The Shank Angle to Vertical as a control parameter for tuning of Ankle Foot Orthoses

Vrije Universiteit
Faculty of Human Movement Sciences
Qualification: MSc in Human Movement Sciences: Sport, exercise and health
Research internship
Authors: M.L.C. Kessels
A. Sterk
Supervisors: H. Houdijk, PhD (VU)
Y.L. Kerkum, MSc (VUmc)
Prof. dr. ir. J. Harlaar, PhD(VU)
F. Steenbrink, PhD (MOTEK Medical)
S. Roeles, MSc (MOTEK Medical)
October 2013
Abstract

Introduction
The effect of Ankle Foot Orthoses Footwear Combinations (AFO-FCs) in children with Cerebral Palsy (CP) seems to be inconclusive and might be dependent on its tuning. Analysis of joint kinematics and kinetics of the lower limb is required for tuning, this is, however, time-consuming. The Shank Angle to Vertical (SAV) is a candidate control parameter which can be easily measured and can control joint kinematics and kinetics. However, the influence of adjusting the SAV on kinematics and kinetics of the ankle, knee and hip during gait is unknown.

Objective
This study investigated what the effects are of various heel heights and footplate stiffness of an AFO-FC on the SAV during gait. Furthermore, the influence of these manipulations on joint angles and moments of the ankle, knee and hip in mid stance and the reliability of the SAV was examined.

Methods
Ten healthy participants wore bilateral orthoses and walked at a self-selected speed on the Gait Real-Time Analysis Interactive Lab (GRAIL) (Motek Medical, Amsterdam, the Netherlands). The SAV was manipulated by heel height (low, medium and high heel height) and footplate stiffness (flexible, rigid). In all the trials, the SAV, joint angles and net internal flexion moments were measured during walking for a period of two minutes on the GRAIL.

Results
The main effect was on the joint angles and moments of the knee. The SAV, flexion angles of ankle, knee and hip and net internal knee flexion moment in mid stance increased significantly with increasing heel height. For the footplate, significant effects were only found on the ankle flexion angle in mid stance. The standard deviations and reproducibility of the SAV show a reliable SAV when measuring 15 steps.

Conclusions
The SAV can be influenced using heel height, but footplate stiffness had no effect on the SAV in mid stance. An exact relationship between the SAV and knee flexion angle in mid stance was not determined and requires further investigation before the SAV can serve as a control parameter for AFO-FC tuning. At least 15 steps must be recorded in order to generate a reliable SAV indication.

Keywords: Ankle Foot Orthoses Footwear Combination, Shank Angle to Vertical, tuning, joint angles and moments.
1. Introduction

In children with cerebral palsy (CP), the central nervous system has been damaged due to lesions or anomalies in the developing foetal or infant brain\(^1\). Patients with CP can develop abnormalities such as a loss of selective muscle control, abnormal muscle tone and imbalance between muscle agonists and antagonists across joints which can lead to abnormal kinematics and kinetics of knee, hip and trunk. This is expressed in gait deviations such as abnormal knee flexion in mid stance of the gait cycle, which may increase the walking energy cost and decreases walking velocity and stability\(^2,3\). Ankle-Foot Orthoses (AFOs) are widely used to improve gait in patients with CP\(^2\). AFOs provide direct control of the foot and ankle and indirect control of the knee and hip by which they can compensate for a loss of function or counteract an excess of function\(^4,5\). An AFO aims to optimise the alignment of the ground reaction force (GRF) (i.e. aligning it closer to the joints) in order to normalise the joint kinematics and kinetics as well as spatiotemporal gait parameters which have been shown to be related to a reduction in walking energy cost\(^4,6,7\).

In literature, the evidence to support AFOs is varying and inconclusive\(^4\). The efficacy of AFOs might be partly dependent on its tuning\(^2,4\). Tuning is the adjustment of AFO footwear combinations (AFO-FCs), aiming to optimise the alignment of the GRF in order to normalise joint kinematics and kinetics of the lower limb during gait\(^2\). However, analysis of joint kinematics and kinetics is extensive and time-consuming. A clear parameter which can be easily measured and can control joint kinematics and kinetics might reduce the time costs of AFO-FC tuning. A candidate control parameter is the ‘Shank Angle to Vertical’ (SAV).

