Stiffness of the Arizona Ankle-Foot Orthosis Before and After Modification for Gait Analysis

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ABSTRACT

The purpose of this project was to measure the stiffness of the Arizona ankle-foot orthosis (AFO) before and after it was modified for gait analysis with a foot specific marker set. Five Arizona AFOs were tested in the intact condition in a custom testing device in the sagittal and coronal planes. After testing in the intact condition, 2.5-cm diameter holes were drilled in the malleoli and the medial, lateral, and posterior aspects of the calcaneus. Three cycles of loading in each plane were averaged for analysis. The slope of the load-displacement curve was calculated to determine the brace stiffness. The coefficient of repeatability was ± 0.17 Nm per degree in plantarflexion, ± 0.16 Nm per degree in dorsiflexion, ± 0.38 Nm per degree in inversion, and ± 0.18 Nm per degree in eversion. The stiffness decreased significantly in plantarflexion and dorsiflexion but not in the coronal plane motions. The change in stiffness in plantarflexion increased with an increase in the height of the medial malleolus holes. This relationship was significant, as determined by the Hotellings t test ($p = 0.04$), which suggests that the Arizona AFO should be reinforced on the medial side before it can be used in gait analysis studies. (*J Prosthet Orthot.* 2009;21:204–207.)

KEY INDEXING TERMS: Arizona AFO, stiffness, biomechanics, gait analysis

Posterior tibial tendon dysfunction (PTTD) is the most common cause of adult acquired flatfoot deformity. The majority of articles on PTTD focus on describing methods for operative treatment of this disorder. There are few studies quantifying the effects of nonoperative treatment. Nonoperative treatments for PTTD include ready-made orthotics and braces or custom-made orthotics, including the Arizona ankle-foot orthosis (AFO), articulated AFOs, and the University of California Biomechanics Laboratory (UCBL) orthosis. The effects of the Arizona AFO were evaluated in patients with PTTD using three clinical measures, the American Orthopaedic Foot and Ankle Society hindfoot score, the Foot Function index, and the SF-36.[1] Significant improvements were seen in all three indices after treatment. Additionally, patients who were dependent on pain medication decreased or eliminated the use of medication for pain. Alvarez et al.[2] evaluated the use of short-articulated AFO with an instep wrap and full-length toe plate and a three-quarter length thermoplastic elastomer foot orthosis with high medial and lateral trim lines in combination with an aggressive therapy program in 47 patients. After treatment, foot and ankle pain decreased significantly, 83% of patients could perform a single heel rise with no pain, and there was a significant increase in eccentric and concentric ankle strength.

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The effects of nonoperative treatment have been quantified in vitro by examining changes in midfoot and hindfoot kinematics and plantar pressures before and after creating a flatfoot deformity. The Arizona AFO, the UCBL, a molded AFO, and various off-the-shelf ankle braces were tested.[3] The off-the-shelf braces had little effect on the restoration of kinematics, the UCBL orthosis partially restored kinematics at the arch and hindfoot and the Arizona AFO restored midfoot height. Although this study provided insight into the effects of different nonoperative treatment on the hindfoot and midfoot kinematics, the tests were completed with static loading during one portion of the gait cycle. To properly evaluate the effects of nonoperative treatment on PTTD, it is necessary to perform a quantitative gait analysis in vivo.

Holes must be placed in the orthotic to measure ankle kinematics, using a lower limb marker set, such as the Helen Hayes marker set. These modifications may include holes over the malleoli and calcaneus.[4–6] Therefore, braces, such as the Arizona AFO (Arizona AFO, Inc., Mesa, AZ), must be modified to allow for placement of the markers. The purpose of this study was 1) to determine the stiffness of the Arizona AFO in the sagittal and coronal planes before and after it is modified for gait analysis and 2) to determine whether the brace should be modified to function as designed.

