Ankle Orthoses Effect on Single-Limb Standing Balance in Athletes With Functional Ankle Instability

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Objective: To test whether a rigid or a flexible ankle orthosis affects postural sway in single-limb stance as quantified by stabilometry.

Design: Crossover trial.

Setting: University laboratory.

Participants: Twenty-two athletes with functional ankle instability (consecutive sample of patients with recurrent ankle sprains but without mechanical instability) and 22 healthy athletes (control group of volunteers matched to age, height, weight, physical activity).

Interventions: Stabilometry in single-limb stance on a force platform. Participants were tested on each leg with and without a rigid or a flexible ankle orthosis. The order of test conditions was randomized.

Main Outcome Measures: Sway velocities, sway pattern, and sway area as calculated from center of pressure movements. The two groups were compared by Mann-Whitney test, and the different orthoses within each group were compared by Wilcoxon test, paired samples (type I error 5%, Bonferroni adjustment).

Results: In athletes with functional ankle instability, both a rigid and a flexible ankle orthosis significantly reduced mediolateral sway velocity. A flexible ankle orthosis also changed sway pattern significantly, by reducing the percentage of linear movements of less than 5° per .01 sec.

Conclusions: In athletes with functional ankle instability, ankle orthoses reduce mediolateral sway velocity, possibly because of improved mediolateral proprioception.

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To prevent recurrent ankle sprains, many athletes wear ankle orthoses for training and competition. Recent studies suggest that, apart from mechanically stabilizing the ankle, orthoses have an effect on proprioception.1 3 Stabilometry in single-limb stance on a force platform has proved to be effective for detecting functional ankle instability.4-5 However, the effects of ankle taping and bracing in functional instability have not been detected with stabilometric techniques so far. Tropp and colleagues6 did not find a significant influence of taping on stabilometric values. Twellaar and colleagues7 also did not find any effect of an ankle brace or tape with stabilometry. Only for mechanical instability shortly after acute ligament rupture, Fridén and colleagues8 found some reduction of speed of frontal (mediolateral) sway movements with an ankle brace; however, this was not statistically significant.

Angle reproduction tests by Feuerbach and colleagues9 clearly demonstrate improved proprioception with ankle orthoses. A single-limb stance test in which the patients stood on a soft surface while the examiner counted the number of failures (eg, touching the ground with the opposite leg) also yielded significantly better results with orthoses than without.10 We therefore hypothesized that although orthoses bring about changes in postural control, the methods used so far for analyzing force platform data were not sensitive enough to detect them. Based on a model for postural control by Nashner and McCollum,11 we developed algorithms to quantify sway patterns from force platform data, thereby introducing two stabilometric parameters: (1) Percentage of linear center of pressure (COP) movements of less than 5° directional change per .01 sec; and (2) Percentage of angular COP movements of more than 175° directional change per .01sec.

In addition to the two new parameters, we calculated sway area according to the method of Tropp and colleagues.6 Mediolateral sway velocity, anteroposterior sway velocity, and total sway velocity in the horizontal plane were also calculated. These six stabilometric parameters were used to answer the following questions: (1) Does a rigid or a flexible ankle orthosis change sway velocity, sway area, and sway pattern in athletes with functional ankle instability and healthy athletes? (2) Do athletes with functional ankle instability differ from healthy athletes in sway velocity, sway area, and sway pattern in single limb stance?

MATERIALS AND METHODS

Subjects

We tested 22 athletes with functional instability of the ankle joint and 22 healthy athletes. Functional instability was defined as recurrent ankle sprains (more than 5 per year) and a feeling of "giving way." Patients with increased talar tilt or positive anterior drawer sign were excluded from the study. None of the subjects had a history of major orthopedic or neurologic lesions. We obtained informed consent from all participants. The study was approved by the local ethics committee and was performed according to the declaration of Helsinki (1983).

The patients (10 men, 12 women) had a mean age of 26 years (range, 19 to 36). Twelve of them suffered from unilateral functional instability; the other 10 were bilaterally affected. Despite their ankle instability, our patients participated in 7 training hours per week on average. Most frequent disciplines were volleyball, handball, basketball, and track and field.

