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The effects of an articulated ankle-foot orthosis with resistance-adjustable joints on lower limb joint kinematics and kinetics during gait in individuals post-stroke

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Abstract

Background: Resistance is a key mechanical property of an ankle-foot orthosis that affects gait in individuals post-stroke. Triple Action[®] joints allow independent adjustment of plantarflexion resistance and dorsiflexion resistance of an ankle-foot orthosis. Therefore, the aim of this study was to investigate the effects of incremental changes in dorsiflexion and plantarflexion resistance of an articulated ankle-foot orthosis with the Triple Action joints on lower limb joint kinematics and kinetics in individuals post-stroke during gait.

Methods: Gait analysis was performed on 10 individuals who were post-stroke under eight resistance settings (four plantarflexion and four dorsiflexion resistances) using the articulated ankle-foot orthosis. Kinematic and kinetic data of the lower limb joints were recorded while walking using a three-dimensional Vicon motion capture system and a Bertec split-belt instrumented treadmill.

Findings: Repeated measures analysis of variance revealed that adjustment of plantarflexion resistance had significant main effects on the ankle ($P < 0.001$) and knee ($P < 0.05$) angles at initial

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Conflict of interest statement

Kobayashi T, Orendurff MS and Lincoln LS were employees of Orthocare Innovations. LeCursi N works for Becker Orthopedic, manufacture of the AFO joint (Triple Action[®]) used in this study.

contact, while dorsiflexion resistance had significant ($P<0.01$) main effects on the peak dorsiflexion angle in stance. Plantarflexion and dorsiflexion resistance adjustments appeared to affect the peak knee flexor moment in stance, but no significant main effects were revealed ($P=0.10$). Adjustment of plantarflexion resistance also demonstrated significant ($P<0.05$) main effects in the peak ankle positive power in stance.

Interpretation: This study demonstrated that the adjustments of resistance in the ankle-foot orthosis with the Triple Action joints influenced ankle and knee kinematics in individuals post-stroke. Further work is necessary to investigate the long-term effects of the articulated ankle-foot orthoses on their gait.

Keywords

AFO; alignment; orthotic; stiffness; stroke; walk

Introduction

Individuals who are recovering from a stroke are often prescribed an ankle-foot orthosis (AFO) to improve stability in stance and prevent foot-drop in swing during walking (Tyson et al., 2013). The prescription of AFOs is not well founded in rigorous methodologies. While evidence for the biomechanical influence of AFOs on gait exists and is accumulating (Gatti et al., 2012; Nikamp et al., 2017b; Yamamoto et al., 2015; Zollo et al., 2015), there remains a paucity of evidence to support the efficacy of using certain orthotic designs to treat specific biomechanical deficits. Customary practice in the field of orthotics relies heavily upon the experience of the clinician and the practitioner's clinical intuition to specify the orthotic design, and to adjust that design to the specific needs of the patient.

The resistance generated from an AFO – the resistance to plantarflexion and the resistance to dorsiflexion – are key mechanical characteristics of an AFO that can influence kinematics and kinetics of the lower-limb joints during walking (Kobayashi et al., 2015; Yamamoto et al., 2013). Restricting pathological motion of the ankle during gait, such as excessive equinus in stance and foot-drop in swing, requires an AFO that is rigid enough to resist the plantarflexion force of the ankle. Resistance to plantarflexion also plays an important role in alleviating knee hyperextension in stance (Kobayashi et al., 2016). However, excessive resistance to plantarflexion may also exacerbate knee flexion during the first rocker of gait. Preventing excessive dorsiflexion and concomitant knee flexion in mid- and late-stance requires an AFO with sufficient rigidity to resist the dorsiflexion force elicited by gravity and possible weakness of the plantarflexor musculature. Therefore, the challenges presented to the clinician during the fitting and adjustment of AFOs centers on balancing the sometimes competing concerns of ankle and knee stability using only the tools available to the practicing orthotist. These tools include observational gait analysis along with objective and subjective clinical indicators.

The amount of resistance of an AFO is typically estimated by the clinician based on visual inspection of patient's gait and incorporated into the fabrication process prior to fitting. In other words, the definitive orthotic design must be determined, implemented and delivered before the overall impact of the orthotic design can be fully determined. Therefore, orthotic

adjustment is an essential aspect in the delivery of orthotic care. This adjustment process relies on both objective and subjective clinical indicators, i.e. patient feedback and is highly subjective and heavily dependent upon the expertise of the clinician.

