Effects of Rigid and Dynamic Ankle-Foot Orthoses on Normal Gait

Bastien Guillebastre, MS; Paul Calmels, MD; Patrice Rougier, PhD
Le Bourget-du-Lac, France

ABSTRACT

Background: As shown through posturographic data, wearing an ankle-foot orthosis (AFO) causes a backward shift in healthy subjects of the mean position of the center of pressure under the limb wearing it, and difficulty in controlling these displacements. This study evaluated whether this particular positioning influenced gait independent of a neurological disorder.

Materials and Methods: Two AFO models, with different mechanical concepts (a rigid-AFO (R-AFO) and dynamic-AFO (D-AFO)), were worn by 11 healthy subjects required to walk on a 12-m electronic mat. Velocity, step time and step length were assessed for each of the five conditions where subjects walked barefoot, and wearing R-AFO or D-AFO (without and with slight and greater stiffness at the elastic band). Spatial and temporal characteristics of each support were also analyzed.

Results: Although wearing R-AFO disturbed velocity, step length and time with an asymmetry between sides, wearing the D-AFO only affected a support characteristic (midline length: length between the pivot points of the two dimensional sensor structure of heel and toe area). No effect was seen when modifying the stiffness of the D-AFO model.

Conclusion: Even though the posturographic data might partly explain this behavior, wearing an orthosis caused different effects on normal gait parameters. Clinical Relevance: These features should be useful when prescribing an ankle-foot orthosis by differentiating what alterations are due to the orthosis and which are due to the gait disorder.

Key Words: Ankle-foot Orthosis; Gait; Self-selected Speed; Healthy Subjects; Electronic Mat

INTRODUCTION

Ankle-foot orthoses (AFO) are most commonly prescribed for neurologic patients to ensure toe clearance during the swing phase of gait, to absorb the body-weight impact at the heel strike, and to support forward propulsion of the body during the mid- to late-stance phase. One of the most important functions of the AFO is to resist abnormal movements of the ankle joint to compensate for muscular deficiencies.

Limitations of many commercially available AFO’s (rigid-AFO (R-AFO)) upon gait control have been emphasized in several studies. Wearing an AFO is, in theory, accompanied by both increased plantarflexion and dorsiflexion resistance. Although the former is desirable, the latter is not. The double function of AFO plantarflexion resistance is to reduce the knee flexion moment by ensuring smooth plantarflexion, and to prevent quick plantarflexion because of insufficient force generation by the dorsiflexors during the initial stance phase. An excessively stiff ankle plantarflexion moment at the time of heel strike, due to excessive R-AFO stiffness, may render the knee unstable. For normal gait, dorsiflexion resistance needs to be reduced as much as possible. This feature is frequently seen in hemiplegic gait patterns and is caused by the triceps surae inability to relax. This excessive resistance to dorsiflexion limits the passive dorsiflexion movement and thus impairs the forward shift of the center of gravity when the foot is on the ground.

Since several parameters (magnitude of plantarflexion resistance and the initial ankle angle) may fluctuate depending on the patient’s gender, the degree of disability, and muscle strength, the AFO needs to be adapted to the particular patient’s requirements. Interestingly, Yamamoto et al. presented data obtained for a new AFO allowing an adjustable plantarflexion resistance and, concomitantly, preventing any dorsiflexion resistance. However, by including only pathological subjects, this study was unable to differentiate what abnormalities arose from the sensorimotor impairments versus the physical features of this experimental AFO. To better assess the function of the AFO, it is important to focus on its physical characteristics. One should keep in...
mind that these devices, even though helpful for hemiparetic patients might affect gait in healthy subjects.\(^1,2\)

On the basis of the above-mentioned desirable characteristics, a new dynamic-AFO (D-AFO) (Figure 1) was proposed. The main innovation of this D-AFO was the adjustable stiffness of the elastic band varying the resistance to plantarflexion. A recent study\(^6\) demonstrated that wearing this dynamic-AFO during stance induces a backward shift of the mean of Center Of Pressure (COP) position, difficulty controlling its movements when the elastic band was stiffened, and was comparable to those observed with standard rigid AFO models.\(^5\)

Based on previous data\(^5,6\) we hypothesized that the gait of healthy individuals wearing orthotic devices would be affected in various ways. Indeed, the backward shift of the COP under the foot wearing the orthosis would likely impair the push-off of the grounded foot. Consequently, a unilateral effect, involving spatiotemporal step characteristics, should be observed. Moreover, the difficulty the foot with the orthosis has controlling the COP displacements during stance should induce an enhanced involvement of the foot on the opposite side.

