Published in final edited form as:

J Biomech. 2018 June 25; 75: 176–180. doi:10.1016/j.jbiomech.2018.04.034.

Effect of plantarflexion resistance of an ankle-foot orthosis on ankle and knee joint power during gait in individuals post-stroke

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Abstract

Plantarflexion resistance of an ankle-foot orthosis (AFO) plays an important role to prevent foot-drop, but its impact on push-off has not been well investigated in individuals post-stroke. The aim of this study was to investigate the effect of plantarflexion resistance of an articulated AFO on ankle and knee joint power of the limb wearing the AFO in individuals post-stroke. Gait analysis was performed on 10 individuals with chronic stroke using a Vicon 3-dimensional motion capture system and a Bertec split-belt instrumented treadmill. They walked on the treadmill under 4 plantarflexion resistance levels ($S1 < S2 < S3 < S4$) set on the AFO with resistance adjustable ankle joints. The ankle and knee joint power calculations were performed using Visual3D, and mean values were plotted across a gait cycle. Statistical analyses revealed significant differences in the peak ankle joint power generation according to the plantarflexion resistance of the AFO ($P = 0.008$). No significant differences were found in the knee joint power. Peak ankle joint power generation [Median (IQR: Interquartile range)] were S1: 0.0517 (0.0238 to 0.1071) W/kg, S2: 0.0342 (0.0132 to 0.0862) W/kg, S3: 0.0353 (0.0127 to 0.0821) W/kg, and S4: 0.0234 (0.0087 to 0.06764) W/kg. Reduction of the peak ankle joint power generation appeared to be related to reduction in the peak plantarflexion angular velocity at late stance due to increases in the plantarflexion resistance of the AFO. This study showed that peak ankle joint power generation

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Conflict of interest statement

Kobayashi T and Orendurff MS were employees of Orthocare Innovations and designed the articulated AFO used in this study.

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was significantly, and somewhat systematically, affected by plantarflexion resistance of the AFO in individuals post-stroke.

Keywords

AFO; gait; hemiplegia; orthotics; stiffness

Introduction

Ankle-foot orthosis (AFO) can generally improve gait, mobility and balance in individuals post-stroke (Tyson and Kent, 2013; Tyson et al., 2013). Improvement in gait with the use of an AFO has been reported in tempo-spatial parameters (Gok et al., 2003; Hesse et al., 1999), ankle or knee joint angles (Nikamp et al., 2017; Yamamoto et al., 2011), ankle or knee joint moments (Kobayashi et al., 2015; Yamamoto et al., 2015) and energy expenditure (Hyun et al., 2015). Even though the AFO has these positive effects on gait, optimization of its resistance is based on trial-and-error in the clinical setting, not on evidence. Therefore, it is necessary to investigate how incremental changes of the AFO resistance affect biomechanical parameters of gait to establish evidence-based optimization process of the AFO in individuals post-stroke.

Plantarflexion resistance of an AFO assists impaired function of dorsiflexors in individuals post-stroke during 1st rocker of stance. This is accomplished by providing sufficient braking for achieving heel strike at initial contact and controlled plantarflexion at loading response (Yamamoto et al., 2011). In addition, during swing it maintains the position of the ankle joint for toe clearance. However, plantarflexion resistance would also brake the ankle joint moving from the AFO's neutral position to the plantarflexion direction during 3rd rocker of stance for push-off. It is generally known that reduction in ankle plantarflexion results in decreases in peak ankle power generation in healthy adults (Huang et al., 2015), which suggests less push-off in gait. And weakness of plantarflexors is known as one factor that limits gait speed in individuals post-stroke (Nadeau et al., 1999).

To the best of our knowledge, no studies have systematically investigated the effect of plantarflexion resistance of an AFO on ankle and knee joint power by incrementally changing the plantarflexion resistance in individuals post-stroke. Therefore, the aim of this study was to investigate the effect of plantarflexion resistance of an articulated AFO on ankle and knee joint power in individuals post-stroke. From our previous work, we showed that plantarflexion resistance of the AFO had systematic effects on ankle and knee joint angle and moment during gait (Kobayashi et al., 2015). Increases in the plantarflexion resistance of the AFO resulted in decreases in the peak plantarflexion angles of the ankle joint and in the peak extension angles in the knee joint. Because joint power is a product of joint moment and joint angular velocity, we hypothesized that ankle and knee joint power would be significantly affected by incrementally changing the plantarflexion resistance of the AFO.

