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

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ABSTRACT

Mechanical tuning of an ankle-foot orthosis (AFO) is important in improving gait in individuals post-stroke. Alignment and resistance are two factors that are tunable in articulated AFOs. The aim of this study was to investigate the effects of changing AFO ankle alignment on lower limb joint kinematics and kinetics with constant dorsiflexion and plantarflexion resistance in individuals post-stroke. Gait analysis was performed on 10 individuals post-stroke under four distinct alignment conditions using an articulated AFO with an ankle joint whose alignment is adjustable in the sagittal plane. Kinematic and kinetic data of lower limb joints were recorded using a Vicon 3-dimensional motion capture system and Bertec split-belt instrumented treadmill. The incremental changes in the alignment of the articulated AFO toward dorsiflexion angles significantly affected ankle and knee joint angles and knee joint moments while walking in individuals post-stroke. No significant differences were found in the hip joint parameters. The alignment of the articulated AFO was suggested to play an important role in improving knee joint kinematics and kinetics in stance through improvement of ankle joint kinematics while walking in individuals post-stroke. Future studies should investigate long-term effects of AFO alignment on gait in the community in individuals post-stroke.

1. Introduction

Ankle-foot orthoses (AFOs) improve gait in individuals post-stroke (Tyson et al., 2013). Individuals post-stroke generally present with muscle weakness and contracture in their lower-limb, which affect their gait. AFOs assist the affected lower-limb by providing proper structural resistance to dorsiflexion and plantarflexion and help to achieve appropriate ankle joint positions while walking. AFOs are generally classified into two groups: non-articulated and articulated AFOs. An articulated AFO has a movable joint at the ankle, while a non-articulated AFO does not have a movable joint, although there may be some motion due to bending of the AFO material.

Alignment of an articulated AFO is generally defined as the initial angle set on the joint in the sagittal plane. This ankle joint position is the location that resistive moments of the AFO joint are initiated in plantarflexion and/or dorsiflexion directions. The importance of tuning alignment is commonly recognized in lower limb prostheses, and its effect on gait has been widely studied in individuals with limb loss (Boone et al., 2013; Schmalz et al., 2002). Alignment of an AFO is also believed to be important to improve gait in individuals post-stroke in a clinical setting. Generally, this angle is set depending on the available range of motion of the ankle joint and muscle strength of lower limb joints in individuals post-stroke in order to assist rocker functions in stance and toe clearance in swing.

Mechanical tuning of an AFO is also accomplished through adjustment of the resistance in some articulated AFOs. The resistance in the sagittal plane generated by articulated AFOs consists of dorsiflexion and plantarflexion resistance (i.e. resistive moment). Systematic changes in plantarflexion resistance of an

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articulated AFO have been shown to affect ankle and knee joint kinematics and kinetics during gait in individuals post-stroke (Kobayashi et al., 2015). In addition, kinematics of the ankle and knee joints can be regulated by the amount of plantarflexion resistance generated by an articulated AFO (Kobayashi et al., 2017b). Dorsiflexion resistance was also shown to affect ankle kinematics during gait in individuals post-stroke (Kobayashi et al., 2011). Therefore, the existing evidence suggests tuning of resistance of an articulated AFO is important in improving gait in individuals post-stroke.

Some articulated AFOs allow more range of motion of the ankle joint while still providing support for the limb using the resistance provided by the movable joint (Kerkum et al., 2015; Kobayashi et al., 2017a; Yamamoto et al., 2005). These articulated AFOs also allow quick tuning of the mechanical properties while clinically customizing the AFO mechanical characteristics to the patient's unique biomechanical deficits to improve their gait. This may not be possible with non-articulated AFOs, which require irreversible modification of their structure through trimming to decrease their resistance to bending in the dorsiflexion and plantarflexion directions. This approach to AFO tuning may also utilize tuning of the shoe rocker to change the interface between the orthosis and the floor to adjust alignment (Owen, 2010). Therefore, ability to adjust alignment and resistance through a joint mechanism may be advantageous for articulated AFOs.