![Shank Angle to Vertical](Figure 1 Shank Angle to Vertical\(^7\))

The SAV is the angle of the shank relative to the vertical in the sagittal plane (Figure 1). According to Owen\(^6,9\) an SAV of 10 – 12 degrees in mid stance results in an optimal alignment of the GRF to the hip and the knee while wearing an AFO-FC, which may result in a more efficient gait pattern in terms of walking energy cost\(^6,9\). Owen manipulated the SAV by applying different heel heights in the AFO-FC with a fixed ankle angle\(^9\). Furthermore, in clinical practise it is seen that the stiffness of the AFO-footplate influenced the point of application of the GRF involving a shift to anterior with respect to the knee with increasing footplate stiffness. The effect of these manipulations on SAV are only measured in
bench alignment and not recorded during walking. Hence, the sensitivity of the SAV during walking is unknown and the effects on joint kinematics and kinetics during walking remain unclear. To determine whether the SAV can serve as a simple control parameter for AFO-FC tuning, it might be important to evaluate the effects of manipulating the SAV on joint kinematics and kinetics of the lower limb during gait\textsuperscript{9,10}.

The aim of this study was to investigate whether the SAV in mid stance can serve as a control parameter in AFO-FC tuning. Therefore, it was first investigated what the effects are of various heel heights and footplate stiffness of an AFO-FC on the SAV during gait. Secondly, the influence of these manipulations on joint kinematics (i.e. joint angles) and joint kinetics (i.e. joint moments) of the lower limb was examined. Thirdly, the reliability of measuring the SAV was investigated. In this study it was hypothesized that the effects of the manipulations are mainly on the joint angles and moments of the knee while less effect was expected on the angles and moments of ankle and hip.
2. Methods

2.1 Participants

After giving their written informed consent, ten healthy adults (age: 22.6 ± 2.1 years, height: 1.75 ± 0.09 m, weight: 66.2 ± 9.0 kg) participated in this study. Their shoe size had to be between 38 and 44 to fit the available experimental AFOs.

2.2 Material

The participants wore bilateral AFOs and simple sneakers that were supplied by the researchers. The AFO had a fixed ankle angle of 0 degrees and was made of prepreg carbon. To vary the heel height, different heights of heel insole wedges, made of cork, were added to the AFO-FC. The ‘Vertical Inclinometer on a Rail’ (VICTOR) (VU Medical Centre, Amsterdam, the Netherlands) was used to determine the height of the insole wedges that was required to achieve an imposed SAV of 5 degrees (low heel height), 11 degrees (medium heel height) and 20 degrees (high heel height) of the AFO-FC (Table 1). The AFO-footplate stiffness was quantified as flexible. To manipulate the footplate stiffness, a rigid footplate, made of dynima, was placed under the AFO in the sneaker. The mechanical characteristics of the AFO and the footplates were measured with the ‘Bi-articular Reciprocating Universal Compliance Estimator’ (BRUCE) (VU Medical Centre, Amsterdam, the Netherlands) (Table 2).

<table>
<thead>
<tr>
<th>Table 1 Height of heel insole wedges</th>
</tr>
</thead>
<tbody>
<tr>
<td>Heel height [cm]</td>
</tr>
<tr>
<td>Low</td>
</tr>
<tr>
<td>Medium</td>
</tr>
<tr>
<td>High</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Table 2 Stiffness of the Ankle Foot Orthosis (AFO)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiffness [Nm/deg]</td>
</tr>
<tr>
<td>AFO – Ankle</td>
</tr>
<tr>
<td>AFO – Footplate</td>
</tr>
<tr>
<td>Insole footplate</td>
</tr>
</tbody>
</table>
To be able to measure joint angles and moments, the Gait Real-time Analysis Interactive Lab (GRAIL) (Motek Medical, Amsterdam, the Netherlands) was used. The GRAIL is a system which can be used for extensive gait analysis and consists of a dual-belt treadmill with an integrated force plate (ForceLink BV, Culemborg, the Netherlands), and a motion capture system (VICON, Oxford Metrics, Oxford, United Kingdom) with ten cameras. D-flow software was used to control the hardware components and to collect kinematic and kinetic data sampled with a variable frame rate of around 130 Hz.