Factors that may affect resistance of thermoplastic and composite AFOs during fabrication process include, but are not limited to: thermoforming method, layup, resin, trim lines, padding, lining and strapping. How each of these factors and their interactions affect final mechanical properties of AFOs is hard to predict. The shape of an AFO is described by its anatomical contour and trim lines. Trim lines define the bounding edges of the orthosis. Incorrect choice of resistance is difficult to correct once the AFO has been fabricated, especially if the device has been fabricated using composite materials. Adjustment of AFO trim lines is one means used to alter AFO resistance, but is irreversible, affects both resistance to plantarflexion and dorsiflexion, and may result in off-axis deformation or changes in support that can affect orthotic control of the ankle-foot complex during gait. One means to facilitate greater adjustability in the resistance of AFOs is to incorporate orthotic ankle joints. There are many ankle joint options available for AFOs, and choosing an adjustable ankle component is one approach to individually tune resistance in orthotic design.

Currently, several resistance-adjustable articulated AFOs are available in the clinical setting (Kerkum et al., 2015; Kobayashi et al., 2017; Yamamoto et al., 2005). These articulated AFOs may permit more anatomical movement of the ankle joint with proper resistance while walking compared to non-articulated AFOs. They also allow simple and convenient adjustment of AFO resistance, which is beneficial not only to the clinician but also to the patient because it can save time. Biomechanical benefits of adjusting resistance of articulated AFOs on gait in individuals post-stroke have been demonstrated (Kobayashi et al., 2015; Yamamoto et al., 2013), but evidence is still limited.

One common type of adjustable ankle joint is the traditional double upright metal AFO with dual channel components which allow the clinician to adjust the assistance and range of motion at the ankle. However, this type of ankle component typically lacks the capability to adjust and provide sufficient amount of resistance from their springs as well as adjustment of alignment, which are important parameters in AFO tuning. More recently, a Triple Action[®] ankle joint (Becker Orthopedic, Troy, USA) has been introduced which facilitates independent adjustment of plantarflexion resistance, dorsiflexion resistance and alignment of an AFO.

A recently published case study suggested that the ankle and knee angle and moment were responsive to changes in the AFO resistance settings when using the Triple Action joints and gait was improved in an individual post-stroke (Kobayashi et al., 2017). Though this case study showed promising results, it remains unclear if the kinematic improvements would be efficacious and beneficial to individuals post-stroke. Therefore, a larger scale study was warranted. The aim of the current study was to investigate the effects of incremental changes in dorsiflexion and plantarflexion resistance of an AFO with the Triple Action ankle joints on lower limb joint kinematics and kinetics while walking in 10 individuals post-stroke. We hypothesized that ankle, knee and hip kinematics and kinetics would be systematically

affected by changing the dorsiflexion and plantarflexion resistance of the AFO. Specifically, the ankle kinematics and kinetics would systematically change with changes to the amount of plantarflexion and dorsiflexion resistance. Knee and hip joint kinematics and kinetics would be impacted indirectly based on the direct effects of the plantarflexion and dorsiflexion resistance changes of the AFO at the ankle joint.

Methods

Participants

Ten individuals (3 females/7 males) post-stroke, whose mean (SD: standard deviation) time since stroke incidence was 5 (2) years, participated in this study (Table 1). Their mean (SD) age was 58 (13) years old, their mean (SD) body height was 1.74 (0.12) m, and mean (SD) body mass was 84 (20) kg. Each individual had unilateral limb involvement (5 right/5 left). Inclusion criteria of this study were: 1) a minimum of 6-month post-stroke with hemiparesis, and 2) an ability to walk safely on an instrumented level treadmill with the use of an AFO. Participants were excluded from the study if they had any confounding injury, musculoskeletal issues or cognitive issues that might limit their ability to walk on the treadmill. The following clinical assessments were performed on each individual post-stroke: 1) Measurement of manual passive peak dorsiflexion angle (i.e. range of motion: RoM) of the affected limb with the knee in 90° of flexion (Baumbach et al., 2014), 2) Manual muscle testing (MMT) of the ankle plantarflexors and dorsiflexors, knee extensors and flexors and hip extensors and flexors of the affected limb (Kendall et al., 1993). Five non-impaired individuals (1 female, 4 males) also participated as controls in this study. Their mean (SD) age was 36 (17) years old, the mean (SD) body height was 1.76 (0.05) m, and mean (SD) body mass was 72 (8) kg. Written informed consent was obtained from all participants in this study that was approved by the Institutional Review Board of the University of Utah (IRB_00062924).

Orthotic design

AFOs were custom fabricated for each individual post-stroke from 4.8 mm polypropylene homopolymer using two prototype Triple Action® ankle joints, one on each side (Fig. 1). The orthotic design was customized for each subject based on clinical assessment. Intrinsic orthotic supportive elements were designed to appropriately manage the posture of the foot and ankle and were customized to the needs of each individual participant. Where appropriate, lateral supramalleolar support was included in the orthosis to resist hindfoot varus with the aim of enhancing the sagittal plane control of the foot and ankle by resisting off-axis compensatory motion. The Triple Action joints allowed independent adjustments of AFO plantarflexion resistance, dorsiflexion resistance and alignment settings (Fig. 1). The resistance was adjustable by altering compression on the springs located in the plantarflexion and dorsiflexion channels. Resistance settings were adjusted by counting the number of turns of the resistance adjustment screws away from the locked position on top of the joint body. These adjustments changed the pre-load and range of motion of the joint in plantarflexion and dorsiflexion directions. Therefore, loosening the adjustment screws from the locked position decreased pre-load and increased range of motion. The alignment

adjustment rotated the joint body and stirrup around the pivot bushing, and was adjusted by turning the hex on the front of the joint body.

