Electromyographic and biomechanic analysis of anterior cruciate ligament deficiency and functional knee bracing

Dan K. Ramsey, Per F. Wretenberg, Mario Lamontagne, Gunnar Németh

Objective. Examine the neuromuscular response to functional knee bracing relative to anterior tibial translations in vivo.

Design. During randomised brace conditions, electromyographic data with simultaneous skeletal tibiofemoral kinematics were recorded from four anterior cruciate ligament deficient subjects to investigate the effect of the DonJoy Legend functional brace during activity.

Background. Knee braces do not increase knee stability but may influence afferent inputs from proprioception and therefore one might expect changes in muscle firing patterns, amplitude and timing.

Methods. Hoffman bone pins affixed with markers were implanted into the tibia and femur for kinematic measurement. The EMG data from the rectus femoris, semitendinosus, biceps femoris, and lateral head of the gastrocnemius were integrated for each subject in three separate time periods: 250 ms preceding footstrike and two consecutive 125 ms time intervals following footstrike.

Results. With brace, semitendinosus activity significantly decreased 17% prior to footstrike whereas biceps femoris significantly decreased 44% during A2 (P < 0.05). Rectus femoris activity significantly increased 21% in A2 (P < 0.05). No consistent reductions in anterior translations were evident.

Conclusion. Our preliminary findings, based on a limited number of subjects, indicate joint stability may result from proprioceptive feedback rather than the mechanical stabilising effect of the brace. Despite a significant increase in rectus femoris activity upon landing, only one subject demonstrated an increase in anterior tibial drawer.

Relevance

Studies have shown functional braces do not mechanically stabilise the anterior cruciate ligament deficient knee. Perhaps bracing alters proprioceptive feedback. It has been shown that bracing the anterior cruciate ligament deficient knee may affect hamstring and quadriceps activity. Our findings stresses the importance of functional knee bracing combined with proprioceptive and muscular coordination training in order to increase joint stability.

© 2002 Elsevier Science Ltd. All rights reserved.

Keywords: Anterior cruciate ligament; Knee brace; Knee instability; Electromyography; Proprioception

1. Introduction

Knee braces have been designed to improve functional stability following anterior cruciate ligament (ACL) injury (Vailas et al., 1990; Vailas and Pink, 1993). Braces may be effective in reducing anterior translations when subjected to static or low anterior shear forces but do not protect the knee in situations where high loads are encountered or when the load is applied in an unpredictable manner (Vailas et al., 1990; Vailas and Pink, 1993; Cawley et al., 1991; Cook et al., 1989; DeVita et al., 1992). Ramsey et al. (2001) found that bracing resulted in minor kinematic changes with no consistent reductions in anterior tibial drawer when landing from a jump. During heel-strike and pivoting when the limb transitions from non-weightbearing to weightbearing, Beynnon et al. (accepted for publication) reported
functional braces were incapable of reducing abnormal anterior tibial translations to normal.

Following ACL injury, compensatory muscle recruitment is required in maintaining joint stability (Wojtys and Hutson, 1994; Branch et al., 1989). The hamstrings and quadriceps work synergistically in protecting the joint (Branch et al., 1989). Yet it remains unclear whether knee braces influence electromyographic (EMG) changes in the knee's musculature. It has been shown that bracing the ACL deficient knee evoked changes in EMG activity (Branch et al., 1989; Németh et al., 1997; Acierno et al., 1995). Perhaps the brace acts as a proprioceptive mechanism that influences afferent neural inputs to the central nervous system (CNS) and mediates hamstring and quadriceps activity (Németh et al., 1997). Mechanoreceptors about the joint, muscles and tendon convert peripheral information regarding joint motion, position, and muscle tension into neural impulses. Inputs are transmitted along afferent pathways to the CNS to regulate neuromuscular control (Lepart et al., 2000). One might expect changes in muscle firing patterns, timing or reductions in amplitudes.