To the best of our knowledge, only two studies have assessed the effects of an AFO on normal gait. Opara et al.\(^11\) showed that wearing an AFO on the right side caused significantly decreased left stride and right step lengths. However, since the subjects wore their own shoes and the walking cadence was imposed can be considered as possible confounding variables. On the other hand, in healthy subjects at free-walking speed wearing their own comfortable tennis shoes, Balmaseda et al.\(^1\) showed a significant reduction in the mean duration of the stance phase with a plastic AFO and a posterior shift of the point of impact at heel strike. Even though the parameters studied describe what happens under each support, no information is given on gait performance due to spatiotemporal parameters. Thus, in spite of comparable populations and tasks, these studies\(^1,12\) do not allow us to predict the gait behaviors, through spatiotemporal parameters, induced by the rigid and dynamic AFOs at free walking speed.

Thus we evaluated whether rigid and dynamic AFOs could induce comparable gait alterations independent of neurological deficit. We also assessed whether the modulated stiffness of the elastic band changed the gait patterns.

**MATERIALS AND METHODS**

**Subjects**

Eleven volunteer subjects (five females and six males) (median age, 25.8 years - range, 19 to 37; median height, 171.6 cm - range, 154 to 179; median mass, 67.2 kg - range, 51 to 83) were recruited for the study. All participants were free of neurological and orthopedic disorders, gait impairments and provided informed consent.

**Orthoses**

Two models of orthoses with different mechanical characteristics were compared on normal gait behavior. The rigid ankle-foot orthosis (R-AFO) was derived from the Houston orthosis and was manufactured with polypropylene maintaining the foot in a neutral position (i.e. with the ankle at 90 degrees) (Figure 1).

The dynamic ankle-foot orthosis (D-AFO) (Figure 1) had three straps (one around the shank and one around the foot with an elastic band joining these two segments). This device allows the magnitude of resistance to be easily changed by adjusting the length of the elastic band using a Velcro\(^\text{®}\) system.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{fig1.png}
\caption{Rigid (A) and dynamic (B) ankle-foot orthoses.}
\end{figure}
Gait analysis

Each subject participated in five randomly performed experimental conditions: the reference condition consisting in walking without the orthosis (REF), one with the rigid-AFO (R-AFO) and three with the dynamic-AFO in different configurations (without elastic (D-AFOo), with slight (D-AFO-), and greater stiffness (D-AFO+)). This stiffness was defined based on visual markers on the elastic band. The markers were defined to obtain strain levels corresponding to 30% (in the D-AFO- condition) and 70% (in the D-AFO+ condition) of the maximal strain (100%). AFOs were worn unilaterally in this experiment. Since a right leg dominance was determined for all subjects, they were worn on this side. The orthoses were chosen based on the subjects' foot size, and were worn without shoes, but with an elastic bandage to ensure a tight fit between R-AFO and the lower limb.

Subjects were required to walk straight at a self-selected speed over a 12-m distance. To avoid the effects of acceleration and deceleration, measurements were only taken from the median 1.32 m (length of the active GAITRite@ area). The self-selected or comfortable speed has been found to be reproducible.2,7 All participants practiced walking a few steps for all conditions before the measurements were taken. Data were collected for all five conditions through two successive trials. Four minutes of rest were given between each condition and 30s between each trial.

Gait was assessed using an electronic mat (GAITRite, CIR Systems, Clifton, PA). The GAITRite® is an 8.3-m x 0.89-m carpet that forms an electronic walkway. Pressure sensors are embedded into the carpet in a horizontal grid. As the subject walks over the carpet, the sensors close under pressure, enabling data collection on spatial and temporal gait parameters. The active area of the mat is 7.32 m long and 0.61 m wide. The sensors were 12.7 mm apart. The GAITRite® application software was used to process the data and calculate spatial and temporal gait parameters. Moreover, the software uses special algorithms to automatically group sensors and form footprints. Once a footprint has been formed, it is divided into heel, midfoot and toe areas. The sampling rate of the system is 80 Hz.

Several studies have demonstrated the validity and reliability3,4,13 and reproducibility10,14 of the GAITRite® system measurements.

