A FUNCTIONAL KNEE BRACE ALTERS JOINT TORQUE AND POWER PATTERNS DURING WALKING AND RUNNING

Paul DeVita,* Michael Torry,' Kathryn L. Glover' and David L. Speroni'

*Department of Exercise and Sport Science, East Carolina University, Greenville, NC 27858, U.S.A.
'Department of Physical Education, Southern Illinois University at Carbondale, Carbondale, IL 62901, U.S.A.

Abstract—Individuals with anterior cruciate ligament (ACL) injury use greater extensor torques at the hip and ankle and lower extensor torques and joint power at the knee during gait compared to healthy subjects. These adaptations may be mediated by (1) altered neuromuscular strategies due to the injury, (2) training effects produced by rehabilitation protocols, and (3) training effects due to the functional knee brace (FKB) used during rehabilitation. The purpose of the study was to test the hypothesis that a FKB can cause individuals to walk and run with the torques and power patterns observed in rehabilitated ACL-injured individuals. Ten healthy subjects were tested walking and running with and without a FKB. Kinematic and ground reaction data were collected and combined with inverse dynamics to estimate the joint torques and powers. Data were analyzed with a two-way repeated measures ANOVA (gait vs knee condition). In walking, the hip and ankle extensor torques were 14.3% (p < 0.038) and 5.1% (p < 0.003) greater with FKB. In running, the hip extensor torque was 17.0% greater with FKB (p < 0.023). Knee torque was not different between conditions. In walking, the work performed at the hip and knee were 11.6% greater (p < 0.013) and 17.7% lower with FKB (p < 0.025), respectively. Results supported the hypothesis and it was concluded that a FKB may be one causative factor in the development of the unique joint torque and power patterns seen in ACL-injured gait.

Keywords: ACL injury; Knee brace; Gait; Walking; Running.

INTRODUCTION

Individuals with rehabilitated anterior cruciate ligament (ACL) injury perform various movements with different joint torque and power patterns in the lower extremity compared to noninjured individuals. ACL-injured individuals use greater extensor torques at the hip and ankle and a reduced extensor torque at the knee during the stance phase of running (Berchuck et al., 1990; DeVita et al., 1992). The same adaptations at the hip and knee have been observed in walking (Andriacchi and Birac, 1993; Berchuck et al., 1990; Noyes et al., 1992; Timoney et al., 1993). Joint power at the knee was also reduced in both the absorption and generation phases in the stance phase of running, indicating that ACL-injured individuals change the source of mechanical work performed during this movement (DeVita et al., 1992). Similar reductions in the maximum extensor torque at the knee were identified during pivoting movements on the previously injured limb (Andriacchi and Birac, 1993; Shiavi et al., 1991). Numerous isokinetic and isometric evaluations of rehabilitated ACL-injured subjects have also shown reduced extensor torque in the involved knee compared to the contralateral limb and to noninjured subjects (Elmquist et al., 1988; Kannus et al., 1987; Murray et al., 1984; Rosenberg et al., 1992; Tibone et al., 1986).

These results have been supported by observations of increased electromyographic (EMG) activity in the hamstrings and medial gastrocnemius and reduced EMG activity in the quadriceps in ACL-injured individuals during gait (Branch et al., 1989; Kalund et al., 1990; Lass et al., 1991; Limbird et al., 1988; Shiavi et al., 1992; Shiavi and Limbird, 1988; Sinkjaer and Arendt-Nielsen, 1991; Tibone et al., 1986), in pivoting maneuvers (Shiavi et al., 1991), and in isokinetic testing (Elmquist et al., 1988; Grabiner and Weiker, 1993; Solomonow et al., 1987).

It is the consensus that the torque, power and EMG adaptations are beneficial to ACL-injured individuals because they reduce anterior displacement of the tibia relative to the femur and therefore reduce stress on the ACL while they also enable the subjects to perform the desired movement (e.g. Andriacchi and Birac, 1993; Berchuck et al., 1990; Branch et al., 1989; DeVita et al., 1992; Grabiner and Weiker, 1993; Hirokawa et al., 1991; Sinkjaer and Arendt-Nielsen, 1991).