Emerging evidence suggests that adjustment of an articulated AFO's resistance affects gait in individuals post-stroke, however there remains a paucity of evidence related to the effects of alignment. A previous study investigated the effect of alignment of an articulated AFO by manipulating shoe heel-height, but it did not demonstrate a clear effect on ankle and knee joint kinematics and kinetics (Fatone et al., 2009). Therefore, a more systematic experimental approach is necessary to investigate the effects of alignment more comprehensively. The aim of this study was to investigate the effects of changing ankle alignment of an articulated AFO on lower limb joint kinematics and kinetics with constant dorsiflexion and plantarflexion resistance in individuals post-stroke. We hypothesized that ankle, knee and hip kinematics and kinetics would be significantly affected by changing the alignment of the AFO. Specifically, the ankle kinematics and kinetics would change with changes to the alignment at the articulated AFO joint. Knee and hip joint kinematics and kinetics would be impacted indirectly based on the direct effects of the alignment changes of the AFO at the ankle joint.

2. Methods

2.1. Participants

Ten subjects (3 females/7 males) with a history of stroke (N01-N10) participated in this study (Table 1). Their mean (SD: standard deviation) age was 58 (13) years old and mean (SD) time since stroke incidence was 5 (2) years. Their mean (SD) body height was 1.74 (0.12) m and mean (SD) body mass was 84 (20) kg. All subjects had unilateral limb involvement (5 right/5 left). Subjects included in the study were a minimum of 6-month post-stroke with hemiparesis and able to safely walk on an instrumented level treadmill with the use of an AFO. Subjects were excluded from the study if they had any confounding injury or musculoskeletal or cognitive problems that might limit the ability to walk on the treadmill. After informed consent was obtained for this Institutional Review Board approved study, the following clinical assessment was performed on all the subjects: (1) Manual muscle testing (MMT) of the ankle, knee and hip joints of the affected limb (Kendall et al., 1993), and (2) Measurement of manual passive peak dorsiflexion angle (i.e. range of motion: ROM) of the affected limb with the knee flexed at 90° (Baumbach et al., 2014).

2.2. Articulated ankle-foot orthosis

An AFO was custom fabricated for each subject from 4.8 mm polypropylene. Supportive elements were incorporated in the orthotic contour based upon the subject's unique postural control needs. All AFOs were fabricated using two prototype Becker Triple Action® ankle joints (Troy, MI, USA), one medial and one lateral (Fig. 1a). The Becker Triple Action® ankle joints facilitates the independent tuning of ankle alignment, dorsiflexion resistance and plantarflexion resistance (Fig. 1b). The alignment adjustment rotates the joint body and stirrup around the pivot bushing, and is adjusted by turning the hex on the front of the joint body. The resistance is adjusted by altering compression on the springs located in the dorsiflexion and plantarflexion channels. Resistance

Table 1
Demographic data of the subjects post-stroke and outcome of clinical assessments.

Demographic Data	Gender	Age	Height (cm)	Weight (kg)	Affected Side	Cause	Years since stroke
N01	M	40	191	113	L	H	4
N02	F	50	173	77	L	I	6
N03	M	49	180	94	L	I	4
N04	M	59	178	64	R	I	2
N05	M	45	188	96	R	I	6
N06	M	65	170	107	R	H&I	5
N07	M	67	173	60	L	I	9
N08	F	71	152	84	R	I	5
N09	M	80	175	89	L	I	9
N10	F	53	157	53	R	H	2
Clinical Assessment	**Plantarflexor MMT**	**Dorsiflexor MMT**	**Knee Flexor MMT**	**Knee Extensor MMT**	**Hip Flexor MMT**	**Hip Extensor MMT**	**Peak DF ROM**
N01	4−	3−	2+	4	4	2+	0°
N02	4	1+	5	5	4	3+	5°
N03	1	0	4−	5	5	3	5°
N04	0	1+	3+	5	3+	3	5°
N05	3+	1+	3−	5	4	4	20°
N06	4	3+	5	5	4−	4−	20°
N07	2+	1+	3−	4	3	2−	−5°
N08	4-	4	4	4	4−	4−	0°
N09	5	4+	5	4+	4+	4+	5°
N10	3	3	5	5	5	3+	20°

Abbreviations: M, male; F, female; L, left; R, right; H, hemorrhagic; I, ischemic; MMT, manual muscle testing; DF, dorsiflexion; ROM, range of motion.

