The shank-to-vertical angle as a parameter to evaluate tuning of ankle foot orthoses

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ABSTRACT

The effectiveness of an ankle foot orthosis-footwear combination (AFO-FC) may be partly dependent on the alignment of the ground reaction force with respect to lower limb joint rotation centers, reflected by joint angles and moments. Adjusting (i.e. tuning) the AFO-FC’s properties could affect this alignment, which may be guided by monitoring the shank-to-vertical angle (SVA). This study aimed to investigate whether the SVA during walking responds to variations in heel height and footplate stiffness, and if this would reflect changes in joint angles and net moments in healthy adults. Ten subjects walked on an instrumented treadmill and performed six trials while walking with bilateral rigid AFOs. The AFO-FC heel height was increased, aiming to impose a SVA of 5°, 11° and 20°, and combined with a flexible or stiff footplate. For each trial, the SVA, joint flexion-extension angles and net joint moments of the right leg at midstance were averaged over 25 gait cycles. The SVA significantly increased with increasing heel height ($p<0.001$), resulting in an increase in knee flexion angle and internal knee extensor moment ($p<0.001$). The stiff footplate reduced the effect of heel height on the internal knee extensor moment ($p=0.030$), while the internal ankle plantar flexion moment increased ($p=0.035$). Effects of heel height and footplate stiffness on the hip joint were limited. Our results support the potential to use the SVA as a parameter to evaluate AFO-FC tuning, as it is responsive to changes in heel height and reflects concomitant changes in the lower limb angles and moments.
INTRODUCTION

Ankle foot orthoses (AFOs) are frequently applied in patients with neurological disorders, aiming to normalize joint kinematics and joint kinetics during walking\cite{1-4}. Although it has been shown that AFOs can significantly improve sagittal joint kinematics and kinetics\cite{2,3,5-7}, inadequate alignment of ground reaction force (i.e. distant from the joint rotation centers) during walking negatively impacts the effectiveness\cite{4,8,9}.

Tuning of the AFO optimizes the alignment of the ground reaction force with respect to the joint rotation centers, enhancing normalization of the joint kinematics and kinetics\cite{8-12}. Such tuning can be described as the process in which the properties of an AFO-footwear combination (AFO-FC) are manipulated. Commonly used adjustments comprise changing the footplate stiffness to affect the point of application of the ground reaction force, and altering the heel-sole differential (i.e. the difference in height between the heel and forefoot of the shoe), which affects shank orientation\cite{8}. The combined effect of the AFO-FC’s ankle angle and heel-sole differential can be described in terms of the shank-to-vertical angle (SVA). The SVA, i.e. tibia inclination, is the angle between the anterior surface of the tibia and the vertical in the global sagittal plane\cite{8,13}. It is clinically often measured using sagittal video recordings\cite{13}. The SVA is considered inclined, when the shank is tilted forward, or reclined, when it is tilted backward with respect to the vertical. Owen\cite{13} suggested that an appropriate shank orientation at midstance aligns the ground reaction force to the joint rotation centers, which contributes to stability, facilitates adequate switching from flexion to extension moments at the knee and hip, and lowers vertical center of mass excursion. Accordingly, the SVA at midstance may be an important and relatively simple parameter to evaluate the effects of adjustments to the AFO-FC during its tuning process\cite{8,13}, also because information on the ground reaction force and calculations of joint moments are not always available in clinical practice.

Several studies in patients with neurological disorders report the SVA, and describe a normalization of gait parameters following changes of the heel-sole differential\cite{11,12,14,15}. However, in all available studies, the SVA was measured while the patient was in a static position, whereas there is no evidence showing that the SVA in this position represents the SVA at midstance\cite{8}. Evidence on the effects of changing the footplate stiffness on the SVA, as well as on joint kinematics and kinetics is also lacking. Yet, in clinical practice, such manipulations of footplate stiffness, in addition to changing the heel-sole differential are commonly applied. Since tuning of these AFO-FC properties is generally guided by monitoring the SVA at midstance, insight is needed in how the SVA responds to changes
of the heel-sole differential and footplate stiffness, in order to assess its potential as a parameter to evaluate the effects of such manipulations.