Methods

Gait analysis was performed on 10 individuals with chronic stroke (2 females/8 males; Mean (SD): 56 (11) years old; body mass: 99(17) kg; body height: 1.76(0.11) m). Six individuals with stroke had right side involvement. The outcome of clinical evaluations for these individuals were as follows (Kobayashi et al., 2015): 1) The Modified Ashworth Scale (MAS) of the ankle joint ranged from 1+ to 3; 2) The mean (SD) of Timed-up and Go Test (TUG) was 17.72 (5.13) seconds; 3) The Manual Muscle Testing (MMT) of the ankle dorsiflexors ranged from 0 to 4 and the plantarflexors from 0 to 4, the MMT of the knee flexors ranged from 2+ to 5 and the extensors from 3+ to 5; 4) The mean (SD) of manual passive peak dorsiflexion angle while the knee joint kept in extension was 2 (6) degrees of plantarflexion. This study was approved by the Institutional Review Board at the University of Utah.

Reflective markers were placed on the head, trunk and limbs of the participants based on a modified Cleveland Clinic Marker Set defining 8 segments [2 feet, 2 shanks, 2 thighs, 1 pelvis, and 1 HAT (head, arm, and trunk)]. The markers were placed directly on the AFO, and a rigid cluster with four markers was secured to the upright of the AFO and used for dynamic tracking. Each participant was secured in a safety harness and asked to walk on a split-belt instrumented treadmill (ITC-11-20L/ITC-11-20R, Bertec corporation, Columbus, OH, USA). An AFO with a plantarflexion resistance adjustable joint was used for this investigation (Figure 1) (Kobayashi et al., 2015). For all trials, the participants walked at a self-selected walking speed (range: 0.15 – 0.27 m/s) with the AFO under four different spring levels: S1, S2, S3 and S4 (Figure 1) in a randomized order. Three participants chose to use a cane on their non-impaired side during data collection as a precaution because they were unfamiliar with the wide range of plantarflexion resistance settings being tested.

The AFO that was worn was designed with an ankle joint that provided plantarflexion resistance using a steel spring at different spring rates. The joint did not generate resistive moment toward dorsiflexion. No steel spring was set on the AFO under S1, and it was considered as a baseline condition/measurement. The initial baseline adjustment (*i.e.* alignment and heel height) for the AFO was used for all trials. The only change made to the AFO between each trial was the plantarflexion resistance by changing the spring level.

A short acclimatization period to walking on the treadmill was provided to each participant before data collection. Marker trajectory data were collected using a 10-camera VICON motion analysis system (Vicon Motion Systems, Oxford, UK) and kinetic data were collected using the Bertec Fully Instrumented Treadmill. Data were synchronized and recorded using VICON Nexus 2.0 software at a rate of 200Hz for a total of 5 successful steps per participant. Seated rests were given whenever necessary during the data collection.

Data were post-processed using Visual3D (C-Motion, Germantown, USA). A low pass, zero-phase shift Butterworth filter at 6 Hz and 20 Hz was used to filter marker trajectory and kinetic data, respectively based on visual inspection and residual analysis (Winter, 2005). The ankle and knee joint power and angular velocity of 5 steps of the limb wearing the AFO were averaged and normalized to a gait cycle for each spring level in each participant. The

ankle and knee joint power were also normalized to body mass. The mean ankle and knee joint power and angular velocity of the participants were plotted under each spring level (Figure 2).

Five ankle power related parameters were calculated and extracted including: 1. Peak ankle power generation (W/kg), 2. Percent gait cycle of the peak (%), 3. Positive work (i.e. area of positive power) (J/kg), 4. Negative work (i.e. area of negative power) (J/kg), and 5. Peak plantarflexion ankle angular velocity (deg/s) at late stance. Six knee power related parameters were also calculated and extracted including: 1. Peak knee power absorption (early stance) (W/kg), 2. Peak knee power generation (late stance) (W/kg), 3. Positive work (i.e. area of positive power) (J/kg), 4. Negative work (i.e. area of negative power) (J/kg), 5. Peak knee flexion angular velocity at early stance (deg/s), and 6. Peak knee flexion angular velocity at late stance (deg/s).

Statistical analyses were performed to compare the five ankle power and the six knee power related parameters, respectively under each spring level of the AFO. Friedman test was performed because a Kolmogorov-Smirnov test revealed that some parameters were not normally distributed. Post-hoc analysis with Wilcoxon signed-rank tests was performed to compare S1 (baseline condition) with other conditions using Holms-Bonferroni method for multiple comparisons if the Friedman test showed a significant difference ($P < 0.05$).