Data analysis

The spatiotemporal components included velocity (cm/s), step time (s) and length (cm) (Figure 2). Since orthoses were worn unilaterally, spatial and temporal characteristics of each support were also analyzed. The stance phase of each foot (percentage of the gait cycle time elapsed between the heel contact and the toe off of the same foot), the percentage of time during which the rear third of the foot was on the ground (heel off-on, %), and the midline length defined as the length between the pivot points of the two-dimensional sensor structure of heel and toe area (midline length, cm) were evaluated (Figure 2).

A non-significant normality distribution for all parameters studied was found through Kolmogorov-Smirnov tests. Thus, the statistical analysis consisted of a nonparametric one-way ANOVA (Friedman) with repeated measures, followed when necessary by Dunn’s multiple comparison tests. The asymmetry between right and left sides for each condition was assessed using nonparametric Wilcoxon tests. For all tests, significance was set at p < 0.05.

RESULTS

Effects of orthoses on comfortable velocity

The statistical analysis showed that the orthoses had a significant effect for percentage of the left stance phase was shown (H(1, 10) = 9.6947; p < 0.05). More precisely, compared with other conditions, velocity decreased when the subjects were wearing the R-AFO. However, only a significant difference was found between R-AFO and D-AFOo (p < 0.05) (Figure 3).

Spatial and temporal characteristics of steps

In comparison with the other conditions, left step characteristics were disturbed when subjects stood on the right foot wearing the R-AFO. Indeed, statistical analysis showed a significant effect of the orthoses on left step...
length \( (H(11,10) = 0.0084; p < 0.05) \) and time \( (H(11,10) = 0.0008; p < 0.05) \) (Table 1). Post tests revealed that step length significantly decreased \( (p < 0.05) \) when comparing barefoot walking (REF) and the R-AFO. Moreover, step time increased using the R-AFO \( (p < 0.05 \) compared with D-AFO, D-AFO- and D-AFO+) (Table 1).

When subjects walked barefoot (REF), no asymmetry between right and left step spatiotemporal characteristics was found for both length \( (T:21.5, p > 0.05) \) and time \( (T:21.5, p > 0.05) \) (Table 1). Only one experimental condition induced asymmetry between right and left steps characteristics. When the R AFO was worn, results showed step time \( (T:5, p < 0.05) \) and length \( (T:7, p < 0.05) \) asymmetries (Table 1).

**Spatial and temporal characteristics of supports**

A significant effect for percentage of the left single support was shown \( (H(11,10) = 0.0064; p < 0.05) \), i.e., post hoc tests yielded a significant increase between R-AFO and REF conditions \( (p < 0.05) \) (Table 1). On the other hand, the statistical analysis indicated a significant effect of the percentage of time during which the rear third of the left foot (the one without orthosis) was on the ground (Heel Off-On, %) \( (H(11,10) = 0.0012; p < 0.05) \). As illustrated in Table 1, post hoc tests revealed that the left heel off-on percentage significantly decreased when subjects wore the R-AFO on the right side, compared to REF \( (p < 0.05) \) and D-AFO0 \( (p < 0.05) \) conditions.

Finally, the statistical analysis showed a significant effect for the right midline lengths across experimental conditions \( (H(11,10) = 0.0023; p < 0.05) \). More precisely, the D-AFO+ condition was associated with a significantly longer midline lengths than when walking barefoot (REF) \( (p < 0.05) \) or in the D-AFO0 condition \( (p < 0.05) \) (Table 1).

**Differences in asymmetry between sides for each condition**

No asymmetry between right and left stance phase percentages or midline lengths was found for any condition (Table 1). Only an asymmetry of the heel off-on percentage between the right and left supports \( (T = 8; p < 0.05) \) was shown in the R-AFO condition (Table 1).

**DISCUSSION**

The purpose of this study was to evaluate the effects of two models of AFOs (a rigid one and a dynamic one that allowed mechanical adjustments), on gait function, at self-selected speed, in healthy subjects. The lack of concordance between previous results (decrease in step length of the foot wearing the orthosis)\(^2\) and our findings (decrease in step length of the foot without the orthosis) can be mainly explained by experimental differences (shoes worn, imposed walking cadence, type of AFO). Although the experimental protocol of Balmaseda et al.\(^1\) was similar, measured gait parameters were different, hence partially excluding any possibility of comparison. Regarding posturographic data,\(^5,6\) we hypothesized that the backward shift of the COP under the limb wearing the orthosis along with the difficulty controlling these displacements should induce a disturbance.