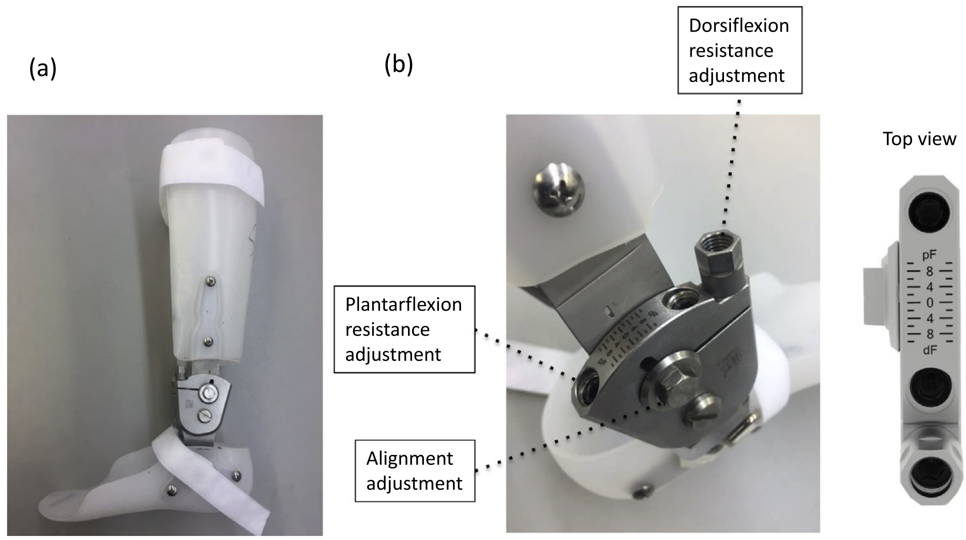

Fig. 1. (a) An articulated AFO with Becker Triple Action® ankle joint, (b) Adjustments of alignment, dorsiflexion resistance, and plantarflexion resistance of the AFO.

settings are adjusted by turning the resistance adjustment screws on top of the joint body.

The mechanical properties (i.e. angle-moment relationship) of the AFO fabricated for subject N01 were quantified as a representative sample. The angle-moment relationship of this AFO was measured within a torque range of ±30 Nm or an angle range of ±15°. This quantification was done using a custom motorized mechanical testing device (Gao et al., 2011) under four alignment conditions (0°, 2°, 4° and 6° of dorsiflexion), where dorsiflexion resistance was kept at 3.0 screw turns away and plantarflexion resistance was kept at 0.5 screw turns away from the locked position, to verify the effect of the alignment changes (Fig. 2). Details on the mechanical testing of the AFO can be found elsewhere (Kobayashi et al., 2017a).

