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Ankle-foot orthosis bending axis influences running mechanics☆

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ABSTRACT

Passive-dynamic ankle-foot orthoses (AFOs) are commonly prescribed to improve locomotion for people with lower limb musculoskeletal weakness. The clinical prescription and design process are typically qualitative and based on observational assessment and experience. Prior work examining the effect of AFO design characteristics generally excludes higher impact activities such as running, providing clinicians and researchers limited information to guide the development of objective prescription guidelines. The proximal location of the bending axis may directly influence energy storage and return and resulting running mechanics. The purpose of this study was to determine if the location of an AFO's bending axis influences running mechanics. Marker and force data were recorded as 12 participants with lower extremity weakness ran overground while wearing a passive-dynamic AFO with posterior struts manufactured with central (middle) and off-centered (high and low) bending axes. Lower extremity joint angles, moments, powers, and ground reaction forces were calculated and compared between limbs and across bending axis conditions. Bending axis produced relatively small but significant changes. Ankle range of motion increased as the bending axis shifted distally ($p < 0.003$). Peak ankle power absorption was greater in the low axis than high ($p = 0.013$), and peak power generation was greater in the low condition than middle or high conditions ($p < 0.009$). Half of the participants preferred the middle bending axis, four preferred low and two preferred high. Overall, if greater ankle range of motion is tolerated, a low bending axis provides power and propulsive benefits during running, although individual preference and physical ability should also be considered.

1. Introduction

Passive-dynamic ankle-foot orthoses (AFOs) are commonly prescribed to improve gait and function [1–3]. The mechanical properties and design features of an AFO are important considerations and previous studies have shown how AFO stiffness, for example, can influence gait performance [4–7] and energy cost [8]. However, the design and prescription process still remains largely qualitative, and further modifications to AFO design characteristics may be advantageous to the end user.

Passive-dynamic AFOs store energy by deforming while loaded and return that energy to assist with important mechanical functions such as forward propulsion. Where this deformation occurs along the AFO may influence energy storage and return. It is generally accepted that the rotational axis of an AFO should coincide with the anatomical axis of the ankle joint [9–12] to prevent tissue damage, gait inefficiency, and mechanical wear on the orthosis [12,13]; however there is little quantitative evidence to support this. Sumiya et al. [14] questioned this rationale on the basis that an orthotic ankle joint aligned perfectly with a biological ankle joint would allow the talocrural joint to move but could impede subtalar joint motion [15] and result in overall unnatural ankle motion. The authors tested this hypothesis by varying the center of rotation of non-articulated AFOs by altering trimlines about the ankle and found that an AFO tends to deform at the level of the ankle joint during plantarflexion movements and deforms distal to the ankle joint center during dorsiflexion movements, resulting in pistoning of the AFO [14].

The majority of studies on AFOs have been conducted during walking, and AFOs are most commonly prescribed to improve or enable walking. However, the ability to perform higher energy activities, such as running, is desirable for many individuals post-injury and there are some AFOs specifically designed to restore running ability for high

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functioning individuals [16,17]. For example, the carbon fiber Intrepid Dynamic Exoskeletal Orthosis (IDEO) constitutes one class of passive-dynamic AFOs that affords many users the ability to return to running and sports participation [17,18]. The design characteristics of these passive-dynamic AFOs should be customized for the individual user and often are implemented in clinical practice, but limited quantitative information exists to guide the specific customization criteria [19]. The effect of AFO strut stiffness on running gait has been previously examined [20], but other design parameters, such as the optimal bending axis for running, are as of yet untested but potentially useful for AFO fabrication and prescription. Therefore, the purpose of this study was to determine if shifting the vertical location of a passive-dynamic AFO bending axis away from the biological ankle joint center alters running mechanics. We hypothesized that a low bending axis closer to the anatomical ankle joint would improve running mechanics compared with more proximal locations.