To this end, we evaluated in healthy young adults i) whether the SVA at midstance can be influenced during walking with an AFO-FC by applying commonly used manipulations within the process of tuning (i.e. changing AFO-FC heel-sole differential and footplate stiffness), and ii) how changes in the SVA, as a result of the manipulations, are reflected in ankle, knee and hip flexion-extension angles and net internal joint moments at midstance. We hypothesized that the SVA would be responsive to changes in AFO-FC heel height and that this would be reflected by increased knee and hip flexion angles and net internal joint extension moments at midstance. As a stiff footplate mainly aims to shift the ground reaction force forward without affecting joint flexion-extension angles, we expected no response of the SVA to changes in footplate stiffness, while it was expected to affect internal net joint moments.

METHODS

Participants

Ten healthy young adults (3 male; mean (SD) age: 24 (3) years; mean (SD) body mass index: 22.8 (2.2)) participated. All subjects provided written informed consent in accordance with the procedures of the Institutional Review Board of the VU University.

Materials

For this study, two pairs of prepeg carbon AFOs were manufactured (European shoe size: 39 and 43) (see Figure 3.1A). Each participant chose the best fitting pair. The stiffness of the AFOs at the ankle and metatarsal joints was measured using BRUCE\[16\], which is an instrument to define AFO mechanical properties. The AFOs were rigid at the ankle (7.9 Nm·deg\(^{-1}\)), aiming to immobilize the ankle joint at 0°.

According to Owen\[13\], important kinematic characteristics at midstance (e.g. thigh inclination) can only be preserved with an SVA ranging from 7°-15°, while an SVA of 10°-12° is suggested to be optimal. In the current study, AFO-FC heel-sole differential was varied using three heel heights by applying insole wedges, aiming to impose an SVA of 5°, 11° and 20° in static position. As such, the effects of SVA manipulations near the presumed optimum and outside the suggested optimal range were investigated. The height of the wedges was pre-defined for both AFO-FCs, using a dedicated instrument to measure heel height and heel-sole differential of an AFO-FC when doffed (Vertical Inclinometer
The shank-to-vertical angle

Figure 3.1. A) Picture of the AFO-FC of the right leg without insole wedges. B) Schematic representation of the VICTOR\(^\text{[17]}\), with virtual markers 12' and 13' as analogue reference points of the anatomical markers at the tibial tuberosity (Figure 2, #12) and tibia (Figure 2, #13). VICTOR was used to determine the height of the insole wedges to impose a SVA of 5°, 11° and 20° during the walking trials. Wedges were added to increase the height (h) of the heel probe until the inclination angle (α) reflected the pre-defined angles of 5°, 11° and 20°. C) Picture of the AFO-FC including insole wedges (pre-defined using VICTOR (h)), resulting in the heel height (HH). The SVA during walking was calculated as the angle between the line at the anterior surface of the tibia (dashed) (i.e. the line connecting the marker at the tibial tuberosity (#12) and tibia (#13)) and the vertical (dotted) in the global sagittal plane. The SVA was expected to represent α.