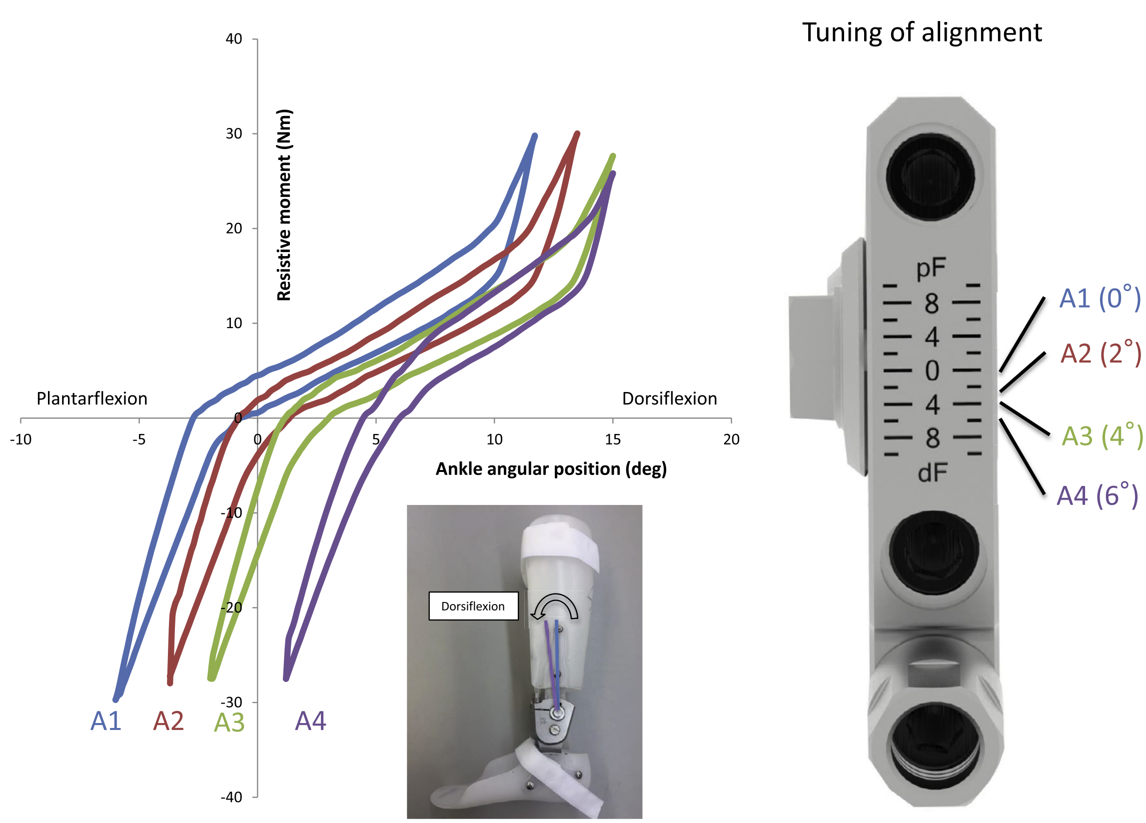

Fig. 2. Mechanical properties of the AFO for subject N01 tested with a custom motorized mechanical testing device under four alignment conditions (A1 to A4: 0°, 2°, 4° and 6° of dorsiflexion), where dorsiflexion resistance was kept at 3.0 screw turns away and plantarflexion resistance was kept at 0.5 screw turns away from the locked position.

2.3. Fitting of the ankle-foot orthoses

The subjects were fit with the previously described custom-made articulated AFO. Each AFO was kinematically tuned for the subject by a certified orthotist using observational gait analysis and feedback from the subjects to establish what was considered clinically appropriate initial settings of the alignment, dorsiflexion resistance and plantarflexion resistance settings of the AFO joints. All the subjects wore the same type and model shoes (New Balance 928, New Balance Inc., USA). Shoe size was selected by the certified orthotist as most clinically appropriate for each subject. These clinically appropriate initial ankle joint settings served as baseline settings during the gait trials. From these settings, the alignment of the AFO joint was randomly changed across trials (i.e. increased or decreased dorsiflexion angles, which reflected general practice in the clinical setting) to one of four angles (A1 to A4) within the range of 2° of plantarflexion to 8° of dorsiflexion with the resistance settings maintained at their clinically-determined settings set initially by the certified orthotist. The initial settings and tested alignment conditions (A1 to A4) for each subject are shown in Table 2. Note that the numbers in the table representing the dorsiflexion and plantarflexion resistance settings indicate the number of turns away from the locked (0° ROM) setting. Therefore, larger number represents less resistance to ankle motion in that direction.

2.4. Gait analysis

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)] for gait analysis. The markers were placed directly on the AFO on the shank and foot of the affected limb of the subjects. A rigid cluster was secured to the lateral side of the AFO for dynamic tracking. The subject was secured in a safety harness and asked to walk on a level split-belt instrumented treadmill (Bertec corporation, Columbus, OH, USA). They walked while wearing the AFO on the affected leg under the four alignment conditions (A1 to A4) at a self-selected speed of 0.21–0.36 m/s. The gait speed of the treadmill was kept constant across the alignment conditions for each subject. They were given a short acclimatization period walking on the treadmill before data collection. Gait data were collected using a Bertec instrumented treadmill and Vicon 10-camera motion analysis system (Vicon Motion Systems, Oxford, UK) at a rate of 200 Hz.

Data were recorded and synchronized using Vicon Nexus software (Vicon Motion Systems, Oxford, UK) and post-processed using Visual3D (CMotion, Germantown, MD, USA). A low pass, zero-phase shift Butterworth filter at 6 Hz and 20 Hz was used to filter marker trajectory and force platform data, respectively. For subject N07, only the kinematic data were included due to technical issues with the kinetic data. The ankle, knee and hip joint angles, moments and power were calculated by averaging over at least 8 gait cycles and normalized to body mass for each of the four alignment conditions of the AFO. The lower-limb wearing the AFO was analyzed for each subject. Subsequently, the mean of the 10 subjects was calculated and plotted for the ankle, knee and hip joint angle, moment and power (Fig. 3). Ankle dorsiflexion, knee flexion and hip flexion were defined as positive for the joint angles, while ankle plantarflexor, knee extensor and hip extensor were defined positive for the internal joint moments.

The following joint kinematic and kinetic parameters were extracted from the gait data of the subjects 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 (°), g) Peak knee extension angle in stance (°), h) Peak knee flexor moment in stance (Nm/kg), i) Hip angle at initial contact (°), j) Peak hip extension angle in stance (°), k) Peak hip extensor moment in stance (Nm/kg), and l) Peak hip flexor moment (Nm/kg).