2. Methods

2.1. Subjects

Twelve military service members with unilateral lower limb weakness due to traumatic injuries participated in this study (Table 1). All subjects were frequent or constant users of the IDEO for walking and all used it for running. Specific inclusion criteria consisted of the following: (1) 18–55 years of age, (2) unilateral lower extremity injury resulting in ankle plantar flexion weakness, (3) currently using a custom carbon fiber AFO (IDEO), (4) able to comply with gait analysis testing, (5) able to walk without an assistive device other than an AFO, and (6) able to provide written informed consent. Exclusion criteria consisted of the following: (1) neurologic, musculoskeletal or other disease states affecting the contralateral limb that limits normal locomotion, (2) ankle plantarflexion weakness as a result of spinal cord injury or central nervous system pathology, and (3) non-ambulatory. Subjects were recruited over a 2-year time period.

2.2. Strut fabrication

The IDEO is a passive-dynamic AFO with a carbon fiber proximal tibial cuff, footplate, and a posterior strut connecting the two (Fig. 1) and is semi-rigid without an articulated “ankle”. The length and stiffness properties of the strut were clinically prescribed by the prosthetist/orthotist. For the purposes of this study, a strut was manufactured to mimic the mechanical properties of the IDEO strut using a selective laser sintering technique (SLS, Vanguards HiQ/HS Sinterstation, 3D Systems, Inc., Rock Hill, SC, USA) with unfilled Nylon 11 (PD D80-ST, Advanced Laser Materials, LLC, Temple, TX, USA). A previously described SLS framework for designing and manufacturing the middle bending axis SLS strut was used and summarized by Harper et al. [21]. The stiffness characteristics were tested and verified using a three-point-bend procedure described by Ranz et al. [22] and were on average 88.98 Nm/kg (SD: 14.94, range: 61.85–105 Nm/kg). A second strut was manufactured to be of identical stiffness properties, but the flex region of the strut was centered 30% of the bolt hole-to-bolt hole distance off-center. This region was created by decreasing the cross-sectional area of the SLS strut using a 2.54 cm radius circular extrusion in the posterior aspect of the strut. The strut with the offset bending axis could be positioned on the IDEO with the extrusion located either high, or flipped to be located low, near the ankle joint center (Fig. 1). Flexible lead tape with negligible stiffness was added to the offset bending axis strut to match the mass of the centrally located bending axis strut. Additional details of the fabrication, testing, and validation framework can be found in Ranz et al. [22].

Table 1
Subject characteristics (MVA, motor vehicle accident; GSW, gunshot wound; blast injuries were all combat-related).

Subject	Age (years)	Height (m)	Mass (kg)	Mechanism of injury
1	30	1.75	79.1	MVA
2	30	1.76	78.2	MVA
3	36	1.78	75.5	Blast
4	22	1.64	80.3	Blast
5	27	1.82	92.9	MVA
6	36	1.95	82.2	Blast
7	26	1.86	110.4	GSW
8	33	1.77	90.7	GSW
9	36	1.76	83.6	MVA
10	22	1.97	99.5	MVA
11	25	1.69	91.4	GSW
12	40	1.77	90.8	GSW
Mean (SD)	**30 (6)**	**1.79 (0.10)**	**87.9 (10.1)**	

2.3. Equipment setup

Subjects were tested using a six-degree-of-freedom marker set with 57 retro-reflective markers according to the procedures described by Wilken et al. [23]. A 26-camera motion capture system (120 Hz, Cortex, Motion Analysis Corporation, Santa Rosa, CA, USA) recorded marker trajectories. A digitizing wand (Visual3D, Inc., Germantown, MD, USA) was used to identify anatomical reference points used for joint center calculations [23]. Three centrally located force platforms (1200 Hz, AMTI, Inc., Watertown, MA, USA) were embedded in the 20 m runway and used to record ground reaction force data.