Since functional braces do not stabilise the ACL deficient knee during activity, reduced agonist/antagonist muscle activity as a result of bracing may place athletes at greater risk of instability and theoretically increase the risk of joint damage. Therefore, the aim of this investigation was to examine ACL deficiency and the neuromuscular response of functional knee bracing relative to anterior tibial drawer (Ramsey et al., 2001). We hypothesised bracing does not improve stability to the ACL deficient knee but causes a reduction in both quadriceps and hamstring activity by altering limb proprioception. Of interest was the transition from non-weightbearing to weightbearing, or the period from footstrike to approximately peak Fz, because this was associated with a fast anterior displacement of the tibia relative to the femur upon landing (Ramsey et al., 2001).

2. Methods

2.1. Subjects

Four male subjects (mean 21.0, SD 2.4 yrs, 81.8, SD 8.4 kg, 179.8, SD 3.7 cm) with ACL deficient knees and no prior surgical treatment were selected by an orthopedic surgeon. Each subject had a history of significant instability causing them to modify their activity. Deficient knees scored +2 on the Lachman's test and were evaluated with the KT 1000 arthrometer (MEDmetric Corporation, San Diego, USA) and compared against their contralateral leg. The Ethics Committee of the Karolinska Hospital approved the experimental procedure. Participants signed an informed consent form to participate in the study.

3. Motion recordings

The surgery, motion recordings and kinematic analysis have been fully described elsewhere (Ramsey et al., 2001). In brief, Hoffman bone pins (Stryker Howmedica AB, 3 mm diameter, #5038-5-80) were inserted into the femur and tibia. Target clusters consisting of four non-collinear 7 mm reflective markers were then affixed. Six 60 Hz MacReflex infrared cameras (Qualisys, Svedalen, Sweden) were paired and affixed to specially designed tripods. The optoelectric motion capture system was synchronised so cameras in each pair recorded alternate frames, or equivalent to three cameras sampling at 120 Hz. The kinematic data were three-dimensionally reconstructed. Angular and linear data were extracted from coordinate transformation matrices (Lenox and Cuzzi, 1978) that established the spatial relationship between tibial and femoral marker reference frames to the anatomical frames of reference. Rotational and translational data were digitally filtered at 6 Hz (Butterworth fourth order, low-pass, critically damped, zero-lag filter). Joint motion was described according to Grood and Suntay’s joint co-ordinate system (Grood and Suntay, 1983). Accuracy was reported to be less than 0.6° for rotations and less than 0.4 mm for translations for volume of 0.0125 m³ when compared against roentgen-stereophotogrammetric X-rays (RSA) (Lundberg et al., 1992).

3.1. Electromyographic and ground reaction recordings

After surgery the skin was dry shaved and cleaned with alcohol. Preamplified surface electrodes (IK Elektronik, Ellös, Sweden) with a gain of 10 were used to record the EMG activity of the rectus femoris, semitendinosus, biceps femoris, and lateral head of the gastrocnemius. Each consisted of bipolar silver-silver chloride electrodes embedded in a rectangular plastic mount (interelectrode distance of 20 mm center-to-center) measuring 25 × 13 × 7 mm. Electrodes were placed on the subject’s deficient limb over the mid muscle belly parallel to the direction of the muscle fibres. Electrode gel was applied between the skin and electrode to ensure low impedance. To facilitate electrode placement, patients were asked to isometrically contract each muscle. The ground electrode (Blue sensor VL-00-S, Medicostet A/S, Denmark) was placed over the greater trochanter. Since EMG electrodes were applied following surgery, no maximal voluntary isometric contractions (MVIC’s) were collected. It was felt that performing MVIC’s with the bone pins implanted increased the risk of pin bending or muscular complications.

Electrode leads were connected to a junction box fitted onto a belt worn around the subject’s waist. Mass of the unit was 450 g and assumed to have negligible effect on performance. A signal cable connected to the
junction box transferred the four raw EMG channels to a multichannel differential amplifier (IK Elektronik) frequency range DC to 25 kHz, common mode rejection ratio 100 dB, and input impedance greater than 10 GΩ. Amplifier gain was set at 100 and the raw EMG signals were high pass filtered (third order) at 8 Hz and low pass filtered (eighth order Butterworth) at 800 Hz. To avoid the influence of movement artefacts, electrodes and cables were taped to the subject.