The mechanical properties (i.e. moment-angle relationship) of the AFO was quantified to demonstrate the effect of changing plantarflexion and dorsiflexion resistance settings of the joint using a custom motorized mechanical testing device (Gao et al., 2011). In this device, an optical encoder (Danaher motion Inc., Wood Dale, USA) was used to measure ankle joint angles, while an inline uniaxial torque sensor (TRT-500, measurement range: 500 in-lb or 56.5 Nm, Transducer Tech Inc., Temecula, USA) was used to measure resistive moment around the ankle joint. The moment-angle characteristics of the AFO were measured within a torque range of ± 30 Nm or an angle range of $\pm 15^\circ$. An NI PCI-6221 M Series DAQ was used for data acquisition. To be able to record, synchronize, and visualize the data as well as control the experimental settings, a graphic user interface was designed using LabVIEW (National Instrument Inc., Austin, USA). The effect of changing plantarflexion and dorsiflexion resistance settings were quantified from 0 to 4 turns with an increment of one half turn of the adjustment screw, respectively. The resistance to dorsiflexion and dorsiflexion angle were defined as positive in this study.

Gait analysis

The individuals post-stroke were fit with the custom-fabricated articulated AFOs by a certified orthotist, and adjustments were made to the orthoses using objective clinical indicators and subjective feedback from the participants. Objective clinical indicators included the observed position of the knee in quiet standing, toe clearance in mid swing, foot position at initial contact and observed knee kinematics in early and late stance as compared to the non-involved side. Subjective clinical indicators included primarily the participant's feedback pertaining to their sense of comfort and stability. Subsequently, the AFO was tuned using observational gait analysis and subjective feedback from the participants to establish the initial settings of the plantarflexion resistance, dorsiflexion resistance, and alignment of the AFO joints. All the participants wore the same shoes (New Balance 928, New Balance Inc., Boston, USA) with appropriate shoe size. The clinically-determined initial ankle joint settings served as baseline settings during the gait trials.

From these baseline settings, the plantarflexion resistance setting of the AFO joint was randomly changed by adjusting the plantarflexion resistance adjustment screw. Component settings were randomly changed by the certified orthotist to four distinct settings (P1 to P4) from the previously described clinically-determined settings for gait analysis (within the range of 0 turn and 4 turns of the plantarflexion resistance adjustment screw away from the locked position based on the available range of adjustability from the initial setting). Subsequently, the dorsiflexion resistance setting was randomly changed to four distinct levels of settings (D1 to D4) from the clinically-determined settings for gait analysis (within the range of 0 turn and 4 turns of the dorsiflexion resistance adjustment screw from the locked position based on the available range of adjustability from the initial setting). The initial settings and tested settings ("P1 to P4" and "D1 to D4") for each participant are shown in Table 2. The numbers in the table for plantarflexion and dorsiflexion resistance indicate the number of turns away from the locked position (i.e. 0 turn position). The

alignment of the AFO was maintained at constant angles (i.e. angles at clinically-determined initial ankle joint settings) throughout tested resistance settings in each participant.

Gait data were collected on a level split-belt instrumented treadmill (Bertec corporation, Columbus, USA) using a Vicon 10-camera motion analysis system (Vicon Motion Systems, Oxford, UK) at a rate of 200Hz. Reflective markers were placed on the feet, shanks, thighs, pelvis and trunk based on a modified Cleveland Clinic Marker Set defining 8 segments [2 feet, 2 shanks, 2 thighs, 1 pelvis, and 1 HAT (combined head, arms, and trunk)]. The markers for the shank and foot of the affected limb were placed directly on the AFO for the individuals post-stroke. A rigid cluster was secured to the lateral side of the AFO for dynamic tracking. A safety harness was used to secure the participants, and they were asked to walk on the level split-belt instrumented treadmill. The individuals post-stroke walked while wearing the AFO on the affected leg under each resistance conditions (“P1 to P4” and “D1 to D4”) at a self-selected speed of 0.21 – 0.36 m/s, while non-impaired controls walked under a shod condition at a self-selected speed of 0.85 – 1.08 m/s. The self-selected walking speed was determined through communication and feedback from each participant. The gait speed of the treadmill was kept constant across the resistance conditions for each individual post-stroke. One participant walked with a forearm crutch on the unaffected side, two participants walked placing their hand on a handrail of the treadmill for balance, and others walked without any assistance. The participants walked on the treadmill for a short period to acclimate before the data collection. This period was kept short to avoid fatigue during data collection.