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**Table 1: Spatiotemporal parameters (Mean ± SD)**

<table>
<thead>
<tr>
<th></th>
<th>REF</th>
<th>R-AFO</th>
<th>D-AFO(^{-})</th>
<th>D-AFO(^{0})</th>
<th>D-AFO(^{+})</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Step length (cm)</strong></td>
<td>Left 68.88 ± 6.26</td>
<td>65.77 ± 6.46(^1)</td>
<td>68.30 ± 6.16</td>
<td>67.76 ± 6.08</td>
<td>67.90 ± 6.29</td>
</tr>
<tr>
<td></td>
<td>Right 68.77 ± 5.05</td>
<td>68.04 ± 5.79</td>
<td>68.90 ± 5.34</td>
<td>68.12 ± 5.80</td>
<td>68.41 ± 5.65</td>
</tr>
<tr>
<td><strong>Step time (s)</strong></td>
<td>Left 0.53 ± 0.02</td>
<td>0.56 ± 0.03,3,4,5</td>
<td>0.53 ± 0.02</td>
<td>0.53 ± 0.02</td>
<td>0.53 ± 0.02</td>
</tr>
<tr>
<td></td>
<td>Right 0.53 ± 0.03</td>
<td>0.53 ± 0.02</td>
<td>0.53 ± 0.03</td>
<td>0.53 ± 0.02</td>
<td>0.53 ± 0.02</td>
</tr>
<tr>
<td><strong>Stance phase (%)</strong></td>
<td>Left 60.88 ± 1.70</td>
<td>62.02 ± 1.05(^1)</td>
<td>61.17 ± 1.28</td>
<td>61.69 ± 1.31</td>
<td>61.62 ± 0.91</td>
</tr>
<tr>
<td></td>
<td>Right 61.36 ± 1.11</td>
<td>61.50 ± 1.58</td>
<td>60.95 ± 1.23</td>
<td>61.32 ± 1.21</td>
<td>61.45 ± 1.45</td>
</tr>
<tr>
<td><strong>Heel Off-On (%)</strong></td>
<td>Left 14.76 ± 3.50</td>
<td>11.54 ± 3.60(^1,3)</td>
<td>14.84 ± 4.01</td>
<td>13.66 ± 3.74</td>
<td>13.99 ± 4.15</td>
</tr>
<tr>
<td></td>
<td>Right 15.36 ± 5.54</td>
<td>12.84 ± 3.55</td>
<td>15.42 ± 5.52</td>
<td>13.44 ± 5.20</td>
<td>14.64 ± 4.44</td>
</tr>
<tr>
<td><strong>Midline length (cm)</strong></td>
<td>Left 13.05 ± 0.85</td>
<td>13.08 ± 0.80</td>
<td>13.10 ± 0.82</td>
<td>13.09 ± 0.82</td>
<td>13.05 ± 0.83</td>
</tr>
<tr>
<td></td>
<td>Right 13.01 ± 0.84</td>
<td>13.18 ± 0.85</td>
<td>13.01 ± 0.86</td>
<td>13.18 ± 0.85</td>
<td>13.26 ± 0.89(^1,3)</td>
</tr>
</tbody>
</table>

In the five conditions, without orthosis (REF), with the rigid orthosis (R-AFO) and the dynamic orthosis (D-AFO0, D-AFO-, D-AFO+), the device was worn on the right foot.

\(^1\): Indicates statistically significant difference between left and right sides scores \( (p < 0.05) \).

\(^2\): Indicates statistically significant difference with the REF condition \( (p < 0.05) \).

\(^3\): Indicates statistically significant difference with the D-AFO0 condition \( (p < 0.05) \).

\(^4\): Indicates statistically significant difference with the D-AFO- condition \( (p < 0.05) \).

\(^5\): Indicates statistically significant difference with the D-AFO+ condition \( (p < 0.05) \).
of the spatiotemporal step characteristics and a preferred involvement of the opposite foot, respectively.