Ground reaction forces (GRF) were simultaneously collected with a Kistler force plate (Kistler Instruments AG, Winterthur, Switzerland). The force platform and EMG recordings were synchronised with an external trigger to collect simultaneously with the optoelectric motion capture system. The analogue force and EMG signals were sampled at 960 Hz, A/D converted at 12-bit resolution (SC/ZOOM, Department of Physiology, Umeå University, Sweden) and stored on a dedicated signal analysis computer.

Subjects were tested during a single experimental session. Subjects were randomised to start with either the braced or non-braced condition. During braced trials, the same person fitted the DonJoy Legend knee brace as prescribed by the manufacturer while ensuring no impingement with the pins. Each subject performed a series of one-legged jumps (OLJ) for maximal horizontal distance. Standing with the deficient limb set back, the subject pushed off from their sound limb and landed onto their deficient limb. Their longest measurement was recorded and marked to determine the proper takeoff distance to the force platform. After subjects were given several trials to familiarise themselves with the pins and testing protocol, five trials were recorded for each of the testing conditions.

3.2. Data processing and analysis

For each subject and brace condition, raw EMG and force platform data were processed offline using Bioproc (Version 1.65c), a biomechanics data processing software (School of Human Kinetics, University of Ottawa, Canada). The point when foot-strike occurred was obtained from the force platform data. Initial contact with the force platform was noted to synchronise EMG with GRF data. The cycle commenced 250 ms prior to footstrike and ended 1.75 s following contact, giving a full 2-s window (Fig. 1). Each trial was considered an independent sample within each bracing condition. EMG signals were full wave rectified and low pass filtered with a cut-off frequency of 10 Hz producing an analogue linear envelope. Integrated EMG (iEMG) was calculated for each subject and muscle in three separate time periods: (A1) 250 ms preceding footstrike and two 125 ms intervals following footstrike, denoted by (A2) and (A3). Subject’s mean iEMG was derived for each muscle and time interval in both experimental settings. Each patient served as their own control with analysis focusing on iEMG differences between brace conditions. Subject’s mean iEMG was combined thereby providing an average for each muscle at each of the time intervals and the data were plotted for both bracing conditions.

Of interest were A1 and A2, the non-weighted or anticipatory phase prior to landing and when the foot contacted the force platform (where peak Fy occurred). The instant about footstrike to approximately peak vertical force (A2) was identified since this manoeuvre was associated with a quick anterior displacement of the tibia relative to the femur upon landing (Ramsey et al., 2001). EMG changes as a result of bracing were discussed relative to anterior tibial displacements (Ramsey et al., 2001).

![Fig. 1. A single trial that illustrates the linear envelope (cut-off 10 Hz) and the iEMG for each muscle at the determined time intervals. A1: 250 ms prior to footstrike, A2 and A3 are consecutive 125 ms time intervals following footstrike.](image-url)
GRF data were used as a control between brace conditions. Vertical and posterior shear forces were scaled to body weight (including the brace and junction box when applicable). It was arbitrarily defined that peak vertical forces within 0.5 times bodyweight and peak posterior shear forces less than 0.3 were considered similar (Ramsey et al., 2001). Since the loaded knee can experience forces up to eight times bodyweight, this criterion was considered to result in no mechanical or clinical significance. Impulse, which is the product of force over the time interval with which the force acts, was also calculated. Of interest was the 250 ms time interval following footstrike.

3.3. Statistics

Because of the small sample size, a within subject repeated-measures ANOVA was employed to verify that variance between repeated measures were low. Additionally, we tested for interaction effects and performed a power analysis. Since subjects served as their own control, the Friedman’s ANOVA by rank for $k$ related samples was used to compare within subject measures in both brace settings. The Friedman test is the non-parametric equivalent of a standard repeated-measures analysis of variance applied to ranks rather than raw scores (Seigel and Castellan, 1988; Howell, 1992). The point for using a matched test was to control for experimental variability between subjects, thus increasing the power of the test. For each subject, matching was achieved by comparing the means of repeated measures for each muscle and time interval for both non-braced and braced conditions. If the ranks between conditions were very different, the notion that differences are coincidences of random sampling was rejected. Differences were considered significant when the probability of an $\alpha$ type error was $<0.05$. Emphasis was to highlight significance to this particular group with respect associated anterior tibial translation.