Dotted, the vertical as used for SVA calculation; dashed, line at the anterior surface of the tibia, representing the long axis of the shank in the global sagittal plane; solid, estimated position of the footplate in the shoe.

on a Rail (VICTOR\(^\text{[17]}\)) (see Figure 3.1B). Using VICTOR, low (size 39: 0.6 cm; size 43: 1.3 cm), medium (size 39: 2.8 cm; size 43: 2.8 cm) and high (size 39: 4.9 cm; size 43: 5.3 cm) heel heights were specified (see Figure 3.1). These heel heights were combined with two different degrees of footplate stiffness, which could be changed by adding a stiff inlay footplate (0.89 Nm·deg\(^{-1}\)) to the AFO’s flexible footplate (0.06 Nm·deg\(^{-1}\)). The provided shoes (i.e. flexible sneakers) were large enough to allow for the insole wedges.
Measurements

Subjects walked on the GRAIL system (Motek Medical BV, Amsterdam, the Netherlands), consisting of a split-belt instrumented treadmill (ForceLink®, Culemborg, the Netherlands) and a passive marker motion capture system (Vicon, Oxford, UK), collecting marker trajectories. Ground reaction forces were captured from force sensors mounted underneath both treadmill belts, and synchronized with kinematic data at 120 Hz.

Reflective markers were placed at anatomical landmarks according to the Human Body Model[18,19] (see Figure 3.2). The SVA was calculated as it is defined in clinical practice[13], i.e. using the line over the anterior surface of the tibia, representing the long axis of the shank, and calculated as the angle between this line and the vertical in the global sagittal plane (see Figure 3.1). In order to do so, additional markers were added to the Human Body Model (see Figure 3.2): at the tibial tuberosity (#12 and #21) and at a distal point on the tibia (#13 and #22, i.e. at 75% of the lower leg, measured from the tibial tuberosity (#12 and #21) to the floor and vertically in line with the marker at tibial tuberosity in the frontal plane). Other additional markers were placed at the dorsal shell of each AFO (#14 and #23), which were horizontally aligned with the tibial tuberosity marker (#12 and #21) in the sagittal plane and vertically aligned to the calcaneus marker (#16 and #25) in the sagittal plane. These markers were used to determine movements of the shank in the AFO, therewith evaluating the immobilization of the ankle. This was done for interpretation of the results, as inadequate immobilization is expected to affect joint flexion-extension angles and moments. The Human Body Model foot markers (#16-18 and #25-27) and the markers at the lateral malleoli (#15 and #24) were positioned on the shoe. None of the markers were replaced between different trials.

Procedure

After being provided with the AFO-FC, the subject accommodated to walking on a treadmill until he/she felt comfortable. Subsequently, the subject’s comfortable walking speed was determined following a standardized protocol. Following this protocol, the participant started walking at an initial speed of 0.8 m·s⁻¹. Treadmill speed was then gradually increased with 0.1 m·s⁻¹ until the participant indicated the speed as comfortable. From thereon, speed was further increased until comfortable speed +0.3 m·s⁻¹ and gradually decreased until the participant indicated the speed as comfortable again. The mean of both self-selected speeds represented the subject’s comfortable
Figure 3.2. Marker model according Human Body Model, with six additional markers (i.e. marker numbers 12, 13, 14, 21, 22, and 23).

*The markers referring to the distal point of the tibia were positioned at 75% of the lower leg, measured from the tibial tuberosity to the floor and in line with the tibial tuberosity marker in the frontal plane.
walking speed. Thereafter, subjects performed six walking trials of 2 minutes at this comfortable speed. For each trial, AFO-FC heel height was set into low, medium or high, and combined with either the stiff or flexible footplate. The sequence of these six combinations was randomly applied.

**Data processing**

Joint flexion-extension angles and net internal moments were calculated using the Human Body Model and D-flow software\(^{[18,19]}\). Joint flexion-extension angles were calculated using the orientation of the distal segment with respect to the orientation of proximal segment and expressed in the sagittal plane of the proximal segment. The SVA, calculated in the global sagittal plane, was defined as the angle between the anterior surface of the tibia and the vertical\(^{[13]}\) (see Figure 3.1). Another line was created using the position of the marker at the dorsal shell of the AFO (#14 and #23) and the lateral malleolus marker (#15 and #24) in the global sagittal plane. The angle between the two lines represented changes of the position of the shank with respect to the AFO (i.e. Shank-to-AFO angle). Assuming that this angle would be unchanged with a fully immobilized ankle joint, smaller angles would indicate movement of the shank towards the AFO’s dorsal shell. Calculations of the SVA and Shank-to-AFO angle were done using Matlab 2011 (The Mathworks, USA).