Gait data were recorded and synchronized using Vicon Nexus software (Vicon Motion Systems, Oxford, UK) and post-processed using Visual3D (CMotion, Germantown, USA). Marker trajectory and force platform data were filtered with a low pass, zero-phase shift Butterworth filter at 6 Hz and 20 Hz, respectively. The ankle, knee and hip joint angles, moments and power were calculated by averaging a minimum of 8 gait cycles and the data was normalized to body mass for each of the resistance conditions of the AFO for the individuals post-stroke and the shod condition for non-impaired controls. The lower-limb with the AFO was analyzed for the individuals post-stroke, while right lower-limb was analyzed for the non-impaired controls. Subsequently, the mean of the 10 individuals post-stroke and 5 non-impaired controls, respectively, was calculated and plotted for the ankle, knee and hip joint angle, moment and power. Due to technical issues with the kinetic data, only the joint angle data were processed for the participant S07. Ankle dorsiflexion, knee flexion and hip flexion were defined as positive for the joint angles, while moments for ankle plantarflexion (ankle plantarflexor moment), knee extension (knee extensor moment) and hip extension (hip extensor moment) were defined as positive for the joint moments. Data were collected from non-impaired controls to show normal range of joint kinematics and kinetics and no comparisons were made between the two groups.

Statistical analysis

The following lower-limb joint kinematic and kinetic parameters were extracted from the gait data of the individuals post-stroke for statistical analyses: a) Ankle angle at initial contact (°), b) Peak dorsiflexion angle in stance (°), c) Peak dorsiflexor moment in stance

(Nm/kg), d) Peak plantarflexor moment in stance (Nm/kg), e) Peak ankle positive power in stance (W/kg), f) Knee angle at initial contact ($^{\circ}$), g) Peak knee extension angle in stance ($^{\circ}$), h) Peak knee flexor moment in stance (Nm/kg), i) Hip angle at initial contact ($^{\circ}$), j) Peak hip extension angle in stance ($^{\circ}$), k) Peak hip extensor moment in stance (Nm/kg), and l) Peak hip flexor moment (Nm/kg).

One-way repeated measures ANOVA was conducted to compare the lower limb joint kinematic and kinetic parameters listed in Table 3 among the four plantarflexion resistance settings (P1-P4) and the four dorsiflexion resistance settings (D1-D4), respectively. Adjustments were made if a violation of sphericity was found (Huynh-Feldt adjustment if the sphericity estimate >0.75 , Greenhouse-Geisser otherwise). Post-hoc multiple comparisons with Bonferroni adjustment were conducted if ANOVAs showed significant differences in the resistance settings. Partial eta squared (η_p^2) was reported as measures of effect size. Statistical analyses were conducted in SPSS v.19.0 (IBM Corp. Armonk, USA) and statistical significance level was set at $\alpha = 0.05$.

Results

Mechanical characteristics of the AFO

The moment-angle relationships of the AFO under each plantarflexion and dorsiflexion resistance setting are presented in Fig. 2. These data demonstrated the capabilities of the Triple Action joint to systematically change its moment-angle relationship by altering the plantarflexion and dorsiflexion resistance settings.

Effect of the AFO resistances on the ankle joints

Significant main effects were found in the ankle angle at initial contact ($F[1.19, 10.73] = 13.78, P < 0.001, \eta_p^2 = 0.61$) for the adjustment of the plantarflexion resistance settings and the peak dorsiflexion angle in stance ($F[3, 27] = 7.48, P < 0.01, \eta_p^2 = 0.45$) for the adjustment of dorsiflexion resistance settings (Fig. 3 and 4, Table 3). For the ankle angle at initial contact, post-hoc multiple comparisons showed significant differences between P1 and P3 ($P < 0.05$), P1 and P4 ($P < 0.05$), P2 and P4 ($P < 0.05$), P3 and P4 ($P < 0.05$). For the peak dorsiflexion angle in stance, post-hoc multiple comparisons showed significant differences between D1 and D3 ($P < 0.05$), D1 and D4 ($P < 0.01$). Significant main effects were found in the peak ankle positive power in stance ($F[3, 24] = 3.34, P < 0.05, \eta_p^2 = 0.30$) for the adjustment of the plantarflexion resistance settings, but the post-hoc multiple comparisons did not show significant differences among the conditions. No significant main effects were found in the ankle moment parameters.

Effect of the AFO resistances on the knee joints

Significant main effects were found in the knee angle at initial contact ($F[3, 27] = 3.66, P < 0.05, \eta_p^2 = 0.29$) for the adjustment of the plantarflexion resistance settings, but the post-hoc multiple comparisons did not show significant differences among the conditions (Fig. 3, Table 3). Both plantarflexion and dorsiflexion resistance adjustments appeared to have affected the peak knee flexor moment in stance, but no significant main effects ($P = 0.10$) were found (Fig. 3 and 4).