4. Results

No subjects experienced significant discomfort. All reported they could move their knees freely and their ability to jump was unaffected by the pins. Of the four patients, the EMG results of four are reported although the kinematic data from three are presented. The exclusion was the result of one subject bending the femoral pin during knee flexion. Consequently, EMG data with associated anterior tibial translations are available for only three subjects.

4.1. EMG

No interaction effects were identified, observed power was relatively strong and the variance between repeated measures were low. Since subjects served as their own control, the results from the Friedman’s ANOVA by rank for $k$ related samples indicate bracing resulted in significant changes in EMG activity. Semitendinosus significantly activity decreased by 17% during A1, the 250 ms interval or anticipatory phase prior to footstrike ($P < 0.05$). Additionally, biceps femoris significantly decreased by 21% and rectus femoris decreased by 44%.

Fig. 2. Group means and standard deviations for each muscle at each time interval. Significance is indicated by an (+) and a percentage of difference.
decreased 44% during A2, the 125 ms interval following footstrike \( (P < 0.05) \). Conversely, rectus femoris activity significantly increased 21% with the braceduring A2, \( (P < 0.05) \). No changes were observed for the gastrocnemius. Fig. 2 illustrates the group means and standard deviations for each muscle at each time interval. Significance is indicated by an (*) and the percentage of difference. Table 1 shows differences in iEMG for each muscle and time interval between non-braced and braced conditions.

### 4.2. Kinematics

In Table 2, the change in iEMG is presented along with the patient’s corresponding kinematic data for period A2. Three subjects are identified since their EMG and associated kinematic data was available. Negative values indicate a reduction in iEMG and anterior displacement magnitudes when the knee was braced. For subjects B and D, small reductions in anterior displacements were observed as a result of bracing. Conversely anterior displacement increased for subject C.

### 4.3. Ground reaction forces

Table 3 depicts mean peak vertical force \( (F_y) \) and mean peak posterior shear force \( (F_x) \) calculated from each subject’s respective non-braced and braced trials. During non-braced testing, the data recording system failed to store GRF data for subject D. Peak forces varied across subjects since each jumped within their own comfort limits. For subjects C and E, the reductions in peak vertical during bracing exceeded the 0.5 times bodyweight criteria but shear forces were generally consistent between conditions. The amount of impulse generated at landing is listed in Table 3. With the knee supported, no changes in impulse magnitudes were observed for subjects B and C whereas one demonstrated a

### Table 1

Differences of the mean iEMG between the non-braced and braced conditions for each muscle, \( (n = 4) \)

<table>
<thead>
<tr>
<th>Muscles</th>
<th>Time interval A1 250 ms prior to foot-strike, ( \Delta ) in area (mV \cdot s)</th>
<th>Time interval A2 125 ms after foot-strike, ( \Delta ) in area (mV \cdot s)</th>
<th>Time interval A3 second 125 ms interval, ( \Delta ) in area (mV \cdot s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rectus femoris</td>
<td>0.002</td>
<td>0.004(^a)</td>
<td>0.002</td>
</tr>
<tr>
<td>Biceps femoris</td>
<td>0.005</td>
<td>(-0.008(^b)</td>
<td>0.003</td>
</tr>
<tr>
<td>Semitendinosus</td>
<td>(-0.004(^b)</td>
<td>0.002</td>
<td>(-0.003)</td>
</tr>
<tr>
<td>Gastrocnemius</td>
<td>(-0.001)</td>
<td>0.000</td>
<td>0.004</td>
</tr>
</tbody>
</table>

\(^a\)Significant increase in the iEMG between bracing conditions for time interval \( (P < 0.05) \).
\(^b\)Significant decrease in the iEMG between bracing conditions for time interval \( (P < 0.05) \).