2.4. Protocol

Subjects ran across the runway at their self-selected speed in three strut conditions: high, middle (centrally located bending axis), and low bending axis (Fig. 1). The order was randomized. Subjects were given a minimum of 30 min accommodation time in each bending axis condition prior to testing. During each condition, subjects performed trials until five good strides were collected on each leg. A good stride was defined as one in which the full foot made contact with one of the force platforms and all markers were present during the stride from heel strike to ipsilateral heel strike. After all sessions were completed, participants indicated their preference ranking for each bending axis condition.

Bilateral isometric ankle strength was tested with a dynamometer (Biodex Medical Systems, Inc., Shirley, NY, USA). Ankle dorsi- and plantarflexor strengths were calculated on both the affected and unaffected limbs during an isometric maximum voluntary contraction in a seated position with the ankle at 15° of plantar flexion, unless these movements were painful and could not be completed by the subject. Ten trials were performed and the maximum value of the 10 trials was recorded.

2.5. Data analysis

Data were initially digitized in Cortex and exported to Visual3D (Visual3D, Inc., Germantown, MD, USA) for further analysis. Kinematic and kinetic data were interpolated using a cubic spline and filtered at 6 and 50 Hz, respectively, with fourth-order Butterworth low-pass filters. A 13-segment full body model was created and scaled by subject mass, inclusive of IDEO mass, and height. Newton–Euler angles were calculated to determine lower extremity kinematics using Cardan rotation sequences [24,25], and inverse dynamics were used to compute joint moments and powers. Moments and powers were scaled to body mass, and ground reaction forces were scaled to body weight. Peak kinetic and kinematic values were identified during stance and/or early swing and ranges of motion (ROM) were identified across the gait cycle. Individual trial peaks were averaged within a subject and then across subjects.

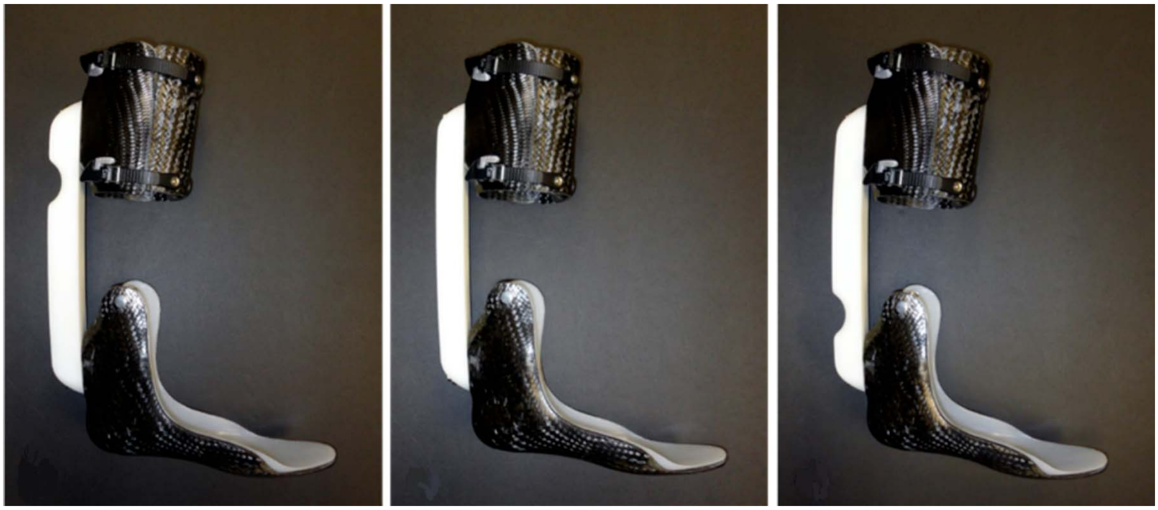

Fig. 1. Passive-dynamic ankle-foot orthosis (AFO) with the middle bending axis (middle image) and with high (left) and low (right) bending axes.

2.6. Statistics

A two-way repeated-measures analysis of variance was used to identify significant differences among conditions, between limbs, and interactions. *Post hoc* paired *t*-tests with sequentially rejective Bonferroni–Holm corrections for three comparisons identified significant differences between bending axis conditions within each limb. The unaffected limb was used as a control and the same *post hoc* analysis was used to compare inter-limb differences within each bending axis condition. The unadjusted criterion for statistical significance was set at $p < 0.05$.