2.5. Statistical analysis

One-way repeated measures ANOVA was conducted to compare the lower limb joint kinematic and kinetic parameters among the four alignment conditions (A1-A4). 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 alignment conditions. 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 $\alpha = 0.05$.

3. Results

3.1. Clinical assessment of subjects post-stroke

The manual muscle testing revealed that plantarflexor muscle strength ranged from 0 to 5, dorsiflexor muscle strength ranged

Table 2
The initial settings and alignment conditions (A1 to A4) of the AFO tested in each subject. The initial alignment setting is highlighted in light gray.

Subject	A1	A2	A3	A4	DF resistance	PF resistance
N01	0°	2°	4°	6°	3.0	0.5
N02	0°	2°	4°	6°	1.0	2.5
N03	2°	4°	6°	8°	1.0	0.5
N04	−2°	0°	2°	4°	1.0	1.0
N05	0°	2°	4°	6°	3.0	1.5
N06	−2°	0°	2°	6°	2.0	2.0
N07	−2°	0°	4°	6°	2.0	1.0
N08	−2°	0°	2°	4°	1.0	1.5
N09	0°	2°	4°	6°	2.0	0.0
N10	−2°	0°	2°	4°	3.0	1.0

Abbreviations: DF, dorsiflexion; PF, plantarflexion.
Note: Resistance settings were adjusted by counting the number of turns away from the locked position. The numbers in the table for dorsiflexion and plantarflexion resistance indicate the number of turns away from the locked position. Thus, the larger number indicates less resistance.