Effect of the AFO resistances on the hip joints

No significant main effects were found in the hip angle, moment and power parameters due to plantarflexion and dorsiflexion resistance adjustments.

Discussion

The aim of this study was to investigate the effects of the incremental changes in dorsiflexion and plantarflexion resistance settings of the AFO with the Triple Action ankle joints on lower limb joint kinematics and kinetics while walking in 10 individuals post-stroke. Our previous study investigated only the effects of plantarflexion resistance of an AFO on ankle and knee joints' angles and moments in individuals post-stroke (Kobayashi et al., 2015). In this study, we also tested the effects of dorsiflexion resistance and included more gait parameters: ankle, knee and hip joint power, hip angle, and hip moment. This study showed that adjustment of plantarflexion resistance had significant main effects on the ankle ($P<0.001$) and knee ($P<0.05$) angles at initial contact, while dorsiflexion resistance had significant ($P<0.01$) main effects on the peak dorsiflexion angle in stance. These outcomes are consistent with the findings of previous work using articulated AFOs with various resistance mechanisms: oil-damper (Kobayashi et al., 2011) and steel springs (Kobayashi et al., 2015).

Clinically objective indicators of the ankle and knee joint angles were used to determine the initial setting of the AFO's resistance. These indicators coincided with the results of the gait analysis. Fig. 3a and 4a suggest toe clearance in mid swing and heel contact at initial contact, while Fig. 3d and 4d suggest knee joints were mostly in the flexed position in early and late stance. Plantarflexion and dorsiflexion resistance adjustments appeared to affect the peak knee flexor moment in stance (Fig. 3e and 4e). In fact, the adjustment of the plantarflexion resistance demonstrated similar effects on the knee flexor moment to a previous study (Kobayashi et al., 2015). But no significant main effects were found in the peak knee flexor moment in the current study. Finally, adjustment of plantarflexion resistance demonstrated significant main effects in the peak ankle positive power in stance. It appeared that the peak ankle positive power was reduced as the plantarflexion resistance adjustment screw was turned toward the locked position (i.e. from P1 to P4) decreasing plantarflexion range of motion. Kinetic parameters, such as knee joint moment and knee power are influenced by gait speed (Lelas et al., 2003). In this study the treadmill was used and the participants walked at the same speed across different AFO conditions. In real world, the gait speed might be increased if the AFO resistance was correctly adjusted to each patient and some joint kinetics would be affected by the gait speed.

Generally, it is known that peak ankle positive power is decreased when plantarflexion is impeded in healthy adults (Huang et al., 2015). Plantarflexion resistance generated from the Triple Action joints demonstrated similar effects on the ankle power in individuals post-stroke. The plantarflexion resistance of AFOs primarily affects the kinematics and kinetics in early stance phase, and during late stance. And it is expected that this may interact with dorsiflexion resistance in mid- to late-stance of the gait cycle if the range of plantarflexion resistance allowed is excessive (e.g. equinus). The transition from early stance to mid-stance is the transition from the resistance to plantarflexion to the resistance to dorsiflexion, and the

opposite transition (dorsiflexion to plantarflexion resistance) occurs again at the transition between mid-stance and late-stance.

Mechanical testing of the AFO demonstrated that adjustment of plantarflexion and dorsiflexion resistance settings of the Triple action joint induced very systematic changes in its moment-angle relationship (Fig. 2). This outcome suggests the possibility of this joint to be clinically tuned to meet each patient's need. Fabricating a non-articulated AFO with specific resistance is very challenging. Use of resistance adjustable AFO joints may be a solution to this issue and potentially effective. Comparison of articulated AFOs and non-articulated AFOs on gait was investigated in children with cerebral palsy (Bennett et al., 2012; Radtka et al., 2005) and individuals with spinal cord injury (Arazpour et al., 2013). These studies used articulated AFOs with plantarflexion stop, and they focused only on the "articulation" function of conventional hinge joints. However, recently developed articulated joints have both "articulation" and "resistance adjustable" functions (Kerkum et al., 2015; Kobayashi et al., 2017; Yamamoto et al., 2005). A robust study is needed for evidence-based practice in choosing articulated or non-articulated AFOs for individuals post-stroke. Moreover, articulated AFOs tend to be bulkier and heavier than non-articulated AFOs. Miniaturization of joints is very important to be fully accepted in the clinical setting.