3. Results

Three subjects were unable to complete the maximal isometric plantarflexion strength test, and four were unable to complete the maximal isometric dorsiflexion strength test. On average, plantarflexion torque was 53 (22)% less on the affected limb than the unaffected and dorsiflexion torque was 22 (35)% less. All subjects were able to complete the running protocol in all three strut configurations.

3.1. Effect of AFO bending axis

Self-selected running speeds were not significantly different among the bending axis conditions. On average, running speeds for the high, middle, and low bending axes were 3.43, 3.45, and 3.53 m/s, respectively ($p = 0.960$). Overall, there was a preference for the middle bending axis with 50% of participants preferring it over the low (33%) or high (17%) bending axis condition. Fifty percent of the participants ranked the low condition as their second preference (33% middle and 17% high).

The primary differences among the bending axis conditions were found at the ankle joint of the affected limb (Fig. 2). There were no significant interactions between bending axis condition and limb for any of the ankle kinematics, moments, or powers. There was a significant main effect of condition ($p = 0.010$) such that ankle ROM progressively increased as the bending axis became more distal ($p < 0.003$ for all comparisons); however, individual peak values were not significantly different among the conditions. The increase in ankle ROM during the low bending axis condition contributed to a significant 10% increase in peak ankle power absorption during stance in the low condition compared to the high ($p = 0.013$), and 15 and 22% greater ankle power generation at push-off in the low condition compared to both the middle ($p = 0.009$) and high ($p = 0.002$) conditions, respectively. Moreover, the peak propulsive ground reaction force at push-off was significantly greater in the low condition compared to the middle condition ($p = 0.013$), but not the high condition ($p = 0.097$).

There were no significant differences among bending axis conditions for peak sagittal plane knee kinematics or kinetics. There was a significant interaction between limb and condition for sagittal plane hip ROM ($p = 0.013$), such that while there were no significant differences in peak values, there was on average 5% greater hip ROM on the affected side in the low condition compared to the high condition ($p = 0.002$). Peak-dependent measures and comparisons among bending axis locations and relative to the unaffected limb can be found in Table 2.

3.2. Effect of limb

The differences between bending axis conditions occurred primarily at the ankle but the differences between limbs spanned all three joints of the lower extremity. There was a significant main effect of limb at the ankle ($p = 0.001$) such that ROM was on average 183% greater in the unaffected limb compared to the affected limb ($p = 0.001$). The unaffected limb had on average 40% greater ankle dorsiflexion during stance compared to the affected limb ($p < 0.002$ for all comparisons). At the knee, there was a significant main effect of limb ($p = 0.018$) for the initial peak internal knee flexor moment during initial stance, such that in the middle and high bending axis conditions, the affected limb produced 35% ($p = 0.012$) and 30% ($p = 0.021$) lower knee flexor moment, respectively, than the unaffected limb. In the low condition, there was no significant difference between limbs in the initial peak internal knee flexor moment ($p = 0.066$). The peak internal knee extensor moment during mid stance was not, however, significantly different between limbs or conditions. There were significant main effects of limb for peak hip extensor moment during initial stance ($p = 0.001$) and late stance/early swing ($p = 0.002$), such that across bending axis conditions, the unaffected limb produced on average 36 and 27% greater hip extensor moment, respectively, than the affected limb ($p < 0.007$ for all comparisons).

There was a significant main effect of limb for peak ankle power absorption ($p = 0.019$) and generation ($p < 0.001$), such that peak ankle power absorption was 40% greater ($p < 0.034$ for all comparisons) and peak ankle power generation was 158% greater ($p < 0.001$ for all comparisons) in the unaffected limb than the affected limb. There were no significant differences between limbs for peak stance phase knee or hip power absorption or generation.