METHODS

Five custom Arizona AFOs, casted from patients with PTTD, were tested (two right and three left). A solid ankle cushion heel (SACH) foot was inserted into the brace, a 1-in spacer was placed on top of the foot, and plaster was used to fill in the remaining portion of the brace (Figure 1). A 4-cm diameter plastic rod was inserted into the plaster before it hardened, to attach the brace to the testing device. Before testing, the plastic spacer was removed to allow for free motion where the ankle joint was located.

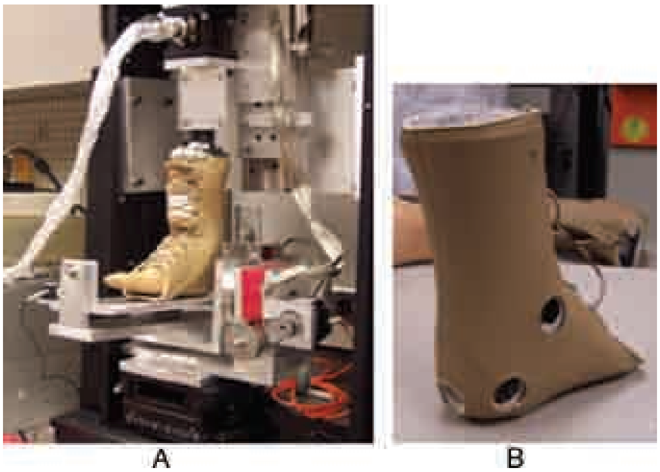

Figure 1. A, Experimental setup for Arizona AFO stiffness testing. B, Arizona AFO modified for gait analysis.

A custom testing device was built to automatically manipulate the brace in the sagittal and coronal planes. A two-axis gimbal system was constructed to make a tilting foot plate on which the foot construct was attached. The method incorporated a six-component load cell (JR3, Inc., Woodland, CA) to measure the torques and forces between the footplate and the fixed tibial component. The gimbal system was made with two worm gear mechanisms (Rino Mechanical Components, Inc., Freeport, NY) powered by stepper motors (Parker Hannifin Corporation, Compumotor Division, Rohnert Park, CA). The angular displacement of the footplate relative to the tibia axis was measured with a tilt sensor (CXTLA02, Crossbow Technology, Inc., San Jose, CA). Control of the gimbal and data collection was performed with a National Instruments motor control and A/D hardware in conjunction with a custom program written in Labview software (National Instruments, Austin, TX).

The brace was positioned such that it was aligned with the x and y axes of the load cell. The braces were rotated at 0.5°/sec, and data were collected at 30 Hz. Each brace was tested two times in the intact condition. Between each trial, the brace was removed from the testing device, and the SACH foot and plaster mold were removed from the brace. After the intact brace was tested, 2.5-cm diameter holes were cut in each brace to allow for the placement of reflective markers on the medial and lateral malleoli and on the medial, lateral, and posterior aspects of the calcaneus.

DATA ANALYSIS

Each brace was tested for 6 cycles. Cycles 1–3 were preconditioning and cycles 4–6 were used to calculate the stiffness of the brace. The data were truncated to eliminate the toe region and to analyze each brace in a consistent linear region. The slope of the load-displacement curve was calculated with a custom program written in Matlab (The Mathworks, Natick, MA). The slope of the load-displacement curve defined the stiffness of the brace. The reported stiffness was the average of three cycles. The sagittal plane data were analyzed from 3° to 10° of plantarflexion and from 1° to 9° of dorsiflexion. Inversion cycles were truncated from 2° to 5°. A 3° range of motion was selected in the eversion cycles because it was not possible to select a consistent region for all five braces.

STATISTICAL ANALYSIS

Measurement repeatability was assessed using the coefficient of repeatability.[7,8] The difference in the brace stiffness after holes were inserted was assessed with Hotelling's t-squared multivariate test (StatView, SAS Institute, Inc., Cary, NC). A regression analysis was performed to determine the relationship between the change in stiffness and the height of the medial malleolus (Microsoft Excel, Microsoft Corporation, Redmond, WA). Statistical significance was set at a p level of 0.05.