These adaptations may be mediated by (1) altered neuromuscular strategies due to the injury, (2) training effects produced by rehabilitation protocols, and (3) training effects due to the functional knee brace used during rehabilitation (Andriacchi and Birac, 1993; Berchuck et al., 1990; Branch et al., 1989; DeVita et al., 1992; Shiavi et al., 1992). None of these mechanisms have been directly tested, although some empirical results suggest that neuromuscular changes due to the injury...
(Andriacchi and Birac, 1993; Berchuck et al., 1990; Shiavi et al., 1992) and the functional knee brace (DeVita et al., 1992) may at least partially explain the adaptations. The purpose of this study was to test the hypothesis that a functional knee brace can cause individuals to walk and run with the torque and power patterns observed in rehabilitated ACL-injured individuals. Specifically, it was hypothesized that subjects would have greater extensor torques and joint powers at the hip and ankle and reduced extensor torque and joint power at the knee during the stance phases of walking and running while performing with a functional knee brace compared to the same movements performed without the brace.

METHODS

Five males and five females (mean age: 20.9 yr, mean mass: 65.8 kg) with no history of lower limb pathology volunteered for the study. All subjects gave informed consent according to University policy. An AMTI force platform located in the center of a 20 m runway was used to measure ground reaction forces at 1000 Hz and a Locam II 16 mm camera was used to film the subjects in the sagittal plane at 100 Hz. Walking and running speeds were set at 1.82 and 3.83 m s⁻¹ and were monitored with a photoelectric timing system.

Circumference measures at the upper thigh, knee, ankle, and metatarsal heads were obtained for each subject along with body weight. The experimental protocol included walking and running with and without the functional knee brace. Ten trials were collected for each subject condition treatment, and the order of the conditions was counterbalanced across subjects.

A functional knee brace is designed to apply sufficient force onto the extremity to prevent abnormal motion, particularly excessive anterior tibial displacement, in the ACL-injured knee, while allowing normal kinematics for skill performance. These braces typically include rigid metallic bars placed on the medial and lateral portions of the thigh and leg and a hinge to provide flexion and extension movements. An Omni Scientific OS-5 functional knee brace [see DeVita et al. (1992) for photograph] was applied to the right knee of each subject for the brace conditions according to the manufacturer's guidelines. The brace was made with a uniaxial hinge, post, and strap design, weighed 8 N and was 0.39 m long and were monitored with a photoelectric timing system.

The magnitude of the segmental masses and the mass center locations of the lower extremity along with their moments of inertia were estimated using a mathematical model (Hanavan, 1964), segmental masses reported by Dempster (1955), and the individual subject's anthropometric data. Joint torques were estimated for the lower extremity through an inverse dynamic analysis combining the anthropometric, kinematic, and kinetic data. The weight of the brace was included in the analysis. Joint powers were estimated as the product of the joint torques and joint angular velocities. The area under the positive phases of the torque and power curves representing extensor angular impulse and positive work were calculated during the entire stance phase.

These variables were entered into a two-way analysis of variance (SuperANOVA) with repeated measures to test the hypothesis. One a priori direct comparison was made between the brace and no-brace group means in each gait mode for each dependent variable using an alpha level of 0.05. The 5% probability of making a Type I error was deemed acceptable because the underlying nature of the hypothesis was to identify the possibility of the brace causing changes in gait which supports the use of a safe but not excessively stringent alpha level. Also, the number of comparisons tested was less than the orthogonal amount for the design of the study and therefore no adjustment to the selected alpha level was performed.

RESULTS

The functional knee brace caused a 14.3% increase (p < 0.038) in the extensor angular impulse at the hip and a 5.1% increase (p < 0.003) in the plantarflexor angular impulse at the ankle, averaged across all subjects, during walking (Fig. 1, Table 1). The brace caused the subjects to use a 17.0% greater (p < 0.023) extensor angular impulse at the hip during running (Fig. 2, Table 1). The torque variables at the knee joint were not statistically different between the conditions in walking or running (Figs 1 and 2).