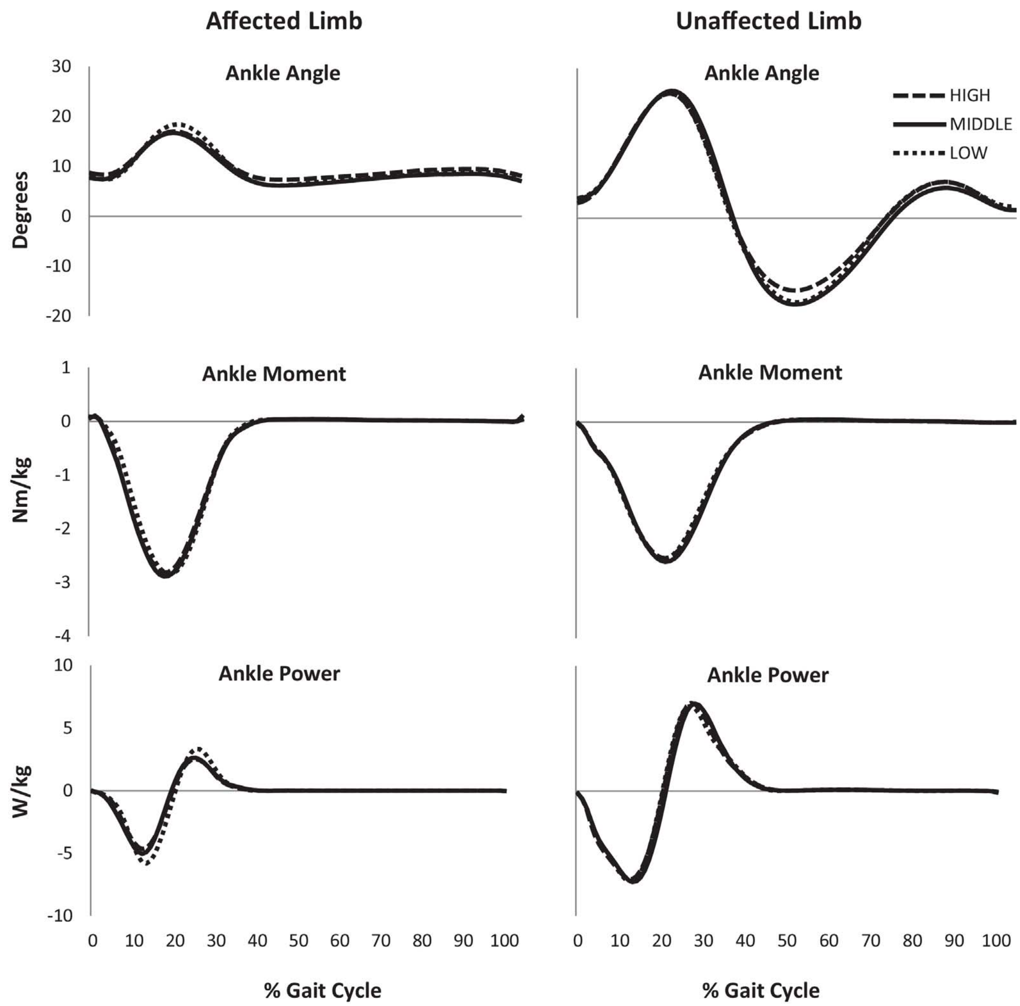

Fig. 2. Mean ankle joint angles (dorsiflexion +), internal joint moments (dorsiflexor +), and joint power across the gait cycle.

Table 2
Mean ± standard deviation for the peak kinematic and kinetic data at the ankle, knee, and hip, and ranges of motion (ROM). Moments are presented as internal joint moments. There were few differences between unaffected limbs in the different bending axis conditions and only the values from the middle condition are presented.