The control group (18 men, 4 women) had a mean age of 25 years (range 22 to 31). Their average number of weekly training hours was 7.5 hours per week on average.
hours was 8, and their main disciplines were as listed above for the patients.

Although the patient and control groups had unequal ratios of men to women, the two groups did not differ significantly in height and weight (Mann-Whitney test, two-sided, \( p > .05 \)). Apart from height and weight, we do not expect any gender-related influence on postural control.

Orthoses

Two different types of orthoses were tested: a rigid one (Caligamed, fig 1) and a flexible one (Malleoloc, fig 2). Both are made of thermoplastic material and attached to the leg with velcro straps.

The main part of the Caligamed orthosis is a rigid polyethylene splint. It extends from the distal third of the fibula to the head of the fifth metatarsal. A lateral opening prevents pressure on the malleolus. The lower border of the splint is formed to receive and support the lateral part of the foot, thereby restricting plantar flexion. An additional wedge under the heel forces the foot into slight pronation. With all straps tightly fastened, patients subjectively reported a substantial reduction in inversion and eversion as well as dorsiflexion and plantarflexion.

The Malleoloc orthosis is a flexible polypropylene stirrup. Its lateral part is positioned in front of the lateral malleolus, the plantar one in front of the heel and the medial one behind the medial malleolus. The inner side is cushioned with a thin layer of foam material. The patients subjectively reported a reduction in inversion and eversion, whereas dorsiflexion and plantarflexion seemed to be less restricted.

Data Collection

Stabilometry was performed on a force platform with piezo sensors (Kistler Z15577-00) connected to an 8-channel charge amplifier (Kistler 9865 B1) and a 12-bit A/D converter (CIO-DAS 1602). Data was collected at a sampling rate of 100Hz with the software “BioWare 1.1” on an IBM-compatible PC. The platform was bolted to the floor. Wooden panels were positioned at a horizontal distance of 1.5 cm around the platform to raise the surrounding floor to platform level on an area of approximately 2m by 1.7m. Each trial lasted for 25sec, with the patient or control subject standing upright on one leg, with the foot positioned on the center of the platform (the heel and the space between the second and third toe on the anteroposterior axis, the medial malleolus over the mediolateral axis of the platform). The other leg was flexed in hip and knee, touching neither the platform nor the weight-bearing leg. Arms were folded in front of the chest. Sports shoes were worn to simulate real-life conditions. Each leg was tested without orthosis, with the rigid orthosis, and with the flexible orthosis. Each subject, therefore, underwent a total of six trials. The order of the three test conditions (without orthosis/with rigid orthosis/with flexible orthosis) and the starting leg (right/left) were randomized. Before each trial, subjects were given 3 to 5 minutes to get used to wearing the orthoses.

Data Analysis

Coordinates of the COP on the platform were calculated with “BioWare 1.1” Calculation of sway parameters and statistical
Analysis were performed using "Statgraphics 5.5D" software. 
The x-axis corresponded to anteroposterior sway and the z axis to mediolateral sway. For each trial, 2,500 x z coordinates were obtained. Sway area in form of a confidence ellipse (A) was calculated according to the method of Tropp and colleagues. For further analysis of COP movements, we chose a vectorial approach. Between successive sampling points, the COP moves a certain distance in a certain direction. At a sampling rate of 100Hz, the time for each of these displacements is .01 sec. It was therefore possible to calculate the average velocity of the COP within each sampling interval. This was expressed as a vector with a certain magnitude and direction. From the 2,500 points in each trial, there was a total of 2,499 of these vectors.

For each trial, we calculated three parameters derived from the vectors' magnitude: (1) the average of the mediolateral component magnitude of all vectors (mediolateral sway velocity, Vz [mm/sec]); (2) the average of the anteroposterior component magnitude of all vectors (anteroposterior sway velocity, Vx [mm/sec]); (3) the average of the total horizontal magnitude of all vectors as obtained by vectorial addition of each mediolateral and anteroposterior component (total sway velocity, Vtot [mm/sec]).