Marker and force plate data were low pass filtered at 6 Hz using the Human Body Model\(^{[19]}\). To select only strides with foot placement on a single belt, a stride was excluded if i) the force of that stride deviated more than 100% from the mean force of all strides, or ii) the length of the stride deviated more than 20 samples from the median length of all strides. For further processing, only correctly recorded strides were selected, based on two criteria i) single-belt foot placement and ii) sufficient marker data (i.e. no occlusion) to calculate the considered parameters. Subsequently, remaining strides were normalized to 100% gait cycle and the SVA, lower limb joint flexion-extension angles and net internal moments, and Shank-to-AFO angles of the right leg were determined at midstance, defined as the moment that the malleolus marker of the contralateral leg (#24) passed the malleolus marker of the ipsilateral leg (#15). The parameters were limited to midstance, as the SVA is clinically used to evaluate the effects of tuning at this stage of the gait cycle\(^{[13]}\). The parameters were averaged over 25 steps, which were selected starting from the end of the trial.
Statistics

The effects of the different AFO-FC conditions on the SVA, joint flexion-extension angles and net moments, and Shank-to-AFO angle were analyzed for statistical significance using a two-way repeated measures analysis of variance (ANOVA) with two within-subject factors (i.e. heel height (three levels) and footplate stiffness (two levels)), using Bonferroni post-hoc adjustments (α=5%). Statistical analyses were done using IBM SPSS Statistics, version 20 (SPSS Inc, Chicago, USA).

RESULTS

SVA

The SVA at midstance significantly increased with increasing heel height (see Figure 3.3A). The SVA during walking (mean (SD) walking speed: 0.96 (0.07) m·s⁻¹) was larger in all heel height conditions compared to the imposed SVA of 5°, 11° and 20° in static position (see Table 3.1). Footplate stiffness had no effect on SVA, and also no interaction effect of heel height and footplate stiffness on SVA was found (F=0.71, p=0.505) (see Figure 3.3A).

Knee joint

The effects of the AFO-FC manipulations were most prominent at the knee joint, with the knee flexion angle and internal knee extensor moment at midstance significantly increasing with increasing heel height (see Table 3.1). The stiff footplate tended to decrease the knee flexion angle and internal knee extensor moment at midstance, although this was only significant for the internal knee extensor moment. An interaction effect of heel height and footplate stiffness on the knee flexion angle (F=3.54, p=0.050) and internal knee extensor moment (F=4.06, p=0.035) was found, indicating an inhibiting effect of the stiff footplate for the low and high heel height conditions, but not for the medium heel height condition (see Table 3.1; Figure 3.3B-C).
Table 3.1. Main effects of heel height and foot plate stiffness manipulations on mean (SE) SVA, joint flexion-extension angles and internal net moments at midstance (n=10).

<table>
<thead>
<tr>
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<tbody>
<tr>
<td>low</td>
<td>14.3 (0.81)</td>
<td>10.5 (1.75)</td>
<td>17.2 (0.99)</td>
<td>13.1 (2.57)</td>
</tr>
<tr>
<td>med</td>
<td>17.4 (0.79)</td>
<td>13.4 (2.31)</td>
<td>24.5 (1.45)</td>
<td>16.4 (2.73)</td>
</tr>
<tr>
<td>high</td>
<td>24.3 (1.52)</td>
<td>16.7 (3.12)</td>
<td>32.7 (2.05)</td>
<td>19.1 (2.59)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>F</th>
<th>p</th>
<th>post hoc</th>
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<tbody>
<tr>
<td>71.7</td>
<td>&lt;0.001</td>
<td>l-m^a; l-h^b; m-h^a</td>
</tr>
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| ankle moment [Nm·kg^-1] | 0.69 (0.07) | 0.74 (0.10) | 0.91 (0.07) |
|_____|_____|_____|

| knee moment [Nm·kg^-1] | 3.81 (0.042) | 2.33 (0.020) | 3.74 (0.085) |
|_____|_____|_____|

| hip moment [Nm·kg^-1] | -0.01 (0.05) | -0.02 (0.04) | 0.02 (0.04) |
|_____|_____|_____|