### Table 2

The difference in the iEMG between the brace conditions during A2 for each subject and muscle and matched against the change in anterior tibial displacements (mm)

<table>
<thead>
<tr>
<th>Subject</th>
<th>Difference in muscle iEMG</th>
<th>Anterior tibial drawer</th>
<th>Difference drawer (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>R. Fem (mV/s)</td>
<td>B. Fem (mV/s)</td>
<td>Semi (mV/s)</td>
</tr>
<tr>
<td>B</td>
<td>0.007(^a)</td>
<td>(-0.004(^b)</td>
<td>0.002</td>
</tr>
<tr>
<td>C</td>
<td>0.005(^b)</td>
<td>(-0.011(^b)</td>
<td>0.004</td>
</tr>
<tr>
<td>D</td>
<td>0.002(^a)</td>
<td>(-0.001(^b)</td>
<td>(-0.009)</td>
</tr>
</tbody>
</table>

Positive value represents an increase in the area during bracing.
\(^a\)Significant increase in the iEMG between bracing conditions for time interval A2 \( (P < 0.05) \).
\(^b\)Significant decrease in the iEMG between bracing conditions for time interval A2 \( (P < 0.05) \).

### Table 3

Mean peak vertical and mean peak posterior GRF (SD) normalised to bodyweight and mass of the brace across subjects and conditions

<table>
<thead>
<tr>
<th>Subject</th>
<th>Trials</th>
<th>Vertical (Fy)</th>
<th>Impulse for (Fy)</th>
<th>Posterior shear (Fx)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Non-braced</td>
<td>Braced</td>
<td>Non-braced</td>
<td>Braced</td>
</tr>
<tr>
<td>B</td>
<td>( n = 3 )</td>
<td>2.2 (0.3)</td>
<td>2.4 (0.1)</td>
<td>0.36 (0.03)</td>
</tr>
<tr>
<td>C</td>
<td>( n = 5 )</td>
<td>3.5 (0.33)</td>
<td>2.6 (0.6)</td>
<td>0.35 (0.01)</td>
</tr>
<tr>
<td>D</td>
<td>( n = 5 )</td>
<td>n/a</td>
<td>2.9 (0.3)</td>
<td>n/a</td>
</tr>
<tr>
<td>E</td>
<td>( n = 5 )</td>
<td>2.9 (0.6)</td>
<td>2.0 (0.1)</td>
<td>0.46 (0.02)</td>
</tr>
</tbody>
</table>

\( n/a \): Data not available. The impulse for peak vertical force was derived using a 250 ms interval following foot-strike \( (A2 + A3) \). Values expressed relative to bodyweight.
slight reduction. This coupled with peak vertical and peak posterior shear force indicates that jumps onto the force platform were similar between brace conditions.

5. Discussion

Other studies have investigated the relationship between ACL deficiency, functional knee bracing and anterior tibial translations (Ramsey et al., 2001; Beynnon et al., accepted for publication). However this study sought to examine whether the DonJoy Legend knee brace influenced EMG changes in the knee’s musculature and anterior tibial displacements (Ramsey et al., 2001). Our rationale was to reproduce the loading environment in which the subject depends on a brace for protection and to challenge the knee with muscle contraction and bodyweight. This is important because internal loads created by muscle contractions and compressive loading may influence displacements (Beynnon et al., accepted for publication; Wojtys and Hutson, 1994).

Since muscle firing patterns are limb position dependent (Branch et al., 1989), the transition between non-weightbearing and weightbearing was analysed separately to reduce the issue of variance. Three time intervals were analysed: (A1) 250 ms preceding footstrike, (A2) 125 ms following footstrike and (A3) an additional 125 ms interval. Time intervals no less than 125 ms were selected in order to overcome the issue electromechanical delay. The 125 ms time interval (A2) was of primary interest, the period when the foot contacted the force plate and when peak Fy was attained. During this phase, Ramsey et al. (2001) observed rapid anterior tibia displacements relative to the femur until peak Fy. Thereafter the tibia was drawn posteriorly when the limb flexed.

Changes in muscle activity were observed as a result of bracing. Tests of within-subject interaction effects revealed no statistical significance. The Friedman two-way ANOVA by rank for k related samples revealed a significant increase (21%) in rectus femoris activity at footstrike (A2) when the knee was braced. Moreover, significant decreases in both semitendinosus (17%) and biceps femoris (44%) activity were observed during periods A1 and A2 respectively. Reductions in hamstring activity with concomitant increases in quadriceps activity would have the tendency to draw the tibia anteriorly during A2. With respect to the skeletal kinematic data, Ramsey et al. (2001) reported no consistent reductions in anterior tibial translations as a function of the knee brace tested.

