Does Wearing a Prophylactic Ankle Brace During Drop Landings Affect Lower Extremity Kinematics and Ground Reaction Forces?

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The objective of the study was to determine if prophylactic ankle bracing worn by females during landings produces abnormal lower extremity mechanics. Angular kinematic and ground reaction force (GRF) data were obtained for 16 athletically experienced females who performed brace and no-brace drop landings. The brace condition displayed reduced in/external rotation and flexion displacements about the ankle and knee joints and increased vertical and mediolateral GRF peak magnitudes and rate of vertical GRF application (paired t test, \( P < .05 \)). The ankle and knee joints landed in a less plantar flexed and more flexed position, respectively. No significant ab/adduction outcomes may have occurred due to interparticipant variability and/or a lack of brace restriction. Conclusion: During typical landings, this lace-up brace increases vertical GRF, decreases ankle and knee joint displacements of flexion and int/external rotation, but minimally affects ab/adduction displacements.

Keywords: ankle joint, knee joint, preventative bracing, impact force

Ankle and knee injuries are the two most common injuries reported for various physical activities worldwide.1 For the foot-ankle complex, sprains are the most prevalent ankle injury reported, whereas for the knee region, several conditions and injuries are commonplace (eg, meniscal lesions, ACL and collateral ligament tears).2,3 Additionally, the consequences of some types of these knee injuries can lead to chronic effects, such as gonarthrosis.4,5

For injury prevention, ankle bracing can be an effective method for preventing future ankle sprains during sport participation.5,7 Hence, it has become common to have athletes wear prophylactic ankle braces during practice/competition, particularly for sports during which sprains occur when landing.6 The benefits of using a brace to prevent ankle injury would be negated, though, if the brace contributed to the occurrence of knee injury or tissue damage. Fortunately, at present, it is encouraging that semirigid bracing has not been found to increase knee injury during basketball and soccer.8,9 However, more research will be needed to confirm that long-term use of preventative ankle bracing is not associated with an increased risk of knee injury.

Therefore, until sufficient epidemiological evidence exists, we can at least determine if ankle bracing affects the knee joint mechanics associated with causation/reduction of knee injury and articular damage. Some researchers have investigated this question for landing tasks and shown that braces limit sagittal plane movements.10–12 Sagittal plane displacements of the lower extremity, particularly after the initial 100 ms of landing, are important for increasing muscle energy absorption and reducing vertical impact forces.13 Therefore, decreased displacements contribute to increased tibio-femoral joint loads.14

However, for frontal and transverse plane mechanics of the knee joint, much less is known. At present, there has been only one study that has investigated brace effects on knee joint kinematics15 and one study of knee joint kinetics.16 Within the Santos et al study,15 only internal/external rotation was examined. Compared with a control condition, greater knee axial rotation occurred during a single-limb axial-twisting task when the Active Ankle brace was worn. Consequently, Santos et al suggested that the brace may increase injury potential. For investigating ankle and knee joint kinetics, Venesky et al16 had participants perform one-foot drop landings onto an inversion-slanted board. Compared with typical landings, when wearing the Active Ankle-T2, ankle eversion torque and knee external rotation torque increased, but the knee ab/adduction torque did not. They concluded it was unlikely
that wearing this brace would place increased stress to soft tissues of the lateral knee region.

Landings are one common movement during which ankle sprains and some types of knee injuries (e.g., ACL injury) occur. Yet, we know little about the effects of brace wear on knee kinematics exhibited in the transverse and frontal planes during landings. Clinically, this may be germane, as increased knee abduction during landings have been associated with ACL injury, particularly for females.19–25

Furthermore, it is not known if ankle braces potentially influence knee injury causation and/or articular damage of the tibio-femoral joint due to vertical ground reaction forces (GRFv). Findings of brace effects on GRFv are equivocal. Increased impact GRFv have been reported by some,10,26 but not other investigators.11,27–29 Greater vertical impact force magnitudes during landings are linked to greater ACL loading24,30–32 and injury.33 Cadaver tibias, when loaded under simulated landing conditions also exhibit tibio-femoral cartilage damage.34,35

Therefore, until a more substantive corpus of epidemiological and prospective research reveals that long-term ankle brace wear has a low knee-injury risk, investigating the mechanical effects of ankle bracing on knee mechanics is needed. Thus, the purpose of the study was to determine the effects of an ASO lace-up brace (Medical Specialties Inc., Charlotte, NC) on the three-dimensional kinematics of the lower extremity and ground reaction forces (GRF). For kinematics, overall, we hypothesized that the mechanical restriction of the foot and tibia would lead to more neutral contact angles and reduced displacements in all planes. For the foot-ankle kinematics, this was anticipated due to the properties of this brace: a lace-up design that is L-shaped to support the foot-ankle complex in a neutral position (Figure 1).