Sway patterns are characterized by how frequently and how much the COP changes its direction while moving on the platform. We therefore calculated the amount of directional change between subsequent vectors. A difference of 0° means that a vector points in exactly the same direction as the preceding one. A difference of 180° indicates inversion of the original direction. From the 2,499 vectors in each trial, 2,498 angular differences were calculated. Based on an analysis of histograms of these differences in preliminary studies, we defined two parameters to quantify sway patterns: (1) the percentage of angular differences of less than 3° per .01 sec to the left or right (linear pattern, lin [%]); and (2) the percentage of angular differences of more than 175° per .01 sec to the left or right (angular pattern, ang [%]).

Statistics
For statistical analysis, two groups were defined as the 32 unstable ankles of the patient group (n = 32 legs) and the control group of 22 healthy athletes (n = 44 legs). The 12 anamnestically stable legs of the patient group were not included in our statistics. Previous studies had shown impaired postural control even when patients stood on their unaffected leg. It was therefore impossible to define the legs as either healthy or unstable.

Nonparametric tests were chosen because the sample size did not allow valid assumptions about how the variables are distributed in the population. Patient and control groups were compared with the Mann-Whitney test for unpaired samples (control group without orthosis against patients without orthosis, with the rigid orthosis, or with the flexible orthosis). To determine the effect of orthoses within each group, the Wilcoxon test for paired samples was used ([without orthosis—rigid orthosis), (without orthosis—flexible orthosis), and (rigid orthosis—flexible orthosis)].

Our null hypothesis for each test was that there is no difference in parameter P between test condition A and test condition B. The alternative hypothesis for each test was that there is a difference in parameter P between test condition A and test condition B.

A critical significance level of .05 was chosen. Bonferroni adjustment for multiple testing (18 tests in each group) yields a critical value of \( p = 0.05/18 \). With our calculated \( p \) values rounded to 3 decimals, \( p < 0.003 \) is significant.

Using nonparametric tests and keeping type I error low necessarily limits the power of the study. Limited resources did not allow us to increase our sample size. However, because the emphasis of the study was to demonstrate the effect of ankle orthoses with a modified method of analyzing force platform data, we would prefer to risk an error in saying there is no effect where there might be one rather than claiming an effect that in reality does not exist.

RESULTS

Effect of orthoses on patients' unstable ankles (Wilcoxon test). Wearing either of the orthoses, patients had a significantly lower mediolateral sway velocity than without orthoses (fig 3). The percentage of linear sway pattern was significantly reduced by wearing the flexible ankle orthosis (table 1). The rigid ankle orthosis also reduced the percentage of linear sway pattern, but not significantly. Direct comparison between the two orthoses did not show any significant difference.

Effect of orthoses on the control group (Wilcoxon test). In the control group, neither ankle orthosis had a significant effect on any of the parameters (table 2). There was, however, a tendency to lower mediolateral sway velocity, higher anteroposterior sway velocity, and less linear movements of less than 5° per .01 sec with both orthoses as well as a tendency to more angular movements of more than 175° per .01 sec with the flexible orthosis. There was no significant difference between the rigid and the flexible ankle orthosis.

![Graph](https://via.placeholder.com/150)

Fig 3. Effect of a rigid (■) and a flexible (□) ankle orthosis on mediolateral sway velocity (Vz [mm/s]). In functionally unstable ankles, both orthoses reduce Vz significantly (*p < .003). Rigid and flexible orthoses do not differ significantly (Wilcoxon test, paired samples, with Bonferroni adjustment for multiple testing).
Patients’ unstable ankles compared with the control group (Mann-Whitney test). The differences between the patient and control groups were not significant, irrespective of whether patients wore ankle orthoses or not. Compared with the control group, there was a tendency to higher mediolateral sway velocities in patients without orthosis, as well as a tendency to higher anteroposterior sway velocities in patients with and without orthoses (tables 1 and 2).

DISCUSSION

In our study, a rigid or a flexible ankle orthosis significantly reduced mediolateral sway velocity in patients with functional ankle instability. With the flexible orthosis, linear movements of the COP on the platform were significantly less frequent than without orthosis.