This study is not without limitations. First, this study investigated the immediate biomechanical effects of the adjustment of plantarflexion and dorsiflexion resistances of the AFO with the Triple Action joints. This AFO's long-term effects on gait, daily activities and quality of life in individuals post-stroke should be explored in future studies (Nikamp et al., 2017a). This group of individuals post-stroke was not very homogeneous and may not represent their general population (Table 1). Recruiting more specific group of participants may yield more specific outcomes. However, demonstrating significant main effects on gait parameters in this non-homogeneous group of participants is promising as it suggests that the effects of the resistance adjustments of the AFO on gait were general across these individuals. The instrumented split-treadmill was used to collect gait data. A previous study suggested that individuals post-stroke would walk differently on the treadmill compared to walking on the over ground (Kautz et al., 2011). The instrumented treadmill is very effective when collecting kinetic data under controlled gait speed from individuals post-stroke. However, how their gait with AFOs on the treadmill differs from their over ground gait requires further investigation. Moreover, assistance provided through the forearm crutch or the handrail was not quantified. Finally, the initial setting of the AFO was determined based on current clinical practice. It was beyond the scope of this study to define the optimal setting of the AFO. However, we believe that the range of resistances that each individual tested encompassed the full range of clinically reasonable resistances to motion. To go farther than any of these measures was either impossible (maximum resistance) or was so compliant as to be clinically ineffective to halt pathological motion.

Conclusions

This study demonstrated that the adjustments of resistance of the ankle-foot orthosis with the Triple Action joints affected ankle and knee joint kinematics while walking in individuals

post-stroke. Future studies should investigate the long-term effects of the resistance-adjustable articulated AFOs on their gait and daily activities.

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(a) Medial view of the AFO

(b) Anterior view of the AFO

Fig. 1.
An articulated ankle-foot orthosis (AFO) with Becker Triple Action[®] ankle joints: (a) Medial view of the AFO, (b) Anterior view of the AFO.

Fig. 2.

The moment-angle relationships of the AFO under each plantarflexion and dorsiflexion resistance setting. The numbers for the plantarflexion (PF) and dorsiflexion (DF) resistance adjustment indicate the number of turns away from the locked position (i.e. 0 turn position).

Fig. 3.

Effect of the plantarflexion resistance adjustments of the AFO on the mean lower-limb joint kinematics and kinetics on the affected side for the individuals post-stroke during a gait cycle. Note that gray lines are mean data of the non-impaired controls in each graph. Effect of the plantarflexion resistance adjustments of the AFO on (a) the mean ankle angle, (b) the mean ankle moment, (c) the mean ankle power, (d) the mean knee angles, (e) the mean knee moment, (f) the mean knee power, (g) the mean hip angles, (h) the mean hip moment, and (i) the mean hip power. Abbreviations: DF, dorsiflexion in angle and dorsiflexor in moment; EX, extension in angle and extensor in moment; FX, flexion in angle and flexor in moment; PF, plantarflexion in angle and plantarflexor in moment.

Fig. 4.

Effect of the dorsiflexion resistance adjustments of the AFO on the mean lower-limb joint kinematics and kinetics on the affected side for the individuals post-stroke during a gait cycle. Note that gray lines are mean data of the non-impaired controls in each graph. Effect of the dorsiflexion resistance adjustments of the AFO on (a) the mean ankle angle, (b) the mean ankle moment, the mean ankle power, (d) the mean knee angles, (e) the mean knee moment, (f) the mean knee power, (g) the mean hip angles, (h) the mean hip moment, and (i) the mean hip power. Abbreviations: DF, dorsiflexion in angle and dorsiflexor in moment; EX, extension in angle and extensor in moment; FX, flexion in angle and flexor in moment; PF, plantarflexion in angle and plantarflexor in moment.

Table 1.

Demographic and clinical assessment data of the individuals post-stroke.

<i>Demographic data</i>	Gender	Cause	Age	Years since stroke	Affected Side	Height (m)	Body mass (kg)
S01	Male	H	40	4	Left	1.91	113
S02	Female	I	50	6	Left	1.73	77
S03	Male	I	49	4	Left	1.80	94
S04	Male	I	59	2	Right	1.78	64
S05	Male	I	45	6	Right	1.88	96
S06	Male	H&I	65	5	Right	1.70	107
S07	Male	I	67	9	Left	1.73	60
S08	Female	I	71	5	Right	1.52	84
S09	Male	I	80	9	Left	1.75	89
S10	Female	H	53	2	Right	1.57	53

<i>Clinical Assessment</i>	Peak DF RoM (knee flexed)	Hip Flexor MMT	Knee Extensor MMT	Knee Flexor MMT	Hip Flexor MMT	Dorsiflexor MMT	Plantarflexor MMT
S01	0°	4	4	2+	4	3-	4-
S02	5°	4	5	5	4	1+	4
S03	5°	5	5	4-	5	0	1
S04	5°	3+	5	3+	3+	1+	0
S05	20°	4	5	3-	4	1+	3+
S06	20°	4-	5	5	4-	3+	4
S07	-5°	3	4	3-	3	1+	2+
S08	0°	4-	4	4	4-	4	4-
S09	5°	4+	4+	5	4+	4+	5
S10	20°	5	5	5	5	3	3

Abbreviations: H, hemorrhagic; I, ischemic; MMT, manual muscle testing; DF, dorsiflexion; RoM, range of motion.

