input. If this hypothesis is supported empirically, it could contribute knowledge potentially important in understanding the control of biomechanical variables in human gait.
CONCLUSION

This study examined the ability of patients who had undergone recent ACL reconstruction to maintain MVGRF with the surgically reconstructed limb, which remains close to constant during constant velocity running, despite variation in stiffness of the surface underfoot. The difference in MVGRF between running in hard vs. soft shoes was greater for the surgical limbs than for the patients’ nonsurgical limbs and for the limbs of healthy control participants. The shoe hardness variation caused no significant difference between the patients’ nonsurgical limbs and those of healthy subjects. The ACL is equipped with mechanoreceptive nerve endings, stimulation of which has been shown to cause reflex responses in knee muscles. Therefore, one potential explanation for this study’s finding was that lack of neural input from the ACL graft causes a deficit of the involved limb in adjusting force output in response to mechanoreceptive input regarding the stiffness of the surface underfoot.

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