Two statistically significant effects were identified in the joint powers. The positive work performed at the hip during the stance phase of walking was increased 11.6% (p < 0.013) in the brace condition compared to the no-brace condition (Fig. 3, Table 1). The work performed at the knee during the stance phase of walking was decreased 17.7% (p < 0.025) in the brace condition (Fig. 3, Table 1). The joint powers were not significantly affected by the functional knee brace during running (Fig. 4).

Body landmarks including the fifth metatarsal head, back edge of the shoe, lateral malleolus, lateral femoral condyle, greater trochanter, and shoulder and the front corner of the force platform were digitized during the stance phase of the tested limb and in four extra frames before and after the stance phase to improve the accuracy of the data near the performance boundaries (Woltring, 1985). The kinematic data were then smoothed using an interactive cubic spline. The center of pressure was calculated and used to spatially coordinate the kinematic and kinetic data.
A functional knee brace alters joint torque and power patterns during walking and running.

Table 1. Individual subject data for statistically significant variables

<table>
<thead>
<tr>
<th>Sub.</th>
<th>Walking</th>
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<th>Running</th>
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<tr>
<td></td>
<td>Hip torque</td>
<td>Ankle torque</td>
<td>Hip work</td>
<td>Knee work</td>
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<td>1</td>
<td>14.5</td>
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<td>34.3</td>
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<td>2</td>
<td>11.5</td>
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<tr>
<td>3</td>
<td>11.1</td>
<td>9.8</td>
<td>19.3</td>
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<td>4</td>
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<td>25.9</td>
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<td>5</td>
<td>8.9</td>
<td>7.2</td>
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<td>6</td>
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<td>10</td>
<td>6.4</td>
<td>7.9</td>
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<td>20.5</td>
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<tr>
<td>Mean</td>
<td>12.6</td>
<td>14.4</td>
<td>31.4</td>
<td>33.0</td>
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<tr>
<td>S.D.</td>
<td>3.9</td>
<td>6.1</td>
<td>10.0</td>
<td>10.4</td>
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Individual subject values are means of all trials in the condition. No-br. is the no-brace condition. Br. is the brace condition. Torque variables are angular impulse produced by the extensor torque during stance; units are Nm.s. Work variables are work produced by extensor torques during stance; units are J.

Fig. 1. Joint torque curves from a representative subject during the stance phase of walking. Positive values are net extensor or plantarflexor torques. Solid line and dotted lines are mean ± 1 S.D. in the no-brace condition and dashed line is mean in the brace condition. The variance in the brace condition was similar to that in the no-brace condition (Table 1) and was not displayed to improve the clarity of the graph. While walking with the functional knee brace, the angular impulse in the extensor direction was increased 14.3% (p < 0.038) at the hip during the initial half of stance and 5.1% (p < 0.003) at the knee for all subjects compared to walking without the brace. The location of the increased ankle torque varied across subjects and was evident for this subject at about 20–50% of stance. The angular impulse at the knee was statistically identical between conditions.

Fig. 2. Joint torque curves from a representative subject during the stance phase of running. Positive values are net extensor or plantarflexor torques. Solid line and dotted lines are mean ± 1 S.D. in the no-brace condition and dashed line is mean in the brace condition. The variance in the brace condition was similar to that in the no-brace condition (Table 1) and was not displayed to improve the clarity of the graph. While running with the functional knee brace, the angular impulse in the extensor direction was increased 17.0% (p < 0.023) at the hip during the initial half of stance compared to running without the brace. The angular impulses at the knee and ankle were statistically identical between conditions.
The hypothesis tested in this study addressed the question, what factors cause ACL-injured individuals to develop their unique torque and power patterns? The hypothesis was derived from the combined results of a previous investigation and the pilot study associated with that investigation (DeVita and Blankenship, 1989; DeVita et al., 1992). The pilot study showed that the functional knee brace caused a healthy individual to run with greater extensor torque and power at the hip and reduced extensor torque and power at the knee compared to running without the brace. These people used the ACL-injured pattern of greater extensor torques at the hip and ankle and reduced extensor torques at the ankle and ankle and reduced extensor torque and power at the knee compared to healthy runners. From these results it was suggested that the functional knee brace would cause an individual to use the ACL-injured torque and power patterns if the person did not already do so and the brace would not affect individuals who already used the ACL-injured patterns. The present study was the first known attempt to identify the cause of the adaptations in the gait of ACL-injured individuals.