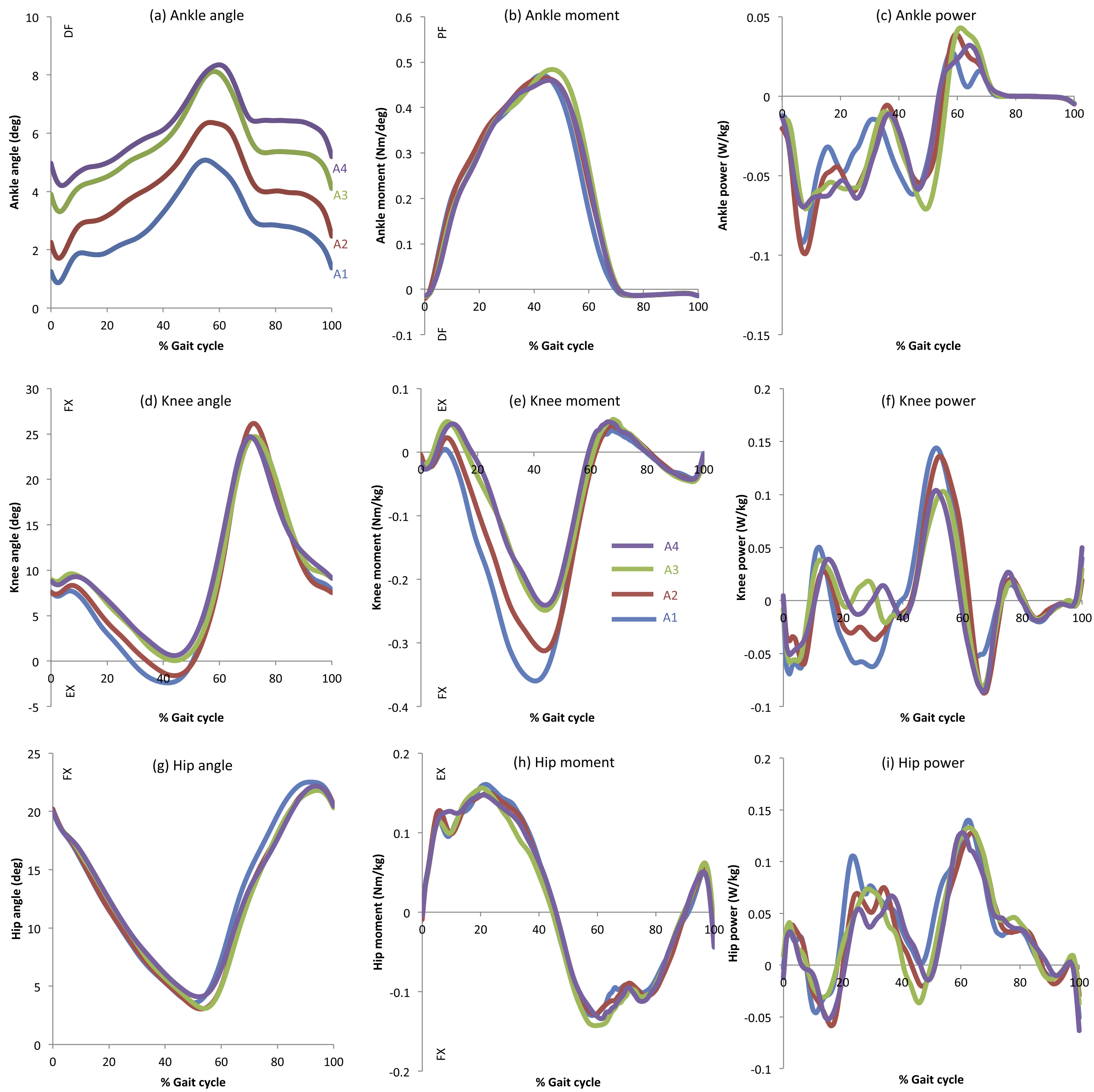

Fig. 3. Effect of the alignment changes of the AFO on the mean lower-limb joint kinematics and kinetics in the affected side for subjects post-stroke in a gait cycle. (a) Effect of the alignment changes of the AFO on the mean ankle angle, (b) Effect of the alignment changes of the AFO on the mean ankle moment, (c) Effect of the alignment changes of the AFO on the mean ankle power, (d) Effect of the alignment changes of the AFO on the mean knee angles, (e) Effect of the alignment changes of the AFO on the mean knee moment, (f) Effect of the alignment changes of the AFO on the mean knee power, (g) Effect of the alignment changes of the AFO on the mean hip angles, (h) Effect of the alignment changes of the AFO on the mean hip moment, and (i) Effect of the alignment changes of the AFO on the mean hip power. Abbreviations: DF, dorsiflexion (for angle) and dorsiflexor (for moment); PF, plantarflexor (for moment); EX, extension (for angle) and extensor (for moment); FX, flexion (for angle) and flexor (for moment).

from 0 to 4+, knee flexor muscle strength ranged from 2+ to 5, knee extensor muscle strength ranged from 4 to 5, hip flexor muscle strength ranged from 3 to 5, and hip extensor muscle strength ranged from 2− to 4. The range of motion of peak dorsiflexion ranged from $-5°$ to $20°$ (Table 1).

3.2. Mechanical properties of the AFO

There were distinct differences in angle-moment relationship across alignment conditions, which verified the effect of changing the alignment of the AFO on its mechanical properties (Fig. 2).

3.3. Effect of the AFO alignment on ankle joints

Significant differences were found in the ankle angle at initial contact ($F[3, 27] = 50.370$, $P < 0.001$, $\eta_p^2 = 0.848$) and the peak dorsiflexion angle in stance ($F[3, 27] = 40.461$, $P < 0.001$, $\eta_p^2 = 0.818$) (Fig. 3, Table 3). For the ankle angle at initial contact, post-hoc comparisons showed significant differences of A1 < A2 ($P < 0.05$), A1 < A3 ($P < 0.001$), A1 < A4 ($P < 0.001$), A2 < A3 ($P < 0.01$), A2 < A4 ($P < 0.001$), and A3 < A4 ($P < 0.05$) (Fig. 4). For the peak dorsiflexion angle in stance, post-hoc comparisons showed significant differences of A1 < A2 ($P < 0.01$), A1 < A3 ($P < 0.001$), A1 < A4 ($P < 0.001$), A2 < A3 ($P < 0.05$), and A2 < A4 ($P < 0.001$) (Fig. 4). Therefore, at the ankle

Table 3
Effect of the alignment changes of the articulated AFO on the lower-limb joint kinematic and kinetic parameters of gait in the affected lower-limb for subjects post-stroke.