<table>
<thead>
<tr>
<th>footplate stiffness</th>
<th>flex</th>
<th>stiff</th>
<th>F</th>
<th>p</th>
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<tbody>
<tr>
<td>low</td>
<td>18.6 (1.07)</td>
<td>18.7 (0.92)</td>
<td>0.07</td>
<td>0.805</td>
</tr>
<tr>
<td>med</td>
<td>14.3 (2.40)</td>
<td>12.7 (2.04)</td>
<td>0.020</td>
<td>0.805</td>
</tr>
<tr>
<td>high</td>
<td>25.8 (1.58)</td>
<td>23.8 (1.14)</td>
<td>3.74</td>
<td>0.085</td>
</tr>
</tbody>
</table>

Positive values represent joint flexion angles, and plantar flexion, knee extension and hip flexion moment.

l, m and h represent low, medium and high heel height respectively.

^a p<0.001
^b p<0.05
Ankle and hip joint

Increasing heel height resulted in a significant increase in ankle dorsal flexion angle, hip flexion angle and internal plantar flexion moment at midstance. The internal ankle plantar flexion moment further increased as a result of the stiff footplate, whereas ankle angle, hip angle, and internal hip moment were not affected by footplate stiffness (see Table 3.1). No interaction effects of heel height and footplate stiffness were found for ankle angle ($F=1.66, p=0.218$), hip angle ($F=0.24, p=0.790$), ankle moment ($F=0.32, p=0.732$), and hip moment ($F=0.05, p=0.953$).

Shank-to-AFO angle

Mean (SE) Shank-to-AFO angle at midstance significantly decreased with increasing heel height ($F=46.9, p<0.001$), with a mean (SE) angle of 18.3° (1.23) for the low, 15.3° (0.92) for the medium, and 13.0° (0.64) for the high heel height condition. Mean (SE) Shank-to-AFO angle was 15.4° (0.91) while walking with the flexible footplate, and 15.6° (0.92) with the stiff footplate ($F=0.945, p=0.356$). No interaction effect of heel height and footplate stiffness was found ($F=1.14, p=0.341$).

DISCUSSION

The present study demonstrates that the SVA is responsive to changes in the AFO-FC heel height, which resulted in an increase in lower limb joint flexion angles and net internal extension moments. In line with our hypothesis, the stiff footplate did not affect the SVA, although it did alter the net internal ankle and knee joint moments. The stiff footplate also affected the knee flexion angle, which is in contrast with our hypothesis.

A recent study of Jagadamma and colleagues\cite{12} showed the effects of tuning rigid AFOs on joint kinematics and kinetics in children with spastic cerebral palsy. In that study, tuning was based on inclining the SVA, starting from 12°, until the ground reaction force alignment during stance was closest to normal. They found that increasing the SVA resulted in an increased knee angle and a non-significant increase in peak hip flexion in stance. Another study of Jagadamma et al\cite{11} also showed that when the SVA was increased from 5.6° to 10.8° after tuning, the peak knee flexion angle in stance increased. This is comparable to our study, as the SVA increased with increasing heel height, resulting in an increase in knee and hip flexion angles. Our results also show an increase in ankle dorsiflexion angle with increasing heel height, while the rigid AFOs aimed to
Figure 3.3. Mean (n=10) shank-to-vertical angle, knee flexion-extension angle, and internal knee flexion-extension moment for different conditions, normalized to 100% gait cycle. Shaded area indicates normal walking.
immobilize the ankle in zero degrees. We presume that this was the effect of an offset between the foot markers and the position of the bony landmarks. More specifically, the foot markers (placed on the shoe) were not replaced between the trials, while the insole wedges lifted the foot inside the shoe.