In healthy control subjects, the effect of ankle orthoses on our stabilometric parameters was not significant. When healthy controls wore ankle orthoses, however, there was a tendency to lower mediolateral sway velocity, higher anteroposterior sway velocity, and a less linear sway pattern. With the flexible orthosis, there was also a tendency to more angular movements of the COP.

The direct comparison between patient and control group did not show any significant difference in sway area, sway velocities, and sway pattern. There was, however, a tendency to higher total sway velocity in patients than in the control group.

When interpreting our findings, we have to keep in mind that the results are based on a very conservative statistical approach. By keeping the likelihood of a type I error (rejecting the null hypothesis when it is actually false) at 5% despite multiple testing, we increased the probability of a type II error (accepting the null hypothesis when it is actually false). Our results do not contradict those of Tropp and colleagues, who found a significantly higher sway area in athletes with functional ankle instability than in healthy control subjects. In our study, patients also had higher mean values for sway area and total sway velocity, which might become statistically significant when increasing the study’s power by increasing the sample size.

The new aspect in our study is the effect of ankle orthoses on single limb standing balance as quantified by stabilometry. Despite the conservative statistical approach, the reduction of mediolateral sway velocity in patients with a rigid or a flexible ankle orthosis and the reduction of linear COP movements with the flexible orthosis were significant.

Other groups had previously studied the effect of ankle taping and ankle orthoses with stabilometric techniques. To determine the effect of ankle taping after acute ankle sprain, Tropp and colleagues performed stabilometry with a seesaw interposed between the patient’s foot and the force platform. Despite this modification of test conditions, no difference between taped and untaped ankles could be found with the confidence ellipse.

Friden and colleagues studied the effect of an air-stirrup brace to 3 to 8 days after ankle ligament injury. Unlike our study, all their patients had a positive anterior drawer sign and were thus mechanically unstable. Sway amplitudes and velocities were calculated for movements in the frontal plane only (corresponding to our mediolateral sway velocity). Significant differences were found between healthy subjects and both legs of unilaterally injured patients, whether they wore the brace or not. No difference was found between the injured leg wearing the brace and the uninjured leg without one. Unfortunately, a comparison between the injured side without orthosis and the injured side with orthosis is not mentioned. It is not clear, therefore, whether the brace had a significant effect or not.

Calmels and coworkers tested the effect of an elastic ankle orthosis in healthy volunteers standing on both legs. The difference in anteroposterior movement was significant when orthoses were worn on one or two legs. Patients, however, were not studied and we do not know whether standing on both legs is an effective method for quantifying ankle instabilities.

Table 1: Stabilometric Parameters of Patients’ Unstable Ankles With a Rigid and a Flexible Ankle Orthosis Compared With Standing Without Orthosis

<table>
<thead>
<tr>
<th></th>
<th>Without Orthosis (n = 32)</th>
<th>With Rigid Orthosis (n = 32)</th>
<th>With Flexible Orthosis (n = 32)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mediolateral sway velocity, Vx (mm/sec)</td>
<td>44.7 ± 5.7</td>
<td>43.3 ± 5.1 (p &lt; .003)*</td>
<td>42.5 ± 5.3 (p &lt; .001)*</td>
</tr>
<tr>
<td>Anteroposterior sway velocity, Vx (mm/sec)</td>
<td>62.0 ± 9.3</td>
<td>61.8 ± 8.7 (p = .509)</td>
<td>62.0 ± 9.4 (p = .701)</td>
</tr>
<tr>
<td>Total horizontal sway velocity, Vtot (mm/sec)</td>
<td>83.8 ± 11.2</td>
<td>82.2 ± 10.7 (p = .153)</td>
<td>82.5 ± 11.5 (p = .150)</td>
</tr>
<tr>
<td>Angular movements of more than 175° per .01sec (%)</td>
<td>7.3 ± 2.9</td>
<td>7.7 ± 2.8 (p = .068)</td>
<td>8.1 ± 2.8 (p = .039)</td>
</tr>
<tr>
<td>Linear movements of less than 5° per .01sec (%)</td>
<td>3.2 ± 1.2</td>
<td>2.8 ± 1.0 (p = .021)</td>
<td>2.8 ± 0.8 (p = .002)*</td>
</tr>
<tr>
<td>Sway area, confidence ellipse A (mm²)</td>
<td>136.5 ± 66.4</td>
<td>126.8 ± 69.3 (p = .181)</td>
<td>150.4 ± 64.1 (p = .483)</td>
</tr>
</tbody>
</table>

Values are given as mean ± standard deviation. Significance levels (p) are given for the Wilcoxon test for paired samples comparing stabilometric parameters with each orthosis to the parameters without orthosis (left column).