Gait parameters	A1		A2		A3		A4	
	Mean (SD)	95% CI	Mean (SD)	95% CI	Mean (SD)	95% CI	Mean (SD)	95% CI
(a) Ankle angle at initial contact (°)	1.26 (2.20)[a,b,c]	−0.316, 2.83	2.25(2.52)[d,e]	−0.45, 4.06	3.91(2.49)[f]	2.13, 5.69	4.96(1.91)	3.60, 6.33
(b) Peak dorsiflexion angle (°)	7.49(2.16)[a,b,c]	5.94, 9.03	8.97(2.65)[d,e]	7.07, 10.87	10.03(2.54)	8.21,11.85	10.81(2.90)	8.74,12.88
(c) Peak dorsiflexor moment (Nm/kg)	−0.08(0.08)	−0.14, −0.02	−0.08(0.09)	−0.15, −0.01	−0.09(0.08)	−0.16, −0.02	−0.09(0.10)	−0.16, −0.02
(d) Peak plantarflexor moment (Nm/kg)	0.54(0.31)	0.30, 0.78	0.55(0.34)	0.29, 0.82	0.55(0.33)	0.30, 0.81	0.56(0.37)	0.28, 0.84
(e) Peak ankle positive power (W/kg)	0.09(0.08)	0.03,0.16	0.11(0.10)	0.03,0.19	0.09(0.08)	0.03,0.16	0.12(0.11)	0.03,0.20
(f) Knee angle at initial contact (°)	7.63(7.51)	2.25, 13.00	7.56(7.67)[e]	2.07, 13.05	9.01(7.36)	3.74, 14.27	8.82(8.28)	2.90,14.74
(g) Peak knee extension angle (°)	−2.62(8.80)[c]	−8.91, 3.68	−1.98(9.20)[e]	−8.56, 4.60	−0.23(8.78)	−6.51, 6.06	0.37(10.11)	−6.86, 7.60
(h) Peak knee flexor moment (Nm/kg)	−0.38(0.20)[b,c]	−0.53, −0.22	−0.34(0.19)[e]	−0.49, −0.19	−0.27(0.20)	−0.42, −0.12	−0.27(0.22)	−0.44, −0.11
(i) Hip angle at initial contact (°)	20.20(8.47)	14.14, 26.26	20.18(8.53)	14.08, 26.28	19.96(8.45)	13.92, 26.00	20.03(8.47)	13.97, 26.08
(j) Peak hip extension angle (°)	2.37(7.59)	−3.06, 7.81	1.67(8.41)	−4.34, 7.69	1.99(8.40)	−4.02, 8.00	2.59(7.58)	−2.84, 8.01
(k) Peak hip extensor moment (Nm/kg)	0.24(0.15)	0.12,0.35	0.23(0.13)	0.13,0.33	0.23(0.14)	0.12,0.34	0.22(0.13)	0.13,0.32
(l) Peak hip flexor moment (Nm/kg)	−0.18(0.07)	−0.24, −0.13	−0.17(0.06)	−0.22, −0.12	−0.19(0.08)	−0.26, −0.13	−0.20(0.06)	−0.25, −0.15

Labels (a)–(f) indicate significant differences (at least $P < 0.05$) between: (a): A1 and A2, (b) A1 and A3, (c) A1 and A4, (d) A2 and A3, (e) A2 and A4, and (f) A3 and A4.
Abbreviations: SD, standard deviation; 95% CI, 95% confidence interval.