RESULTS

The coefficient of repeatability was ± 0.17 Nm per degree in plantarflexion, ± 0.16 Nm per degree in dorsiflexion, ± 0.38 Nm per degree in inversion, and ± 0.18 Nm per degree in eversion. There was a significant decrease in overall brace stiffness when the holes were added to the orthosis ($p = 0.0038$). When the change in stiffness was examined as individual planes of motion, the stiffness decreased significantly during sagittal plane motion, but not with coronal plane motion (Figure 2). Although the decrease in stiffness was statistically significant for inversion, the difference between the intact and modified brace was less than the coefficient of repeatability. Therefore, the decrease in inversion stiffness was not considered to be clinically significant. Plantarflexion stiffness increased with an increase in the height of the hole in medial malleolus (Figure 3). The coefficient of determination (R^2) was 0.874, showing a strong relationship between the height of the medial malleolus and the brace stiffness in plantarflexion.

DISCUSSION

This study found significant changes in sagittal plane stiffness after modification of an Arizona AFO for gait analysis. In comparison, Lai et al.[9] tested the sagittal plane stiffness of five polyethylene AFOs and five hindfoot orthoses (HFO), designed to treat ankle and subtalar joint arthritis before and after modification for gait analysis. Braces were tested intact, with the malleoli cut out and with holes over the medial, lateral, and posterior aspects of the calcaneus (in combination with the malleoli cutouts). No difference was detected in plantarflexion stiffness; however, there was a significant decrease in stiffness in dorsiflexion. The authors concluded that the overall stiffness of the brace was not affected by the cutouts in the braces. The differences in these two studies were probably due to the differences in the construction of the two orthotics. Specifically, different polymers were used in the two orthoses, and there is no polymer material in the heel of the Arizona AFO, which may make the orthotic more susceptible to a decrease in stiffness with modification. Additionally, the AFO tested in the last study was five times stiffer in plantarflexion and 10 times stiffer in dorsiflexion