Table 2.

(a) The initial settings and plantarflexion (PF) resistance conditions (P1 to P4) of the AFO tested in each participant. The initial plantarflexion resistance setting is highlighted in light gray. (b) The initial settings and dorsiflexion (DF) resistance conditions (D1 to D4) of the AFO tested in each participant. The initial dorsiflexion resistance setting is highlighted in dark gray.

(a) PF resistance						
	P1	P2	P3	P4	Alignment	DF resistance
S01	1.5	1.0	0.5	0.0	6°	3.0
S02	3.0	2.5	2.0	1.5	2°	1.0
S03	1.5	1.0	0.5	0.0	6°	1.0
S04	2.0	1.5	1.0	0.5	2°	1.0
S05	2.0	1.5	1.0	0.5	4°	3.0
S06	2.5	2.0	1.0	0.5	2°	2.0
S07	2.5	1.5	1.0	0.0	4°	2.0
S08	4.0	2.5	1.5	1.0	-2°	1.0
S09	2.0	1.5	1.0	0.0	0°	2.0
S10	2.0	1.5	1.0	0.5	2°	3.0

(b) DF resistance						
	D1	D2	D3	D4	Alignment	PF resistance
S01	4.0	3.5	3.0	2.5	6°	0.5
S02	2.5	2.0	1.5	1.0	2°	2.5
S03	1.5	1.0	0.5	0.0	6°	0.5
S04	1.5	1.0	0.5	0.0	2°	1.0
S05	3.0	2.5	2.0	1.5	4°	1.5
S06	3.5	3.0	2.0	1.5	2°	2.0
S07	2.5	2.0	1.5	1.0	4°	1.0
S08	2.5	2.0	1.5	1.0	-2°	1.5
S09	3.5	2.5	2.0	0.0	0°	0.0
S10	3.5	3.0	2.5	2.0	2°	1.0

Table 3.

Effect of plantarflexion and dorsiflexion resistance adjustments of the AFO on lower-limb joint kinematics and kinetics parameters of gait in the individuals post-stroke.