* Significant (p < .003 after Bonferroni adjustment).

Table 2: Stabilometric Parameters of Control Group’s Healthy Ankles With a Rigid and a Flexible Ankle Orthosis Compared With Standing Without Orthosis

<table>
<thead>
<tr>
<th></th>
<th>Without Orthosis (n = 44)</th>
<th>With Rigid Orthosis (n = 44)</th>
<th>With Flexible Orthosis (n = 44)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mediolateral sway velocity, Vx (mm/sec)</td>
<td>41.8 ± 5.2</td>
<td>40.3 ± 5.9 (p = .013)</td>
<td>40.8 ± 5.6 (p = .031)</td>
</tr>
<tr>
<td>Anteroposterior sway velocity, Vx (mm/sec)</td>
<td>57.2 ± 6.1</td>
<td>58.4 ± 8.7 (p = .007)</td>
<td>58.1 ± 8.7 (p = .015)</td>
</tr>
<tr>
<td>Total horizontal sway velocity, Vtot (mm/sec)</td>
<td>77.8 ± 9.8</td>
<td>77.6 ± 12.2 (p = .791)</td>
<td>77.8 ± 10.5 (p = .945)</td>
</tr>
<tr>
<td>Angular movements of more than 175° per .01sec (%)</td>
<td>6.7 ± 2.2</td>
<td>6.9 ± 2.1 (p = .273)</td>
<td>7.3 ± 2.5 (p = .017)</td>
</tr>
<tr>
<td>Linear movements of less than 5° per .01sec (%)</td>
<td>3.1 ± 0.9</td>
<td>2.8 ± 0.6 (p = .003)</td>
<td>2.9 ± 0.8 (p = .025)</td>
</tr>
<tr>
<td>Sway area, confidence ellipse A (mm²)</td>
<td>121.9 ± 44.8</td>
<td>121.2 ± 43.6 (p = .452)</td>
<td>133.8 ± 50.2 (p &lt; .189)</td>
</tr>
</tbody>
</table>

Values are given as mean ± standard deviation. Significance levels (p) are given for the Wilcoxon test for paired samples comparing stabilometric parameters with each orthosis to the parameters without orthosis (left column). Significance level after Bonferroni adjustment p < .003.
Besides stabilometry, other methods have been applied to determine the effect of ankle orthoses on proprioception and coordination. Feuerbach and associates\(^2\) evaluated ankle proprioception in healthy subjects. They had to match reference joint angles without visual feedback. With an air-stirrup brace, errors in matching the reference positions were significantly less than without the brace. As an explanation for this, increased afferent feedback from cutaneous receptors was discussed.

Jerosch and coworkers\(^3\) examined the effect of a lace-on brace, a stirrup brace, and ankle taping on proprioception in functionally unstable athletes. Angle reproduction, hopping on one leg, and standing on a soft surface were performed. Unlike our study, single-limb stance was not evaluated with a stabilometric technique. A score was calculated by counting the number of failures over a period of time (eg, touching the ground with the opposite leg). Patients performed significantly better with either of the braces in hopping and standing on one leg. Taping, however, decreased the ability to balance on the soft surface. Both braces and taping improved angle-reproduction significantly.