2.3 Procedure

Participants were first acquainted with the GRAIL for two minutes while wearing the shoes (provided by the researchers) without the AFOs. Afterwards, they walked on the GRAIL for two minutes wearing the AFO-FC in which no manipulations had been added. At the end of this familiarisation period, the individual comfortable walking speed was determined. Passive reflective markers were placed on the skin at defined anatomical and technical landmarks on the participant’s body according to the model by Van den Bogert\textsuperscript{12}. On each leg, two additional markers were placed on the tuberositas tibiae (TUT-marker) and distal on the margo anterior of the tibia (DIT-marker).

Measurements consisted of seven trials walking on the GRAIL for two minutes at the preselected, comfortable walking speed. Firstly, a baseline trial was performed wearing the shoes without AFOs. This was necessary to calibrate the segments of the human body model\textsuperscript{12} for the calculation of joint angles and moments. For the remaining walking conditions, there was a randomisation for the sequence of the trials. The conditions involved walking with a low, medium and high heel height combined with a flexible footplate or combined with a rigid footplate. The GRF and position of the markers were continuously recorded during walking using D-flow.

2.4 Data analysis

All data analysis was collected and processed in D-flow. Only the right leg in mid stance was analysed in this study. Mid stance was defined as the moment where the distance between the left and right malleoli marker of the ankle in the anterior-posterior direction was minimal. The SAV was calculated in the sagittal plane as the angle between the line between DIT-marker and POT-marker and a vertical reference, as shown in figure 2. The calculation of joint flexion angles and net internal flexion moments of the ankle, knee and hip was based on the model of Van den Bogert\textsuperscript{12}. 


Figure 2 To calculate the Shank Angle to Vertical during gait, additional markers were placed on the shank (A) and allowing a triangle to be created (B).

2.5 Statistics

The SAV, flexion angles of ankle, knee and hip and net internal flexion moments of ankle, knee and hip in mid stance of the last 30 seconds of each trial were averaged. Statistical analyses were performed using SPSS, version 20. The effect of heel height (low, medium, high) and footplate (flexible, rigid) on SAV and joint angles and moments of the ankle, knee and hip in mid stance was examined with a two-way repeated measures analysis of variance (ANOVA). Differences were determined using Bonferroni’s post hoc test. The statistical significance level was set at \( p < 0.05 \). The reproducibility of the SAV in mid stance was evaluated by the intraclass correlation coefficient (ICC) (3,1) using the SAV-data of 15 single steps. The conditions of the flexible footplate with low, medium and high heel height were used. ICC values were interpreted with a rating > 0.75 excellent agreement, 0.40 – 0.75 fair to good agreement and < 0.40 poor agreement\(^\text{13}\). Furthermore, standard deviations of the SAV for all trials were analysed.
3. Results

Ten healthy participants walked with a mean comfortable walking speed of 0.93 ± 0.06 m/s. The average number of recorded steps during the last 30 seconds was 23.93 ± 1.96 over all trials. On average for all participants the defined mid stance was at 29% of the gait cycle. Due to technical problems, only the kinetic data of six participants could be used for analysis.

3.1 SAV

The mean measured SAV in mid stance increased with increasing heel height \((F(1.15, 10.35) = 50.84, p < 0.001, \eta_p^2 = 0.85)\)(Figure 3). Post hoc analysis revealed significant differences between each heel height \((p < 0.01)\). However, the low heel height, with an imposed SAV of 5 degrees, did not result in an SAV of 5 degrees in mid stance but was on average 8.97 degrees \((\pm 1.35)\). The same applied for the high heel height, with an imposed SAV of 20 degrees, which led to a mean measured SAV of 17.22 degrees \((\pm 4.26)\). Only the medium heel height, with an imposed SAV of 11 degrees, resulted in a mean SAV of 11.04 degrees \((\pm 1.90)\) in mid stance. The footplate had no significant main effect on the SAV in mid stance \((p > 0.05)\) and there was no interaction effect of heel height and footplate\((p > 0.05)\).