Our findings combined with what has been reported suggest increased afferent inputs from knee proprioceptors and the brace-skin-bone interface to the CNS evoke adaptive motor response and modify EMG activity (Németh et al., 1997). During maximal isokinetic knee extension tests, Acierno et al. (1995) observed symptomatic ACL deficient patients increased quadriceps activity by as much as 40% when wearing a brace. Antagonist hamstring activity decreased throughout the range of motion. The authors speculated the increase in quadriceps EMG stabilised brace by exerting a posteriorly directed force to the superior tibia as a result from increased quadriceps tension. Therefore the need for antagonist hamstring intervention was reduced and the brace compensated externally for the absence of the ACL. However, Branch et al. (1989) reported that during the stance phase of side step cutting, ACL deficient subjects without braces significantly reduced quadriceps and gastrocnemius activity, both in iEMG and peak when compared to normal controls. Medial hamstring activity significantly increased. When fitted with a brace, further reductions in total quadriceps and peak EMG were noted as well as a concomitant decrease in total hamstring activity. This suggests bracing the ACL deficient limb required less stabilisation by agonist/antagonist co-contractions (Branch et al., 1989). In a dynamic EMG study of skiers with ACL injury and varying degree of instability, Németh et al. (1997) reported quadriceps, hamstring and gastrocnemius activity all increased without brace, when the knee was maximally flexed. The more unstable knees exhibited greater biceps femoris activity. Bracing the injured leg resulted in increased medial gastrocnemius and semimembranosus EMG just prior to maximal knee flexion. It must be noted that magnitudes will vary dependent upon the methodologies employed and activities involved.

Inter-subject comparisons were not possible since the EMG data were not normalised to peak across the bracing conditions or to MVIC. The major disadvantage lies in the measurement scales. This provides no information on the degree of muscular activation upon landing. With no reference to the subject’s capacity, whether the reference level is 10% or 90% of the subject’s maximum capacity is unknown. Knowledge of the proportion of a subject’s muscle capacity required to perform a task may be important.

Average peak vertical force at foot-strike, peak posterior shear force and impulse were generally consistent between unsupported and braced conditions (Ramsey et al., 2001). The consistency in the kinematic, impulse and ground reaction data are indications that landings onto the force platform were similar. Therefore, changes in iEMG and anterior tibial drawer cannot be attributed to differences in landings but rather to the brace itself. However, due to the invasive nature of our protocol and since subject’s jumped onto their deficient limb, jumps were within the patient’s comfort limits. The knee can experience peak vertical forces of up to eight times bodyweight during high dynamic activity yet patients
only reached about four times bodyweight. This may not have been enough to yield differences between test conditions.

Our findings suggest joint stability may result from proprioceptive feedback rather than the mechanical stabilising effect of the brace. The homogeneous group enables the exclusion of variations such as type of injury and functional status as a cause for differences in values. However, caution is warranted when interpreting our findings because of the small number of subjects and generalisations should be restricted to our procedure. Moreover, the DonJoy Legend brace was solely used in this study and therefore one may not be certain that the results can be applicable to other functional braces. Secondly, application of this methodology may be limited primarily due to its invasiveness. One cannot exclude the fact that local anaesthesia, incisions into the muscle and the introduction of the pins might adversely affect EMG activity. Future studies employing larger sample sizes will be needed to substantiate these results.

6. Conclusion

Joint stability may result from proprioceptive feedback rather than the mechanical stabilising effect of the brace. As a result of bracing, we observed decreased semitendinosus and biceps femoris activity but increased rectus femoris activity. We suggest increased afferent input from knee proprioceptors and brace-skin-bone interface modifies EMG activity. Despite increased quadriceps activity, which would have the tendency to increase anterior tibial translations, two subjects’ demonstrated small reductions. Therefore, our data stresses the importance of functional knee bracing combined with proprioceptive and muscular coordination training in order to increase joint stability.

Acknowledgements

The work was funded in part from the Swedish National Centre for Research in Sports, the Karolinska Institute and the Department of Orthopaedics, Karolinska Hospital. Braces were provided from Smith & Nephew, DonJoy Inc. Carlsbad, California.

References