Results

The mean ankle and knee joint power and angular velocity normalized to a gait cycle collected from the participants are shown in Figure 2. The outcome of the data analysis of each parameter in Median (IQR: Interquartile range) is summarized in Table 1. Statistically significant differences, according to the plantarflexion resistance of the AFO, were found at the ankle for peak ankle power generation (chi-square = 11.760, $P = 0.008$), positive work (chi-square = 12.760, $P = 0.006$) and peak plantarflexion angular velocity (chi-square = 14.760, $P = 0.002$). For the knee, negative work (chi-square = 9.240, $P = 0.026$) was found to be statistically significantly different. In addition, with increases in the plantarflexion resistance, at the ankle a reduction in the peak ankle power generation, positive work of the ankle, and peak plantarflexion angular velocity was observed. At the knee, increases in the plantarflexion resistance of the AFO resulted in a reduction of negative work of the knee with increases in the plantarflexion resistance (Table 1).

Post-hoc analysis with Wilcoxon signed-rank test with the Holms-Bonferroni method for multiple comparisons showed statistically significant differences between S1 and S2 ($Z = -1.988$, $P = 0.047$), S1 and S3 ($Z = -2.395$, $P = 0.017$), S1 and S4 ($Z = -2.803$, $P = 0.005$) in the peak ankle power generation, between S1 and S2 ($Z = -2.497$, $P = 0.013$), S1 and S3 ($Z = -2.497$, $P = 0.013$), and S1 and S4 ($Z = -2.701$, $P = 0.007$) in the positive work of the ankle, between S1 and S2 ($Z = -2.293$, $P = 0.022$), S1 and S3 ($Z = -2.599$, $P = 0.009$), and S1 and S4 ($Z = -2.599$, $P = 0.009$) in the peak plantarflexion angular velocity, and between S1 and S4 ($Z = -2.497$, $P = 0.013$) in the negative work of the knee.

Discussion

The aim of this study was to investigate the effect of plantarflexion resistance of an articulated AFO on ankle and knee joint power in individuals post-stroke. This study showed that the peak ankle power generation, positive work of the ankle, peak plantarflexion angular velocity and negative work of the knee were significantly affected by plantarflexion resistance of the AFO in individuals post-stroke (Table 1, Figure 2). It appeared that the changes in the peak ankle power generation were related to changes in ankle angular velocity (*i.e.* kinematics) more than ankle moment (*i.e.* kinetics) because the ankle moment did not appear to show differences in late stance due to incremental changes of the plantarflexion resistance of the AFO (Kobayashi et al., 2015). Therefore, reduction of the peak ankle power generation seems to be related to reduction in the peak plantarflexion angular velocity due to increases in the plantarflexion resistance of the AFO. However, no significant differences were demonstrated in the knee power due to incremental changes in the plantarflexion resistance of the AFO. This is the first study that demonstrated the effects of the plantarflexion resistance of the AFO on the peak ankle power generation and its mechanism by systematically changing its plantarflexion resistance in individuals post-stroke.

Plantarflexion resistance of an AFO is necessary to achieve heel strike at initial contact and toe clearance during swing. However, reduction in the peak ankle power generation and the positive work with increased plantarflexion resistance suggests that it also limits push-off at the 3rd rocker. Figure 2 suggested that timing of the peak ankle power generation might be delayed as the plantarflexion resistance of the AFO is increased in individuals post-stroke, but no significant differences were found (Table 1). This study also showed that negative work of the ankle is much larger than positive work during gait in individuals post-stroke. This suggests that absorbed energy during 1st and 2nd rocker was not efficiently returned during 3rd rocker. The plantarflexion resistance that is needed for 1st rocker may result in diminished push-off and increases in metabolic cost to the individuals post-stroke (Farris et al., 2015).

Generally, peak ankle power generation of individuals post-stroke is lower compared to non-disabled individuals and the amount is associated with their functional levels (Jonkers et al., 2009). Individuals walking slowly and having minimal push-off power or plantarflexion range of motion might be less affected during the 3rd rocker by the plantarflexion resistance of the AFO than individuals walking faster with more push-off. Considering the participants' self-selected walking speed was slow in this study, future study should investigate the influence of difference in gait speed on the lower limb joint power while wearing an AFO.

Use of an AFO may hinder ankle plantarflexion, but an AFO with plantarflexion resistance joints (*i.e.* oil-damper joints) was demonstrated to increase the peak ankle power generation compared to non-AFO use (*i.e.* shod condition) in individuals post-stroke (Yamamoto et al., 2015). This may be attributed to the increase in the ankle plantarflexor moment with the AFO, rather than the influence of ankle angular velocity. Therefore, use of an AFO itself may not necessarily have negative impacts on the 3rd rocker of stance compared to gait without an AFO, but the outcome of our study also suggested that it would be necessary to

find the balance between enough plantarflexion resistance for 1st rocker, while not hindering push-off for 3rd rocker more than necessary.