![Figure 3 The Shank Angle to Vertical measured in mid stance. Blue and red bar indicate flexible and rigid stiffness of the footplate respectively.](image)

3.2 Ankle kinematics and kinetics

Despite the fixed ankle angle of the AFO, the mean ankle angle in mid stance increased with increasing heel height \((F(2, 18) = 29.02, p < 0.001, \eta_p^2 = 0.76)\)(Figure 4a). The Bonferroni pair-wise comparisons showed significant differences between the low and high heel height \((\text{mean difference} = 7.26 \text{ degrees}, \text{SE} = 1.20)\) and between the medium and high heel height \((\text{mean difference} = 5.36 \text{ degrees}, \text{SE} = 0.81)\). The ankle angle in mid stance was also influenced by the footplate \((F(1,9) = 8.32, p < 0.05, \eta_p^2 = 0.48)\) Post hoc analysis revealed that participants had a significantly smaller ankle angle
in mid stance with the rigid footplate compared with the flexible footplate (mean difference = 1.84, SE = 0.64) (p < 0.05). There was no significant interaction effect of heel height and footplate (p > 0.05). The heel height and footplate had no effect on the moments of the ankle in mid stance (p > 0.05) (Figure 4d).

3.3 Knee kinematics and kinetics

The mean knee flexion angle in mid stance increased with increasing heel height (F(2,18) = 91.78, p < 0.001, $\eta^2_p = 0.91$) (Figure 4b). The Bonferroni pair-wise comparisons showed significant differences between each heel height (p < 0.001). The mean difference between low and medium heel height was 6.20 degrees (SE = 0.77) and between medium and high heel height was 9.37 degrees (SE = 1.25). Effect of the footplate showed a trend (p = 0.087) indicating that the rigid footplate reduces the knee angle in mid stance compared with the flexible footplate (mean difference = 1.28 degrees, SE = 0.67). The results showed an interaction effect of heel height and footplate (F(2,18) = 4.82, p < 0.05, $\eta^2_p = 0.35$). Only the high heel height showed a significant difference between the flexible and stiff footplate (mean difference = 2.91, SE = 0.89), (t(9) = 3.28, p = 0.009, r = 0.74, CI [0.91, 4.91]) indicating the footplate only had an effect for a high heel sole differential. As with the knee angle, the mean knee flexion moments in mid stance increased significantly with increasing heel height (F(2,10) = 30.76, p < 0.001, $\eta^2_p = 0.79$) (Figure 4e). The Bonferroni pair-wise comparisons showed significant differences between each heel height (p < 0.001). The mean difference between low and medium was 0.15 Nm/kg (SE = 0.02) and between medium and high was 0.26 Nm/kg (SE = 0.04). There was no significant main effect for footplate (p > 0.05) and no significant interaction effect of heel height and footplate (p > 0.05) on the knee moment in mid stance.

3.4 Hip kinematics and kinetics

The mean hip flexion angle in mid stance was affected by heel height (F(2,18) = 21.15, p < 0.001, $\eta^2_p = 0.70$) (Figure 4c). The absolute differences in hip angle in mid stance are small but nevertheless post-hoc analysis revealed that they are significantly different from each other (p < 0.05). The mean difference between low and medium heel height was 2.23 degrees (SE = 0.69) and between medium and high heel height was 2.22 degrees (SE = 0.44). There was no significant main effect of the footplate (p > 0.05) and no significant interaction effect of heel height and footplate on the hip angle in mid stance (p > 0.05). The heel height and footplate did not affect hip moments in mid stance (p > 0.05) (Figure 4f).
Figure 4 Mean joint angles (a, b, c) and moment (d, e, f) of ankle, knee and hip in mid stance for different heel heights. Blue and red bar indicate flexible and rigid footplate stiffness respectively.