Although the self-selected speed results complied with normal values for all conditions, data in the R-AFO condition gave lower values. This is in accordance with the study conducted by Holt et al.,7 which showed that to preserve an economic gait, healthy subjects reduced their walking speed. Wearing the R-AFO thus disturbs locomotion performance. In addition to an overall disturbance, wearing the R-AFO induced changes in step spatial and temporal characteristics, i.e., with this rigid orthosis, asymmetry between stepping performance occurred since step spatial and temporal characteristics of the foot without the orthosis were smaller and longer than in the REF and D-AFO conditions, respectively. The fact that the R-AFO prevents dorsiflexion at push-off could explain this change. In addition, the backward shift of the mean position during undisturbed stance of the foot wearing the R-AFO6 seems to disturb gait through a decrease in the opposite step length, confirming our first hypothesis. This backward shift must therefore be viewed as an index expressing a decreased capacity to use plantarflexion and thus a lessened capacity to accelerate the center of gravity during gait.

Furthermore, for the barefoot condition, wearing the R-AFO increases the percentage of opposite side stance time. The foot without the orthosis appears to be preferably exploited to ensure an efficient gait. Interestingly, this finding could be considered as a compensatory phenomenon related to the difficulty of controlling the support with the orthosis in undisturbed stance5 and thus would confirm our second hypothesis.

When subjects wear the R-AFO, a single support phase pattern of the opposite lower limb is disturbed and leads to an asymmetry. The percentage of time during which the rear third of the foot without the orthosis was on the ground (heel off-on) was significantly decreased with respect to the REF and D-AFO6 conditions. Considering an equal ankle angle at heel strike, the disturbance of heel off-on percentage time of the foot without the orthosis could be explained through the decrease in step length, inducing a shorter braking action time for the foot coming down.

The unilateral disturbance observed when subjects wore the R-AFO provides information on the relation with walking speed change. It seems that walking speed change was determined by unilateral disturbances.

Interestingly, the D-AFO only affects midline lengths of the foot wearing it. In the D-AFO+ condition, longer midline lengths under the foot wearing the orthosis were noted with respect to REF and D-AFO6 conditions. However, this effect doesn’t induce any asymmetry between the midline lengths of each support. Thus, the main factor of the dynamic-AFO influencing this gait change is the stiffness of the elastic band. To maintain an efficient use of the foot on the orthosis side, subjects had to exploit the maximal distance that could be in contact with the ground, inducing longer midline lengths.

This phenomenon could be associated with the problems controlling displacement when subjects stood quietly wearing this orthosis with the same adjustment.6

It is worth noting that there was no difference between D-AFO- and D-AFO+ conditions. This feature could be explained by a low-level effect of the elastic band during gait due to low plantarflexion. Indeed, only 30% of normal gait cycle time is performed with ankle plantarflexion.11 With patients, one can expect significant effects since hemiparetics have been shown to spend 90% of the gait cycle in plantarflexion.11 Lastly, this lack of difference between D-AFO- and D-AFO+ conditions may also be due to an insufficient strain of the elastic band to affect normal gait parameters in the D-AFO- condition, or to an efficient compensatory strategy in the D-AFO+ condition.

In addition to structural and material differences, each R-AFO or D-AFO model induced particular and different effects on normal gait parameters. Although these devices have similar functions, the effects induced by each model on gait appear complementary. While R-AFO influences walking speed, step time and length data with an asymmetry between sides, D-AFO affects only midline lengths, without any impact upon the step spatiotemporal characteristics. A neuromuscular adaptation with D-AFO characteristics could explain this difference.

CONCLUSION

Compared with barefoot walking, the use of an R-AFO unilaterally causes changes in walking speed, step (length and time) and support (percentage of stance and heel off-on time) characteristics, with asymmetry between left and right sides (except the stance stage percentage) in healthy subjects. Wearing a D-AFO orthosis induces changes in midline lengths, especially when this is accompanied by a high level of stiffness. On the contrary, with a low level of stiffness, normal gait is seen. Our results might be helpful for better understanding the influence of a biomechanical disturbance of the AFO independent of neurological disorders. Moreover, relating undisturbed stance and gait provide complementary insights and thus facilitate our understanding of the results. The effects have provided new insights about the functional use of these orthotic devices in healthy subjects. Naturally, this kind of investigation should be extended to include impaired individuals. Thus, further research is needed to reveal the effects of different types of AFOs and the effects of adjusting the mechanical characteristics for patients with footdrop on standing and gait performance. However, the expected gait behaviors for these patients should largely differ from our results, due to their specific needs. Indeed, what can be disturbing for healthy subjects could be helpful for footdrop patients.
ACKNOWLEDGMENT

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REFERENCES