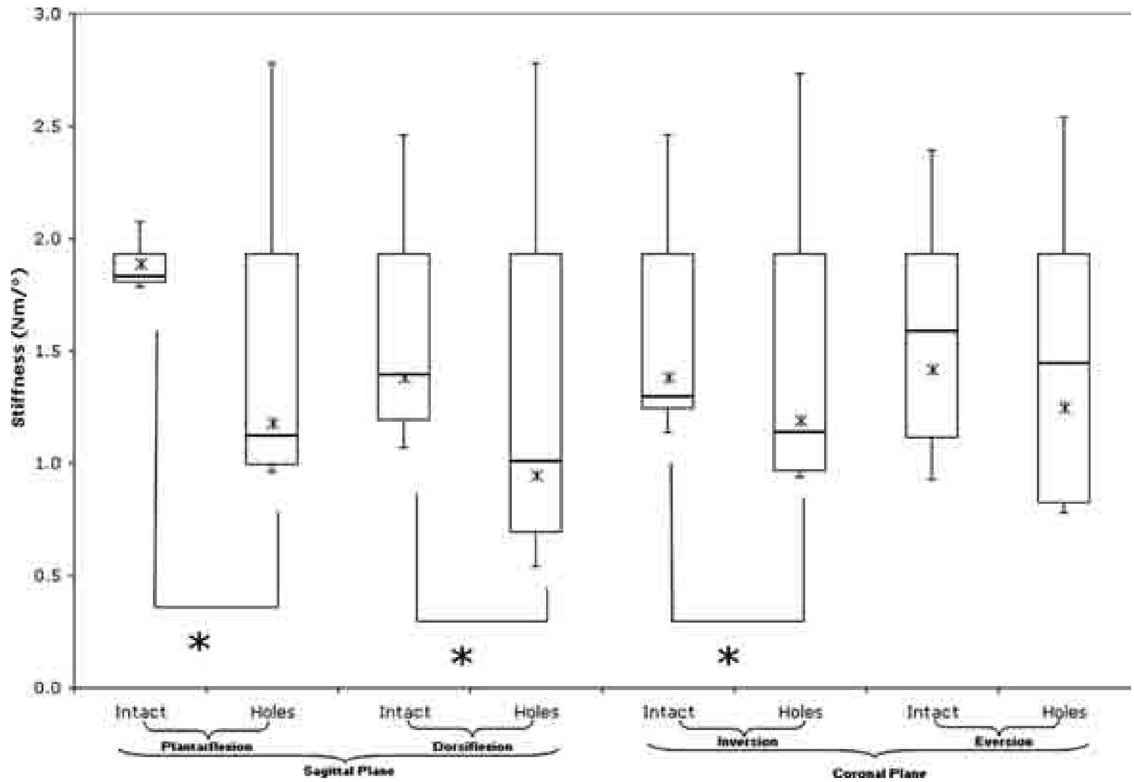

Figure 2. Box Plot showing the mean values and ranges of the stiffness for the intact and modified Arizona AFOs in the sagittal and coronal planes. *Statistically significant difference. The decrease in sagittal plane stiffness was significant in both directions (i.e., plantarflexion and dorsiflexion). The decrease in inversion stiffness was statistically significant; however, the value of the difference was within the limits of the coefficient of repeatability. Therefore, the change was not considered to be clinically significant.