Parameter	Affected limb			Unaffected limb
	Low	Middle	High	Middle
Ankle ROM (deg)	**16.36 ± 9.68**^{M,H}	**14.65 ± 10.17**^{L,H}	**13.93 ± 10.24**^{L,M}	42.90 ± 11.69
Ankle dorsiflexor moment (Nm/kg)	0.20 ± 0.10	0.16 ± 0.12	0.17 ± 0.10	−0.01 ± 0.15
Ankle plantarflexor moment (Nm/kg)	−2.97 ± 0.59	−2.97 ± 0.54	−2.87 ± 0.56	−2.78 ± 0.33
Ankle power abs. (W/kg)	**−6.83 ± 1.80**^{H}	−6.52 ± 1.49	**−6.18 ± 1.55**^{L}	−9.09 ± 2.76
Ankle power gen. (W/kg)	**4.22 ± 1.49**^{M,H}	**3.66 ± 1.44**^{L}	**3.45 ± 1.44**^{L}	9.46 ± 2.98
Knee ROM (deg)	94.60 ± 19.80	89.31 ± 17.72	90.92 ± 19.05	88.19 ± 19.69
Knee flexor moment (Nm/kg)	−0.95 ± 0.44	**−0.90 ± 0.31**	**−0.87 ± 0.44**	−1.22 ± 0.42
Knee extensor moment (Nm/kg)	1.23 ± 0.58	1.17 ± 0.52	1.20 ± 0.52	1.65 ± 0.52
Knee early stance power abs. (W/kg)	−4.30 ± 2.57	−4.06 ± 1.96	−3.82 ± 2.12	−5.13 ± 2.43
Knee mid-stance power gen. (W/kg)	2.54 ± 1.38	2.32 ± 1.03	2.09 ± 1.19	2.89 ± 1.18
Knee late stance power abs. (W/kg)	−2.41 ± 1.45	−2.25 ± 1.27	−2.32 ± 1.48	−2.92 ± 2.08
Hip ROM (deg)	58.34 ± 10.49^{H}	57.24 ± 7.32	55.32 ± 9.94^{L}	56.78 ± 11.39
Hip flexor moment (Nm/kg)	−1.19 ± 0.50	−1.18 ± 0.44	−1.19 ± 0.48	−1.19 ± 0.45
Hip extensor moment (Nm/kg)	**2.80 ± 0.96**	**2.73 ± 0.88**	**2.58 ± 1.03**	3.78 ± 1.23
Hip stance power abs. (W/kg)	4.16 ± 2.76	3.64 ± 1.86	3.60 ± 2.58	3.83 ± 2.49
Hip stance power gen. (W/kg)	−2.98 ± 2.07	−2.90 ± 1.99	−2.74 ± 2.21	−2.14 ± 1.64
Propulsive GRF (BW)	**0.21 ± 0.09**^{M}	**0.20 ± 0.07**^{L}	**0.20 ± 0.08**	0.29 ± 0.10

ROM, range of motion; GRF, ground reaction force; Abs., absorption; Gen, generation. Superscript H, M, and L indicate significant differences from high, middle, and low bending axis conditions, respectively, on the affected limb. Bold values indicate a significant difference from the unaffected limb in the same bending axis condition.

4. Discussion

The purpose of this study was to determine how altering the vertical location of a passive-dynamic AFO's bending axis affected running mechanics. A better understanding of the relationship between the bending axis location and running mechanics could contribute to the development of evidence-based prescription guidelines and, thus, the improvement of rehabilitation outcomes for AFO users. These present findings suggest that a low bending axis, near the vertical location of the anatomical ankle joint, provides greater ankle ROM, power generation, and propulsive ground reaction force than more proximal locations. These results would be considered an improvement in AFO design for individuals able to tolerate increases in ankle joint motion on their affected limb. However, sometimes, AFOs are specifically prescribed to control ROM about the ankle joint due to pain or injury and an AFO with a high bending axis may be preferable for these individuals. Half of the individuals preferred the middle bending axis, suggesting that the middle bending axis may provide the best compromise between the varied tolerance for ROM and need for power generation.