We assumed that, for any plane of motion, as the foot-tibia complex moves from a neutral alignment, increased mechanical resistance would occur. For knee and hip kinematics, predicted outcomes for the sagittal and frontal planes were predicted partly due to tibial restriction affecting the alignment and motions of the femur. When the proximal tibia rotates in either plane, the femur is constrained to rotate, too, due to the joint tissues that act to stabilize the tibio-femoral joint region.

Vertical ground reaction force magnitudes and rates of application were expected to increase during braced landings. The rationale is that reduced sagittal plane displacements about the lower extremity joints are associated with less effective impact force attenuation strategies.13,24,36–38

Methods

Design

For our independent variable, ankle brace, two ankle brace conditions were tested: “no-brace” and “brace” in a repeated-measures, counterbalanced design. For the brace condition, the lace-up brace was worn on each leg.

Participants

Sixteen physically active females (mean ± SD: age = 21.2 ± 2.9 y; mass = 57.9 ± 8.2 kg; height = 164.8 ± 7.6 cm) who had prior and/or current competitive experience in volleyball, basketball, and/or soccer volunteered to participate. An informed-consent form was completed by all participants. Our institutional review board approved this and all forms and protocols. A health-status screening questionnaire was used to assess that participants were healthy and had no symptoms or medical conditions possibly affecting their safety or performance and no musculoskeletal injuries that required medical attention within the past two years.

Instrumentation

Ground reaction force signals of one foot were captured at 1200 samples/s using an AMTI OR-6-6-0 force platform (Advanced Mechanical Technology, Inc., U.S.A.) and low-pass filtered electronically (cut-off frequency = 10.5 Hz. Seven digital video cameras (Vicon MX-40; Vicon Motion Systems Ltd., UK) captured at 240 frames/s the spatial locations (Figure 2) of 55 reflective markers (14 mm) placed on the participant’s body and her own court shoes.39 The malleoli markers were removed for brace trials.

Drop Landing Task

The participant hung from a bar with both hands (height between midpoint of the lateral malleolus to the ground = 0.43 m), then dropped downward and landed with both feet on the ground, with the leg of interest contacting the force platform and the arms still extended above the
head. The participant then remained stationary for 2 s. We selected a level, stationary-landing surface, as that is the most typical landing situation in many sports.

**Procedures**

Anthropometric data were obtained. The landing leg of interest then was determined by observing the limb used by the participant to kick a ball. The participant performed a 5 min warm-up on a stationary bicycle at a self-selected pace. Next, if brace was the condition to be tested, the braces were placed on the participant. The braces were tightened by the participant as instructed and per manufacturer recommendation. For each brace condition, a standing reference trial was captured. Next, two practice trials and 10 test trials of drop landings were performed.

**Data Reduction and Analysis**

For data analyses (via laboratory MATLAB programs), the landing phase began at initial contact with the force platform and ended at maximum knee flexion. Maximum GRF magnitudes were generated for the vertical (GRF_V) and medio-lateral (GRF_M-L) directions. Other GRF variables included the time to maximum GRF_V and maximum rate of GRF_V application that occurred during the initial impact phase (detected from slopes calculated using a central, finite-difference equation). GRF values were scaled to body mass.

For kinematic quantities, 3-dimensional coordinate data were generated from the 2-D camera coordinate data via a proprietary Vicon algorithm and smoothed (GCVSPL). Malleoli marker locations were reconstructed for both landing conditions. Segmental reference coordinate systems for the leg of interest and trunk were generated, and joint center locations relative to the segmental reference coordinate systems were determined for the reference trial. Joint coordinate systems were used to define the Cardan joint angles of the lower extremity joints of the leg of interest. Angles were adjusted to the reference trial angles. The initial contact (θ_IC) and maximum angles were calculated. For angular displacement about a given joint and axis, the greatest difference between two angular positions that occurred during the landing phase was selected (eg, ankle joint dorsiflexion displacement = maximum dorsiflexion angle – initial contact dorsi-/plantar flexion angle). Paired t tests were used to test differences between brace conditions and eta-squared (η^2) used to assess effect sizes. To minimize the number of statistical comparisons, ankle and knee θ_IC and displacements of all joints were tested first to answer specific hypotheses. For a significant displacement, maximum angle (and for hip displacement, hip θ_IC) was then tested. Statistical comparisons of P < .050 were deemed significant. 95% confidence intervals (CI) were generated using difference scores to assess the clinical relevance of the differences between the brace conditions. The difference score for each variable = (brace value) – (no-brace value).