In table 3, the results of the standard deviations and the reproducibility of the SAV are shown. The standard deviations indicate a high variability in measurements of the SAV. The results of the ICC indicate an ‘excellent’ agreement for SAV in mid stance in the medium and high heel height conditions and a ‘fair to good’ agreement for the SAV in mid stance in the low heel height condition.

Table 3 Standard deviations and intraclass correlation coefficient (3,1) of the measured Shank Angle to Vertical at low, medium and high heel height.

<table>
<thead>
<tr>
<th>Heel height</th>
<th>Low</th>
<th>Medium</th>
<th>High</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard deviation [deg]</td>
<td>1.05</td>
<td>1.14</td>
<td>1.80</td>
</tr>
<tr>
<td>ICC (3,1)</td>
<td>0.66</td>
<td>0.79</td>
<td>0.89</td>
</tr>
</tbody>
</table>
4. Discussion

Aiming to investigate the SAV as a control parameter, it was examined what the effect is of varying the heel height and footplate stiffness of the AFO-FC on the SAV and joint angles and moments of ankle, knee and hip in mid stance. Furthermore, the reliability of the SAV in mid stance was analysed. As expected, the SAV in mid stance increased with increasing heel height. Contrary to our expectations, the footplate stiffness had no effect on the SAV in mid stance. Analysis of the joint angles of the lower limb showed that most effects were seen on the knee. The effect on ankle angle in mid stance was striking, where there was a significant effect for both heel height and footplate stiffness on the ankle angle.

As expected, the SAV in mid stance increased with increasing heel height. There were clear differences in SAV between the low, medium and high heel height. Using VICTOR, the heel height required to achieve an imposed SAV of 5°, 11° and 20° in bench alignment was determined. There appears to be a difference between the bench aligned SAV and the SAV in mid stance (Figure 3). The results indicate that participants prefer a walking strategy whereby their SAV in mid stance is normalised in the direction of the optimal values as reported by Owen (2002)⁸. For this strategy, ankle range of motion (ROM) is required. The results of measurements with BRUCE indicated that the stiffness of the AFO was appropriate to limit ankle ROM. There was a reduction from 26.8 degrees ankle ROM in the only shoes condition to 12.5 degrees in walking with AFO-FCs. The participants used this residual ankle ROM to influence their SAV in mid stance as walking strategy. Due to ankle ROM, the standard deviations of the SAV are relatively high which indicate a high step-to-step variability. However, the ICC results (determined over 15 steps) indicate that the SAV is well reproducible. Therefore it is preferable to measure the SAV over several steps. Taking this into consideration, the SAV is sensitive to varying heel height and has to be recorded over at least 15 steps in order to obtain a reliable SAV indication.

As well as the SAV in mid stance, the knee flexion angle in mid stance increased with increasing heel height. This is in accordance with Jagadamma et al.¹⁰ (2010), who manipulated the SAV of an AFO-FC with heel insole wedges in bench alignment and reported an increase in the knee flexion angle in mid stance. They did not measure the SAV during walking but assumed that the bench aligned SAV results in the same SAV during mid stance¹⁰. Our study showed that with increasing heel height, the SAV increased but also the knee flexion angle. It can be suggested that an increasing SAV leads to an increase in knee flexion angle and therefore, we suggest, that the SAV can be used to control the knee flexion angle in mid stance. However, based on this study, no statements can be made about the exact relationship between the SAV and knee angle in mid stance. Besides the effects of the heel
height, there was a trend \((p = 0.087)\) of the footplate stiffness on the knee flexion angle indicating that the rigid footplate decreased the knee flexion angle in mid stance compared with the flexible footplate. Besides the agreements of the heel insole wedges, Jagadamma et al. used a rigid rocker to tune the knee angle during gait which can be compared to our rigid footplate because in both situations the roll-over of the foot is limited. In accordance with Jagadamma’s study, we did not find an effect of the rigid footplate on the knee flexion angle in mid stance. However, Jagadamma et al. found an effect of the footplate in terminal stance and pre-swing\(^{10}\). These parts of the gait cycle were not analysed in this study so therefore these results cannot be confirmed or rejected. Taken generally for the knee flexion angle in mid stance, patients who walk with extended knees in mid stance can use heel insole wedges to increase their knee flexion angle. On the other hand, patients who walk with increased knee flexion angle in mid stance can use a rigid footplate to normalise their knee angle. Although there were no significant results of the footplate, footplate stiffness can be used to achieve subtle changes in knee angle.