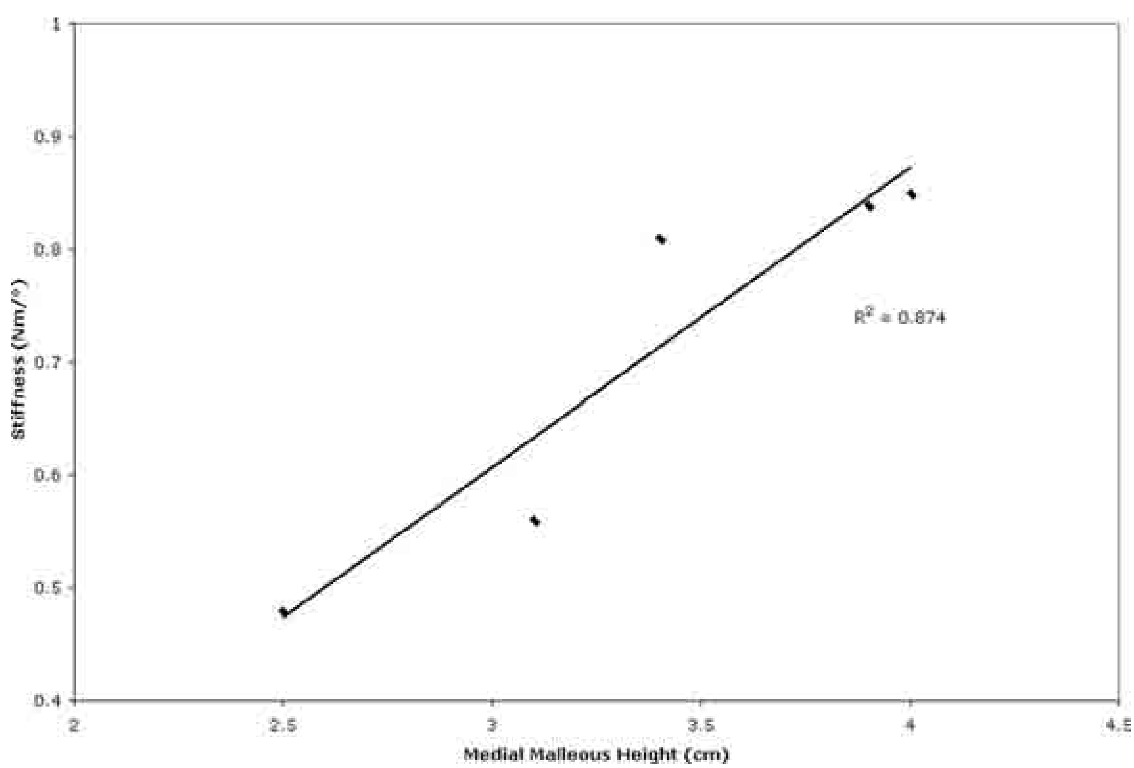

Figure 3. The change in plantarflexion stiffness increased as the height of the medial malleolus increased.

than the Arizona AFO, when intact. The intact HFO was almost twice as stiff in plantarflexion and almost three times stiffer in dorsiflexion than the Arizona AFO. The decrease in stiffness was greater in the AFO and the HFO once holes were placed in the orthotics, although these changes were not significantly different (e.g., ~4 Nm per degree from intact to holes in plantarflexion was observed in the AFO, which was a 40% decrease in stiffness). Therefore, a recommendation is made that studies examining the magnitude of the changes in stiffness, not just the statistical significance.

The stiffness of five Arizona AFOs was calculated during motion to a maximum of 10° in plantarflexion, 9° in dorsiflexion, 5° in inversion, and 5° in eversion. In a previous study of foot and ankle kinematics during gait,[10] the mean maximum plantarflexion and dorsiflexion was 4.6° and 7.6° in a rigid HFO and 4.3° and 5.3° in a rigid AFO, respectively. Maximum inversion and eversion was 3.5° and 3.2° in the HFO and 2.3° and 3.8° in the AFO, respectively.[10] Although the Arizona AFO has not been tested during gait, we expect it to limit the hindfoot motion similarly to the rigid AFO or the HFO. Therefore, we believe that the Arizona AFOs were tested in a range of motion consistent with in situ displacements.

Few studies have been performed to quantify AFO stiffness. In previous studies of various AFO designs, the stiffness in plantarflexion has ranged from 1.3 to 7.2 Nm per degree and the stiffness in dorsiflexion has ranged from 0.7 to 7.0 Nm per degree.[11–13] The measured sagittal plane stiffness of the Arizona AFO was within the range of other AFOs. Yamamoto et al.[14] tested 11 AFOs in inversion and eversion. The eversion stiffness ranged from 0.4 to 1 Nm per degree and inversion stiffness ranged from 0.1 to 1.25 Nm per degree. The coronal plane stiffness of the Arizona AFO was higher than the AFOs tested by Yamamoto et al., which could be due to differences in trimlines in the AFO designs.

Based on the results of this study, the Arizona AFO would need to be reinforced to achieve the intact stiffness. The holes drilled significantly decreased the sagittal plane stiffness of the brace. Furthermore, the decrease in plantarflexion stiffness was related to the medial malleolus hole height (Figure 3). This suggests that if holes are drilled in an Arizona AFO, it should be reinforced on the medial side before it can be used in a gait analysis study. All five Arizona AFOs were 10-cm high. Therefore, as the ratio of medial malleolus hole height to brace height increases (i.e., when the height of the hole in the medial malleolus increases, whereas the brace height did not change), more reinforcement will be needed.

This study had several limitations. AFOs are designed to be worn in a shoe. To our knowledge, no other study examining the stiffness of an AFO tested the orthosis in a shoe. However, there may be no significant decrease in stiffness when a shoe is worn. In the experimental set up, there was no simulated ankle joint, so that there would not be any motion restrictions. It is possible that when a human foot and ankle are placed in the brace, the decrease in stiffness may not be significant. Future studies should include a test of the Arizona AFO with an intact human limb, while a shoe is worn in combination with the AFO.

CONCLUSION

This study indicated that an Arizona AFO modified for gait analysis studies must be reinforced on the medial side.

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