**Figure 2** — Reflective marker set for the lower extremities and pelvis only. The malleoli markers were captured only for the static, standing trial and reconstructed later for the drop landing trials.

**Figure 3** — Means and SD of GRF variables. *Significant difference (P < 0.05) between brace (BR) and no-brace (NO-BR) landings.
Table 1  The 95% confidence intervals and the means of the difference scores for vertical ground reaction force (GRF<sub>v</sub>) variables. A positive or negative value indicates that the magnitude of the brace value was greater or lesser, respectively, than the no-brace value.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Confidence Limits</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lower bound</td>
<td>Upper bound</td>
</tr>
<tr>
<td>Max. GRF&lt;sub&gt;v&lt;/sub&gt; (N·kg&lt;sup&gt;–1&lt;/sup&gt;)</td>
<td>1.1</td>
<td>4.2</td>
</tr>
<tr>
<td>Time to Max. GRF&lt;sub&gt;v&lt;/sub&gt; (ms)</td>
<td>–11.2</td>
<td>–0.4</td>
</tr>
<tr>
<td>Max. Rate of GRF&lt;sub&gt;v&lt;/sub&gt; Applic. (N·kg&lt;sup&gt;–1&lt;/sup&gt;·s&lt;sup&gt;–1&lt;/sup&gt;)</td>
<td>21.4</td>
<td>56.7</td>
</tr>
</tbody>
</table>

### Results

For the GRF outcomes, shown in Figure 3 and presented in Table 1, the maximum GRF<sub>v</sub> magnitude for the brace condition was 3 N·kg<sup>–1</sup> greater (P = .001, η<sup>2</sup> = 0.521) than no-brace landings and occurred 6.4 ms earlier (P = .026, η<sup>2</sup> = 0.290). Hence, the maximum rate of GRF<sub>v</sub> displacement also was greater during the brace condition landings (P = .001, η<sup>2</sup> = 0.541). The brace, compared with no-brace landings, also displayed greater maximum values for GRF<sub>M.L</sub> (P = .029, η<sup>2</sup> = 0.265).

For sagittal plane kinematics (Figure 4), the brace condition compared with no-brace exhibited 10° less displacement for ankle dorsiflexion (P < .001, η<sup>2</sup> = 0.897). In addition, the ankle joint was 8° less plantar flexed (P < .001, η<sup>2</sup> = 0.821) at initial contact and achieved 2° less maximum dorsiflexion angle (P = .010, η<sup>2</sup> = 0.365). The difference-score CI (Table 2) outcomes are congruent with the inferential statistics. The knee flexion displacement was 3° less, as the knee flexion θ<sub>IC</sub> (P = .021, η<sup>2</sup> = 0.305) was 3° more flexed (P = .010, η<sup>2</sup> = 0.363) during braced landings, but the maximum knee flexion angle was the same (mean = 73°). Hip displacement was nonsignificant, so no other hip variables were tested. The knee and hip joint CI for the difference scores (Table 2) also were congruent with the inferential statistics outcomes. One exception was that the difference-score CI for knee joint displacement contained 0° [–5.4°, 0.6°].

Ad/abduction (in/eversion) outcomes (Figure 4) demonstrated no differences for any variables at any of the joints. Differences between brace condition means were 1° or less. Moreover, for difference-score CI (Table 2), all but one variable (ankle joint displacement) ranged from –1.9° to +0.9°. Qualitatively, compared with the majority of participants, for both brace conditions and all joints, two to six participants displayed displacements in the opposite direction (eg, these individuals displayed knee adduction motion, while other participants displayed knee abduction).

For internal/external rotation (Figure 4), the only significant brace condition differences were displacements about the ankle and knee joints. Ankle external rotation and knee internal rotation displacements during brace landings were less (P = .003, η<sup>2</sup> = 0.451 and P = .009, η<sup>2</sup> = 0.374, respectively), and hip external rotation displacement tended to be less (P = .060, η<sup>2</sup> = 0.198). However, of these variables, only ankle external rotation displacement displayed a meaningful difference-score CI [–1.7°, –3.3°]. Interestingly, regardless of brace condition, participants externally rotated about the ankle joint, but at the knee and hip joints, the displacement direction of four to six individuals was opposite to that of the rest of the participants.