In order to maximize the peak ankle power generation for push-off at the 3rd rocker, it is necessary to find the minimum amount of plantarflexion resistance of an AFO that can optimize heel strike at initial contact, plantarflexion at loading response and toe clearance during swing. This tradeoff relationship requires further investigation and exploration of optimal gait of individuals post-stroke with an AFO. An AFO that can provide sufficient plantarflexion resistance at 1st rocker of stance and swing, but eliminate the resistance at 3rd rocker or provide positive push-off power may be clinically ideal. A powered AFO that can actively plantarflex for push-off has been developed (Sawicki and Ferris, 2009), but it is not designed for everyday use.

Limitations of this study include: 1) a shod condition (*i.e.* non-AFO condition) was not tested; 2) no electromyography (EMG) was measured thus muscle activation pattern and their contribution were unclear; 3) data were collected on the split treadmill, which might have affected the gait pattern compared to overground gait; and 4) The outcome of the clinical evaluations suggests that this group of subjects were not homogeneous and may not represent the general population of individuals with chronic stroke. However, demonstrating significant effects in this non-homogeneous group of subjects suggest that the findings of this study may be extrapolated to chronic stroke individuals with a wide range of functional levels.

Acknowledgments

This study was supported by the National Institutes of Health, specifically the Eunice Kennedy Shriver National Institute of Child Health & Human Development (2R44HD069095) and the National Center for Research Resources (S10RR026565).

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Figure 1.
A plantarflexion resistance adjustable AFO and its mechanical characteristics (Kobayashi et al., 2015).

Figure 2.

(a) Mean ankle joint power, (b) mean ankle angular velocity, (c) mean knee joint power, and (d) mean knee joint angular velocity of the limb wearing the AFO in a gait cycle collected from the participants post-stroke.

Abbreviations: DF, dorsiflexion; EX, extension; FX, flexion; PF, plantarflexion

Table 1

Ankle and knee joint power related parameters of the affected limb [Median (IQR)] under each AFO condition collected from the participants post-stroke. An asterisk (*) indicates a significant difference ($P < 0.05$) from S1.

<i>Ankle</i>	Ankle power generation (W/kg)	% GC of peak (%)	Positive work (J/kg)	Negative work (J/kg)	Ankle PF angular velocity (deg/s)
S1	0.0517 (0.0230 to 0.1071)	65 (61 to 67)	0.0023 (0.0013 to 0.0081)	-0.0356 (-0.0566 to -0.0238)	-36.10 (-46.80 to -30.21)
S2	0.0342 * (0.0132 to 0.0862)	65 (62 to 66)	0.0015 * (0.0005 to 0.0067)	-0.0329 (-0.0518 to -0.0254)	-32.47 * (-37.27 to -21.48)
S3	0.0353 * (0.0127 to 0.0821)	66 (63 to 68)	0.0020 * (0.0004 to 0.0046)	-0.0426 (-0.0499 to -0.0235)	-24.63 * (-32.45 to -9.48)
S4	0.0234 * (0.0087 to 0.06764)	66 (61 to 68)	0.0007 * (0.0003 to 0.0044)	-0.0378 (-0.0447 to -0.0233)	-18.56 * (-28.86 to -13.94)

<i>Knee</i>	Knee power absorption (early stance) (W/kg)	Knee power generation (late stance) (W/kg)	Positive work (J/kg)	Negative work (J/kg)	Knee FX angular velocity (early stance) (deg/s)	Knee FX angular velocity (late stance) (deg/s)
S1	-0.0764 (-0.2267 to -0.0332)	0.1520 (0.0101 to 0.2983)	0.0300 (0.0074 to 0.0496)	-0.0314 (-0.0394 to -0.0121)	20.32 (-1.34 to 55.01)	107.67 (77.17 to 118.79)
S2	-0.1309 (-0.1770 to -0.0530)	0.1339 (0.0169 to 0.2941)	0.0246 (0.0162 to 0.0445)	-0.0263 (-0.0388 to -0.0153)	33.36 (-10.08 to 50.58)	99.16 (70.07 to 115.31)
S3	-0.0814 (-0.2250 to -0.0305)	0.1024 (0.0292 to 0.1784)	0.0256 (0.0164 to 0.0361)	-0.0236 (-0.0275 to -0.0144)	25.81 (4.34 to 36.93)	95.81 (64.73 to 121.11)
S4	-0.0876 (-0.1656 to -0.0550)	0.0764 (0.0172 to 0.1597)	0.0255 (0.0176 to 0.0364)	-0.0182 * (-0.0303 to -0.0100)	23.37 (8.18 to 35.91)	104.88 (79.14 to 115.94)

Abbreviations: FX, flexion; GC, gait cycle; PF, plantarflexion