Contrary to our expectations, there was an effect of the heel height and the footplate stiffness on the ankle angle. This could be due to the fact that markers for calculating the ankle angle were placed on both the shoe and the participant. The markers were not repositioned during the experiment. This is important as far as the marker on the lateral malleolus is concerned because with increasing heel height, this marker became less in line with the actual position of the lateral malleolus which is required for reliable ankle kinematics. As a result, the marker on the lateral epicondyle of the knee assumes a more anterior position with increasing heel height with respect to the lateral malleolus marker. This leads to overestimation of the ankle angle in mid stance and explains the increase of the ankle angle with increasing heel height.

Our results showed that the hip flexion angle in mid stance increased with increasing heel height. The absolute difference in angles between the low and high heel height were small (Figure 4c). In the study by Maas et al.\(^{7}\) (2009), the SAV in mid stance was manipulated from 10 to 14 degrees using heel insole wedges in children with CP. They found no differences in hip angles and moments but they used more subtle changes in the SAV\(^{7}\). In this study more pronounced differences in heel height, and therefore in SAV in mid stance, were used. These pronounced differences in heel height only provide a small difference in hip flexion angle in mid stance. Therefore, it can be stated that the SAV has a minimal effect on the hip flexion angle in mid stance.

In the same way as with the joint angles, the main effect of heel height is on the joint moment of the knee meaning that the knee flexion moment increased with increasing heel height. There was no effect of the heel height on the flexion moments of the ankle and the hip. It is assumed that the
vector of the GRF is oriented towards the body centre of mass in walking with AFO-FCs as it is in normal walking. With increasing heel height, the SAV inclined but the vector of the GRF remained in the same direction (Figure 5) and consequently the GRF shifted more posterior to the knee resulting in an increased net external flexion moment. However, there was no effect on the moment of the hip because the hip stayed in approximately the same position relative to the upright direction of the GRF (Figure 5). In order to make this possible, the thigh must recline to compensate the increased knee flexion angle. In contrast to the varying heel height, there was no effect of the footplate stiffness on the joint moments. This could be due to the fact that in mid stance, the footplate stiffness did not change the direction of the GRF. In other gait events, such as terminal stance, the body centre of mass is not above the foot resulting in an effect on joint moments. This is confirmed by the study of Jagadamma et al.\textsuperscript{10} (2010) who found no effect on joint moments in mid stance but an effect in terminal stance and pre-swing by using a rigid footplate.

![Figure 5 Alignment of the ground reaction force with respect to the ankle, knee and hip for the flexible footplate. Low, medium and high represents the heel height.](image)

The conclusions of this study are subject to some limitations. Our constant walking velocity for all trials could be seen as a limitation for its implementation in a clinical setting. For clinical practise, it is relevant to study the effect on walking velocity in children with CP with the manipulations used. Nevertheless, this constant walking velocity enables changes in joint angles in mid stance to be detected. Jagadamma’s study\textsuperscript{10} and this present study demonstrate that the SAV in mid stance can be influenced by heel height resulting in an increase in knee flexion angle net moment. However, neither of both studies tested this in children with CP. A study such as performed by Maas et al.\textsuperscript{7} is required to confirm if the same applies for children with CP. However, more pronounced differences in heel heights will be required.
5. Conclusion

In this study, it has been proven that the SAV can be influenced using heel insole wedges but that footplate stiffness had no effect on the SAV in mid stance. The main effects of these manipulations were on the joint angles and moments of the knee. Based on the ICC values and standard deviations of the SAV in this study, at least 15 steps have to be recorded to obtain a reliable SAV indication. To determine whether the SAV can serve as a control parameter for AFO-FC tuning, the exact relationship between the SAV and joint angles and moments of the knee should be further examined.
6. References