### Discussion

The aim of this study was to determine if wearing a lace-up prophylactic brace would alter lower extremity kinematics and GRFs. We had hypothesized that during brace compared with no-brace landings, increased vertical impact forces; and, for all planes of motion and joints, more neutral θ<sub>IC</sub> and reduced displacements. The basic assumption for the kinematic outcomes was that this brace (ASO) design increases mechanical restriction of the foot and tibia during angular displacement from a neutral position.45,46 Thus, this mechanical restriction was expected to directly influence the ankle and knee joint kinematics and indirectly influence the knee and hip joint kinematics via tibio-femoral joint restraint. Increased peak vertical impact force also was anticipated due to reduced lower extremity flexion displacement.

Our anticipated outcomes for the sagittal plane were partially correct. For the θ<sub>IC</sub>, for brace compared with no-brace landings, a more neutral landing position was displayed for the ankle and knee joints. As expected, the brace ankle θ<sub>IC</sub> likely was due, in part, to the L-shaped design that mechanically restrained the foot to a more neutral position. At the knee joint, a 3° more flexed landing position was unanticipated. This may have occurred to orient the foot to avoid a more flat-footed landing. The participant compensated for a more neutral ankle θ<sub>IC</sub> by using tibial flexion about the tibio-femoral joint to adjust the foot orientation.

For sagittal plane displacements, during brace landings, the ankle and knee joint values decreased as expected, but the hip displacement did not change. Less
Figure 4 — Means and SD for kinematic variables. *Significant differences between brace (BR) and no-brace (NO-BR) landings ($P < 0.05$) are shown for the initial contact angle (IC), maximum angle (MAX), and displacement (DISP).
ankle and knee joint displacements were likely influenced more by their respective $\theta_{IC}$ rather than maximum flexion angle. The maximum flexion angles for these joints were nonsignificant; and the difference-score CI (Table 2) supports no difference at the knee joint. Hip displacement probably was not affected by the brace, as 0° difference is contained within the difference-score CIs. Our sagittal plane findings were similar to those of comparable landing studies. For the same ASO brace used in the current study and a semirigid brace, respectively, DiStefano et al and Cordova et al showed the same results as ours. The only contradictory outcome was decreased maximum dorsiflexion angle during brace landing observed by DiStefano et al.

The sagittal plane kinematic outcomes have potential clinical relevance. Landing with an extended knee may be one factor involved in ACL injury. Hewitt et al, though, did not find this association in a prospective study of 205 female athletes. If an extended knee $\theta_{IC}$ is detrimental, then contacting the ground in a more flexed knee position due to brace wear could suggest a positive effect. In contrast, decreased ankle and knee joint displacements may have negative consequences. Reduced lower extremity displacements are less effective landing strategies for absorbing mechanical energy. Santos et al also found the effects of brace wear on knee rotation ambiguous. During static, one-leg standing tasks requiring axial rotation, for brace versus no-brace displacement, knee rotation decreased for ball catching, but increased during trunk twisting.

Frontal plane kinematics were nonsignificant, which we did not anticipate. One possible explanation is that wearing a brace did not influence frontal plane kinematics. Further support for this deduction is that, at all joints, the differences between brace condition means were less than 2°; the difference-score CIs included 0°. The exception was the difference-score CI for ankle eversion displacement [$-1.7°$, $-3.3°$], indicating slightly decreased rotation during brace landings. Therefore, one potential explanation is that the brace did not restrict foot eversion sufficiently enough to influence knee and hip kinematics. During typical landings onto a flat surface, there is little need for eversion restriction, as compared with landing when excessive inversion could occur. Consequently, for the knee and hip joints, if the distal tibial motion was not restricted in the frontal plane, then femoral alignment would not be affected, either. Another potential explanation affecting the knee joint results, suggested by Kernozek and Ragan, is that increased GRF can lead to increased tibio-femoral compressive forces, and consequently, knee abduction/adduction displacement is constrained.

Table 2 The 95% confidence intervals of difference scores for joint kinematic variables (degrees): Lower (LB) and upper (UB) bounds and the mean of the difference score are shown. A positive or negative value indicates that the magnitude of the brace value was greater or lesser, respectively, than the no-brace value.