Also comparable to our results, the tuned AFO-FC in the Jadamma study\(^\text{[12]}\) resulted in an increased internal peak knee extensor and ankle plantar flexion moment, whereas the peak hip moment remained unchanged. While we did expect to see these changes at the ankle and knee joints, the unchanged internal hip moment between conditions was in contrast to our hypothesis. In our study, subjects may have positioned the thigh such that the ground reaction force was aligned close to the hip joint at midstance, independent from shank kinematics, therewith showing similar internal hip joint moments between heel height conditions at this stage. Although AFO-FC heel height manipulations and the resulting changes in joint angles and moments found in Jagadamma’s studies\(^{[11,12]}\) were smaller compared to our study, the nature of their effects was similar. Hence, our study confirms the responsiveness of the SVA to changes in AFO-FC heel height, though providing a more systematic change of heel height and, additionally, analyzing the effect of adjusting footplate stiffness.

Since literature on the effect of footplate stiffness on joint angles and moments is lacking, our results might best be compared to a study on the effect of different AFO footplate lengths\(^{[20]}\). Similar to the non-significant decrease in knee flexion angle as a result from the stiff footplate in our study, Fatone and colleagues\(^{[20]}\) found a non-significant decrease in knee flexion angle while walking with the full-length footplate. The increase in the internal ankle plantar flexor moment as a result of the stiff footplate is also in agreement with that study\(^{[20]}\), and may be explained by the ground reaction force shifting forward early in stance. On the contrary, Fatone’s study\(^{[20]}\) showed a non-significant increase in the internal peak knee extensor moment in early stance while walking with a full-length footplate, compared to the three-quarter footplate. Yet, as they found that subtle changes in sagittal AFO-FC alignment had relatively less effect on the knee moments during stance compared to changes in the length of the footplate, Fatone et al.\(^{[20]}\) suggested that adjustments in footplate length should be used to control the knee joint moments during stance. The interaction effect of heel height and footplate stiffness on the internal knee extensor moment found in our study, emphasizes the importance of considering footplate characteristics within AFO-FC tuning. In this context, tuning using footplate stiffness characteristics should however preferably be done using the ground reaction force, as the stiff footplate showed no effect on SVA.
A limitation of the study is the calculation of the SVA, which was expressed in the global sagittal plane. Although the SVA was calculated according to methods used in other research and in clinical (2D) settings, it may have introduced a small underestimation of the SVA. Another limitation is poor fitting of the AFOs to some subjects, enabling compensation to the AFO-FC manipulations. This is supported by the results on the Shank-to-AFO angle, which decreased with increasing heel height, indicating that the lower leg was pushed more into the dorsal shell of the AFO when heel height increased. Moreover, the AFO may have been lifted inside the shoe, therewith affecting joint flexion-extension angles and moments.

Our results indicate that the SVA is responsive to AFO-FC heel height manipulations in young healthy adults walking with bilateral rigid AFOs. An increase in SVA was accompanied by increased joint flexion angles and internal net extension moments, especially at the knee joint. Whereas the SVA was not responsive to changes in footplate stiffness, the stiff footplate increased the internal ankle plantar flexion moment, and an interaction effect of heel height and footplate stiffness showed an opposite effect of the stiff footplate on the internal knee extensor moment in the low and high heel height conditions. These findings emphasize the consideration of footplate characteristics in the tuning process. In conclusion, the SVA may serve as a parameter to evaluate AFO-FC tuning, which has to be elaborated on in the clinical target population.
REFERENCES


The shank-to-vertical angle