A primary limitation in the study was the use of healthy instead of ACL-injured subjects. Healthy subjects were used in this initial test of the hypothesis to eliminate the possibility that the rival adaptive mechanisms described herein may have caused the observed results. Indeed, the observation of the expected adaptations in healthy subjects while performing with the functional knee brace would strongly support the hypothesis since no other factors would cause these changes in this population. In contrast, a protocol involving injured subjects performing with and without a brace may not provide valid results because each subject may rely on one causative factor (e.g. the brace) to produce the adaptations and then rely on another factor if the first is removed. Further depending on the adaptation present protocol brace can cause instead of energy generation and absorption. Solid line and dotted lines are mean ± 1 S.D. in the no-brace condition and dashed line is mean in the brace condition. The variance in the brace condition was similar to that in the no-brace condition (Table 1) and was not displayed to improve the clarity of the graph. The functional knee brace did not affect the joint powers during running and all sample means were statistically identical between conditions.

DISCUSSION

The observed adaptations seen in ACL-injured subjects performing with and without a brace may not provide valid results because each subject may rely on one causative factor (e.g. the brace) to produce the adaptations and then rely on another factor if the first is removed. Further depending on the adaptation present protocol brace can cause instead of energy generation and absorption. Solid line and dotted lines are mean ± 1 S.D. in the no-brace condition and dashed line is mean in the brace condition. The variance in the brace condition was similar to that in the no-brace condition (Table 1) and was not displayed to improve the clarity of the graph. The functional knee brace did not affect the joint powers during running and all sample means were statistically identical between conditions.

Fig. 3. Joint power curves from a representative subject during the stance phase of walking. Positive and negative values indicate energy generation and absorption. Solid line and dotted lines are mean ± 1 S.D. in the no-brace condition and dashed line is mean in the brace condition. The variance in the brace condition was similar to that in the no-brace condition (Table 1) and was not displayed to improve the clarity of the graph. While walking with the functional knee brace, the positive work performed at the hip was increased 11.6% (p < 0.013) during the initial 40% of stance and this increase was directly attributable to the increased torque observed at this time. Walking with the brace also caused a 17.7% (p < 0.025) decrease in the positive work performed at the knee for all subjects. The decreased knee power was evident in all three phases of positive power. The work performed at the ankle was statistically identical between conditions.

Fig. 4. Joint power curves from a representative subject during the stance phase of running. Positive and negative values indicate power generation (concentric muscle contraction) and power absorption (eccentric muscle contraction). Solid line and dotted lines are mean ± 1 S.D. in the no-brace condition and dashed line is mean in the brace condition. The variance in the brace condition was similar to that in the no-brace condition (Table 1) and was not displayed to improve the clarity of the graph. The functional knee brace did not affect the joint powers during running and all sample means were statistically identical between conditions.
movements investigated, the subjects' characteristics, or the variables measured.

It was hypothesized that a functional knee brace will cause individuals to walk and run with joint torque and power patterns previously associated with ACL-injured gait. A group of healthy individuals was evaluated as they walked and ran with and without a single functional knee brace. While walking with the brace, the subjects used greater extensor torques at the hip and ankle and produced more work at the hip and less at the knee compared to walking without the brace. While running with the brace, the subjects had a greater extensor torque at the hip compared to running without the brace. All of these changes are consistent with those exhibited by individuals who have sustained ACL injury compared to noninjured individuals. The hypothesis was therefore supported and it was concluded that a functional knee brace may be one causative factor in the development of the unique joint torque and power patterns seen in ACL-injured gait.

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REFERENCES


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