Standing on one leg is a complex coordinative task performed within a set of mechanical constraints. It is therefore difficult to determine which of the orthoses' effects is based on mechanical factors and which is from proprioceptive effects. Nashner and McCollum\(^4\) designed a model for postural sway in the sagittal plane. According to their theory, there are two basic strategies for maintaining single-limb stance: an ankle strategy that makes use of torques around the ankle joint and a hip strategy that generates shear forces. Substantial disturbances of balance are corrected by the hip strategy, whereas fine-tuning is achieved by the ankle strategy. Nashner and McCollum\(^5\) predict that hip strategy prevails if mechanical constraints (like standing on a narrow surface) make ankle strategy ineffective.

Tropp and Odenrick\(^6\) measured sway amplitudes of ankle, hip, and sternum in the frontal plane with an optoelectronic device in healthy subjects standing on one leg on a force platform. They found that the COP is highly correlated to the position of the ankle and showed that the theory of hip and ankle strategy by Nashner and McCollum\(^5\) also applies to mediolateral movements.

In single-limb stance, ankle strategy more efficiently controls anteroposterior than mediolateral sway, simply because the foot is longer than it is wide. The foot's narrow base of support makes it necessary to use hip strategy to control substantial mediolateral disturbances of balance, whereas ankle movements can only achieve fine-tuning of mediolateral sway. The degree of inversion and eversion movements necessary for this fine-tuning are small and, according to our observations, well within the limits set by the orthoses used in our study. A mechanical effect of the orthoses on mediolateral sway is therefore unlikely. Rather than a mechanical effect, the reduction of mediolateral sway velocity might be from improved proprioception. By stimulating cutaneous afferents, ankle orthoses provide additional information about joint position.\(^7\) This could lead to more accurate correctional movements in single-limb stance, thereby reducing mediolateral sway velocity. The fact that the reduction of mediolateral sway velocity in our study is stronger in patients than in healthy subjects (fig 3) supports this theory.

With either orthosis, there was a tendency to less linear movements of the COP on the platform. For patients wearing the flexible orthosis, this effect was statistically significant. With the flexible orthosis, there was also a tendency to more angular movements of the COP. How sway patterns as defined by our parameters reflect the two strategies of postural control\(^8\) cannot be determined with the methods used in our study. The fact that sway velocities are reduced at the same time suggests, however, that the changes in movement patterns brought about by wearing ankle orthoses are useful for maintaining single-limb stance. A simple mechanical model illustrates the connection between sway patterns and velocities: The longer the center of mass is accelerated in one direction, the higher the resulting sway velocity. Frequent, abrupt change in sway direction therefore reduces the chance of reaching high velocities. The higher the sway velocity, the longer it will take a subject with given body mass and muscular force to decelerate and finally change sway direction. If this process takes too long, the center of mass reaches the limits of the supporting surface and balance is at risk. We therefore interpret less linear sway patterns in subjects wearing ankle orthoses as another indicator of improved balance in single-limb stance.

We are aware that our testing conditions hardly reflect real-life conditions. Single-limb stance is much more static and requires a much smaller range of ankle movement than most athletic activities. With a 12.5% reduction in linear COP movements and approximately 5% reduction in mediolateral sway velocity effect size in our study is moderate to small. In what way are these results clinically significant? According to our interpretation, there is evidence of a proprioceptive effect of ankle orthoses. It appears that this effect already exists under relatively static conditions with a small range of movement, we expect it to be present and possibly even stronger in dynamic real-life movements. It is therefore probable that athletes with functional ankle instability might benefit from wearing ankle orthoses.

**CONCLUSIONS**

In athletes with functional ankle instability, single-limb standing balance can be improved by wearing ankle orthoses, the main effect being a reduction in mediolateral sway velocity. A flexible orthosis changes sway pattern by reducing the amount of linear movements of the COP on the supporting surface. Our findings suggest that, in addition to mechanically stabilizing the ankle, orthoses appear to have a measurable effect on proprioception in the mediolateral direction.

**References**


**Suppliers**

b. Bauerfeind GmbH & Co., Postfach 10 03 20, D-47880 Kempen, Germany.
c. Kistler Instruments AG, Eulachstr. 22, CH-8408 Winterthur, Switzerland.
d. Keithley Instruments GmbH, Landsberger Str. 65, D-82110 Germering, Germany.
e. Manugistics Group, Inc., 2115 E. Jefferson St., Rockville, MD 20852-4999.