<table>
<thead>
<tr>
<th>Joint and Variable</th>
<th>Rotation Axis</th>
<th>Flexion / Extension</th>
<th>Ad/Abduction (In/Eversion)</th>
<th>Internal / External Rotation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>LB</td>
<td>UB</td>
<td>Mean</td>
<td>LB</td>
</tr>
<tr>
<td>Hip Displacement</td>
<td>–0.9</td>
<td>3.1</td>
<td>1.1</td>
<td>–1.1</td>
</tr>
<tr>
<td>Knee IC Angle</td>
<td>0.8</td>
<td>5.1</td>
<td>2.9</td>
<td>–0.4</td>
</tr>
<tr>
<td>Max. Angle</td>
<td>–3.8</td>
<td>3.6</td>
<td>–0.1</td>
<td>–0.9</td>
</tr>
<tr>
<td>Displacement</td>
<td>–5.4</td>
<td>0.6</td>
<td>–3.0</td>
<td>–0.8</td>
</tr>
<tr>
<td>Ankle IC Angle</td>
<td>–10.7</td>
<td>–6.3</td>
<td>–8.5</td>
<td>–1.9</td>
</tr>
<tr>
<td>Max. Angle</td>
<td>–3.9</td>
<td>–0.6</td>
<td>–2.2</td>
<td>–1.9</td>
</tr>
<tr>
<td>Displacement</td>
<td>–12.7</td>
<td>–8.7</td>
<td>–10.7</td>
<td>–3.4</td>
</tr>
</tbody>
</table>

The effect of bracing on int/ext rotation kinematics is unclear. The brace had little effect on $\theta_{IC}$. The means between conditions were not different, and the confidence intervals supported a lack of an effect. In contrast, as anticipated, the internal/external rotation displacements at the ankle and knee joints were reduced, with a similar tendency about the hip joint. However, it is uncertain if reduced knee displacement has practical importance. Effect size (.357) was sufficient to be meaningful, but the difference-score CI suggests otherwise [$-2.6°$, $+0.2°$]. Santos et al also found the effects of brace wear on knee rotation ambiguous. During static, one-leg standing tasks requiring axial rotation, for brace versus no-brace displacement, knee rotation decreased for ball catching, but increased during trunk twisting.
Clinically, the lack of a brace effect on knee ab/adduction findings may potentially have beneficial relevance. First, greater abduction $\theta_{AB}$ and ab/adduction displacement are believed to contribute to excessive ACL strain during a noncontact ACL injury, especially for females. In Hewett et al., female athletes who later suffered an ACL injury displayed 8° greater abduction displacement than noninjured athletes. Second, increased frontal plane malalignment is an important determinant in the development of knee osteoarthritis during weight-bearing activity. Also of practical importance for knee ab/adduction displacement is that the brace effect was low (Table 2) regardless of the participant’s displacement direction (abduction or adduction). This is beneficial, as there is much variation among individuals for typical landing mechanics.

For the GRF$_V$ for brace compared with no-brace landings, the increases of 0.31 times body weight of the maximum magnitude, decreased time to maximum GRF$_V$ and 25% greater maximum rate of application confirmed our predictions. We believe that reduced ankle and knee flexion displacements partly account for these findings. Sufficient lower extremity displacement is part of an energy-absorption strategy shown to attenuate impact forces. For brace versus no-brace values, Hodgson et al and Cordova et al observed greater first-peak GRF$_V$ magnitude but no second-peak GRF differences. Other studies have reported no changes for peak GRF$_V$ magnitude. Our outcomes may have varied from these studies due to the use of a different task and/or brace.

The effect of wearing the ASO lace-up brace over a long period of time (eg, years) on joint or bone tissue integrity due to impulsive GRF$_V$ loading is complex and requires further investigation. High magnitudes and repetitions of GRF$_V$ have been implicated in lower extremity overuse injuries and joint tissue degeneration. via a review of literature, however, refuted that high GRF$_V$ magnitudes cause such injuries. Rather, Nigg stressed that high magnitudes and rates of application (within optimal range and conditions) were needed for stimulating bone growth.

Another potential implication of increased GRF$_V$ when using this brace relates to ACL loading. The role of GRF$_V$ on ACL stress and injury is complex and controversial. As GRF$_V$ contributes to tibial axial loading during landings, GRF$_V$ have been suggested to be an important determinant of increased tibio-femoral shear force and ACL forces. Hewett et al reported that ACL-injury athletes had demonstrated a 20% greater peak GRF$_V$ compared with their noninjured cohorts at the start of the prospective study.

In summary, for female athletes, during typical landings, wearing an ASO lace-up brace created greater peak vertical impact force magnitude and rate of application. These outcomes may have potential negative and positive long-term consequences not yet known. For kinematics, greater knee flexion angle at touchdown and less tibial internal/external rotational displacement were displayed, and abduction/adduction angles and displacements were not increased. We conclude that the use of a common, lace-up ankle brace does not appear to adversely affect knee joint kinematics, and potentially may improve lower extremity alignment during landings.

Acknowledgments

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