Previous studies examining the effect of AFO bending axis location have been performed during walking. Biomechanical data support the use of a low bending axis or orthosis joint axis located near the anatomical ankle joint [13,14], with a bending location exactly at the intermalleolar axis being preferable for articulated AFOs [26]. During running, greater deflection and deformation of the posterior strut would be expected than during walking, and deformation of the posterior strut at an optimal location could be particularly beneficial during running.

For many individuals undergoing lower limb reconstruction procedures, running is often not an option or recommended. As a result, the inability to return to sports and physical activity, among other daily activities, leads some to consider amputation. However, advancements in AFO design and rehabilitation have allowed many individuals to maintain significant activity levels post-injury [17]. As the number of individuals using AFOs for running and sports participation increases, the design and manufacturing criteria are of greater focus. While the IDEO used in this study is custom-fabricated for each individual, a more generic model would provide increased availability to a broader population. Therefore, design considerations such as stiffness [20] and bending axis location are important features for future AFO designs.

Although AFOs are commonly prescribed to improve function in individuals with lower limb impairments, gait asymmetries still exist in almost all cases. Previous work has shown that passive AFOs reduce ankle ROM while walking [22,27] and running [20] in an effort to support the ankle joint. In the present study, the limited dorsiflexion and total ankle ROM in the affected limb are of approximately the same magnitude as previously reported values [20]. Decreased ankle power absorption in the affected limb was also in the same range as previously reported while running in the same style AFO [20]. Compared to able-bodied running studies, peak ankle power absorption was greater in the unaffected limb, but similar in the affected limb [28,29]. Any differences are attributed to our participants using a mixture of footstrike patterns, since previous work has shown that forefoot striking runners have greater peak ankle power absorption than rearfoot striking runners [30].

One limitation of this study is that the unaffected limb was used as a control in comparison to the affected limb. However, it cannot be assumed that the unaffected limb's running mechanics match the mechanics of an able-bodied control subject, as some compensation with the unaffected limb may have occurred. In addition, the population investigated in this study was heterogeneous in terms of injury type, severity, and length of rehabilitation. Therefore, the effects of bending axis on running mechanics may not fully translate across other populations with different injuries or AFO designs. Next, passive ankle ROM was not recorded as part of this study, and it is not definitively known the extent to which the AFO or the injury limited ankle motion relative to the unaffected limb. However, the orthotists who fit these patients typically position them to avoid end range of motion and allow function within their available range during daily tasks (which includes running in the present patient cohort). In addition, ankle ROM during walking in an IDEO remains nearly identical across studies that require passive ankle ROM as an inclusion criteria [31] and those that do not specify [7,20,22]. Thus, ankle ROM when ambulating in the IDEO is likely due primarily to the design properties of the device. Lastly, the subjects who participated in this study required an AFO for running and, therefore, data were not collected on running mechanics without the device, due to concern for patient safety.

5. Conclusion

The design features of an AFO can have a significant impact on the user's movement biomechanics. This study investigated the effects of bending axis location on running mechanics. A low vertical location of the bending axis, closer to the anatomical ankle joint, provides increased power and propulsive benefits during running. A low bending axis may be useful for individuals able to tolerate the increased ankle ROM. The biomechanics of proximal joints were relatively unaffected by changes in the bending axis location. Individual preference and physical ability should also be considered when prescribing an AFO with an offset bending axis for running.

Conflict of interest

Elizabeth Russell Esposito, Ellyn C. Ranz, Kelly A. Schmidtbauer, Richard R. Neptune, and Jason M. Wilken certify that he/she has no commercial associations (e.g., consultancies, stock ownership, equity interest, patent/licensing arrangements, etc.) that might pose a conflict of interest in connection with the submitted article. The authors declare they have no conflicts of interest.

Ethical review

This study has been approved by the Brooke Army Medical Center Institutional Review Board.

Location

The experimental data collection and analysis were performed at the Center for the Intrepid at Brooke Army Medical Center, JBSA Ft. Sam Houston, TX, USA. The AFO strut fabrication was performed at The University of Texas at Austin.

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