Effect of PF resistance adjustment	P1			P2			P3			P4		
	Mean (SD)	95% CI		Mean (SD)	95% CI		Mean (SD)	95% CI		Mean (SD)	95% CI	
(a) Ankle angle at initial contact (°) *	0.42 (2.15) ^{a,b}	-1.12, 1.95		2.02 (2.44) ^c	0.27, 3.76		2.88 (2.57) ^d	1.04, 4.71		4.61 (3.22)	2.31, 6.91	
(b) Peak dorsiflexion angle (°)	8.37 (3.35)	6.00, 10.76		8.53 (3.39)	6.11, 10.96		8.44 (3.48)	5.95, 10.93		9.01 (3.54)	6.48, 11.54	
(c) Peak dorsiflexor moment (Nm/kg)	-0.07 (0.07)	-0.13, -0.02		-0.08 (0.07)	-0.13, -0.03		-0.08 (0.08)	-0.14, -0.02		-0.08 (0.08)	-0.14, -0.02	
(d) Peak plantarflexor moment (Nm/kg)	0.55 (0.32)	0.31, 0.80		0.55 (0.30)	0.32, 0.78		0.53 (0.34)	0.27, 0.79		0.54 (0.32)	0.30, 0.79	
(e) Peak ankle positive power (W/kg) *	0.11 (0.10)	0.03, 0.19		0.10 (0.10)	0.03, 0.18		0.09 (0.09)	0.02, 0.16		0.09 (0.81)	0.02, 0.15	
(f) Knee angle at initial contact (°) *	7.43 (7.75)	1.89, 12.98		8.42 (7.53)	3.03, 13.81		8.38 (7.45)	3.05, 13.71		9.42 (6.91)	4.48, 14.37	
(g) Peak knee extension angle (°)	-2.69 (9.09)	-9.20, 3.81		-1.19 (8.63)	-7.37, 4.99		-1.27 (7.86)	-6.89, 4.35		0.31 (7.36)	-4.95, 5.58	
(h) Peak knee flexor moment (Nm/kg)	-0.38 (0.23)	-0.55, -0.20		-0.32 (0.20)	-0.48, -0.16		-0.31 (0.17)	-0.44, -0.18		-0.25 (0.17)	-0.38, -0.12	
(i) Hip angle at initial contact (°)	18.91 (8.99)	12.48, 25.34		19.52 (9.01)	13.08, 25.96		19.63 (8.95)	13.23, 26.04		19.40 (8.66)	13.21, 25.59	
(j) Peak hip extension angle (°)	2.28 (8.71)	-3.95, 8.51		3.04 (9.34)	-3.64, 9.72		3.60 (8.72)	-2.64, 9.84		3.23 (9.35)	-3.46, 9.92	
(k) Peak hip extensor moment (Nm/kg)	0.24 (0.17)	0.17, 0.37		0.23 (0.15)	0.12, 0.35		0.25 (0.15)	0.13, 0.37		0.23 (0.15)	0.12, 0.34	
(l) Peak hip flexor moment (Nm/kg)	-0.19 (0.08)	-0.25, -0.13		-0.17 (0.07)	-0.23, -0.12		-0.17 (0.06)	-0.21, -0.13		-0.18 (0.07)	-0.24, -0.13	
Effect of DF resistance adjustment	D1			D2			D3			D4		
	Mean (SD)	95% CI		Mean (SD)	95% CI		Mean (SD)	95% CI		Mean (SD)	95% CI	
(a) Ankle angle at initial contact (°)	3.36 (1.96)	1.95, 4.76		3.21 (2.63)	1.33, 5.09		2.94 (2.14)	1.41, 4.47		2.86 (2.06)	1.39, 4.33	
(b) Peak dorsiflexion angle (°) *	10.42 (2.95) ^{e,f}	8.31, 12.54		9.51 (2.60)	7.65, 11.38		8.94 (2.57)	7.10, 10.80		8.53 (2.11)	7.02, 10.04	
(c) Peak dorsiflexor moment (Nm/kg)	-0.08 (0.09)	-0.15, -0.01		-0.09 (0.08)	-0.15, -0.03		-0.08 (0.09)	-0.14, -0.01		-0.09 (0.08)	-0.15, -0.02	
(d) Peak plantarflexor moment (Nm/kg)	0.55 (0.32)	0.30, 0.80		0.53 (0.35)	0.27, 0.80		0.54 (0.34)	0.28, 0.80		0.55 (0.32)	0.30, 0.79	
(e) Peak ankle positive power (W/kg)	0.13 (0.13)	0.03, 0.22		0.11 (0.13)	0.01, 0.21		0.11 (0.11)	0.02, 0.19		0.08 (0.09)	0.02, 0.15	
(f) Knee angle at initial contact (°)	8.59 (7.40)	3.30, 13.88		8.31 (7.26)	3.12, 13.51		8.32 (7.39)	3.04, 13.61		8.05 (6.91)	3.11, 13.00	
(g) Peak knee extension angle (°)	-0.31 (9.36)	-7.00, 6.39		-1.19 (8.23)	-7.09, 4.70		-0.97 (8.33)	-6.93, 4.99		-1.58 (7.95)	-7.27, 4.11	
(h) Peak knee flexor moment (Nm/kg)	-0.28 (0.19)	-0.42, -0.13		-0.30 (0.17)	-0.43, -0.17		-0.31 (0.17)	-0.44, -0.17		-0.32 (0.16)	-0.44, -0.20	
(i) Hip angle at initial contact (°)	19.97 (8.04)	14.22, 25.72		19.65 (9.17)	13.09, 26.21		20.03 (8.10)	14.24, 25.83		20.00 (8.42)	13.97, 26.02	
(j) Peak hip extension angle (°)	1.84 (7.72)	-3.68, 7.37		2.12 (9.29)	-4.53, 8.76		2.99 (8.31)	-2.95, 8.94		2.18 (8.05)	-3.58, 7.94	

Effect of PF resistance adjustment	P1		P2		P3		P4	
	Mean (SD)	95% CI	Mean (SD)	95% CI	Mean (SD)	95% CI	Mean (SD)	95% CI
(k) Peak hip extensor moment (Nm/kg)	0.23 (0.14)	0.12, 0.33	0.24 (0.15)	0.12, 0.35	0.23 (0.15)	0.12, 0.35	0.22 (0.15)	0.10, 0.34
(l) Peak hip flexor moment (Nm/kg)	-0.18 (0.06)	-0.23, -0.14	-0.19 (0.07)	-0.24, -0.13	-0.18 (0.07)	-0.23, -0.13	-0.17 (0.06)	-0.22, -0.13

(*) Asterisks indicate significant main effects (at least $P < 0.05$) by one-way repeated measures ANOVA.

Superscripts (a) –(f) indicate significant differences (at least $P < 0.05$) between:

(a) P1 and P3

(b) P1 and P4

(c) P2 and P4

(d) P3 and P4

(e) D1 and D3

(f) D1 and D4.

Plantarflexion resistance settings (P1 to P4) and dorsiflexion resistance settings (D1 to D4) for each participant are defined in Table 2.

Abbreviations: DF, dorsiflexion; PF, plantarflexion; SD, standard deviation; 95% CI, 95% confidence interval