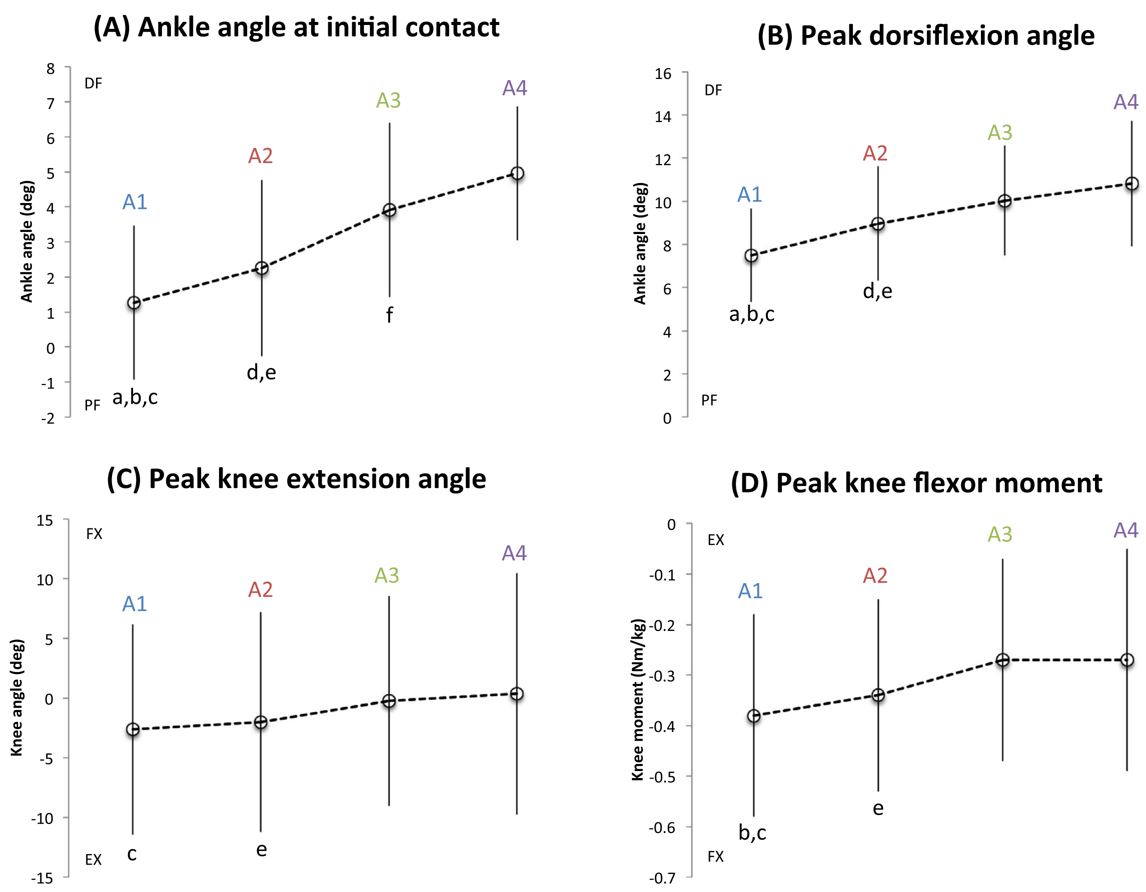

Fig. 4. Significant effects of the alignment of the AFO on the representative ankle and knee joint kinematic and kinetic parameters: (A) Ankle angle at initial contact, (B) Peak dorsiflexion angle, (C) Peak knee extension angle, (D) Peak knee flexor moment. Labels (a) –(f) indicate significant differences (at least $P < 0.05$) between: (a): A1 and A2, (b) A1 and A3, (c) A1 and A4, (d) A2 and A3, (e) A2 and A4, and (f) A3 and A4. Abbreviations: DF, dorsiflexion; PF, plantarflexion; EX, extension (for angle) and extensor (for moment); FX, flexion (for angle) and flexor (for moment).

joint, the joint angles at initial contact and peak dorsiflexion angles in stance were significantly increased as the dorsiflexion alignment angle of the AFO was increased. Despite these angular changes observed at the ankle joint, there were no significant kinetic changes due to alignment changes ($P = 0.534$ for peak dorsiflexor moment in stance; $P = 0.925$ for peak plantarflexor moment in stance).

3.4. Effect of the AFO alignment on knee joints

Significant differences were found in the knee angle at initial contact (F[3, 27] = 3.638, $P < 0.05$, $\eta_p^2 = 0.288$), the peak knee extension angle in stance (F[3, 27] = 6.192, $P < 0.01$, $\eta_p^2 = 0.408$), and the peak knee flexor moment in stance (F[2.018, 16.140] = 18.969, $P < 0.001$, $\eta_p^2 = 0.703$) (Fig. 3, Table 2). For the knee angle at initial contact, post-hoc comparisons showed significant differences of A2 < A4 ($P < 0.05$). For the peak extension angle in stance, post-hoc comparisons showed significant differences of A1 < A4 ($P < 0.05$), and A2 < A4 ($P < 0.05$) (Fig. 4). For the peak knee flexor moment in stance, post-hoc comparisons showed significant differences of A1 > A3 ($P < 0.01$), A1 > A4 ($P < 0.001$), and A2 > A4 ($P < 0.01$) (Fig. 4). Therefore, at the knee joint, the peak knee extension angles and the peak knee flexor moments in stance were significantly decreased as the dorsiflexion alignment angle of the AFO was increased.

3.5. Effect of the AFO alignment on hip joints

No significant differences were found in the hip kinematic and kinetic parameters by changing the alignment of the AFO (Fig. 3, Table 3).

4. Discussion

The aim of this study was to investigate the effects of changing ankle alignment of an articulated AFO on lower limb joint kinematics and kinetics with constant dorsiflexion and plantarflexion resistance in individuals post-stroke. The changes in alignment of the articulated AFO significantly affected ankle and knee joint angles and knee joint moments for individuals post-stroke walking on a level treadmill (Fig. 4, Table 3). From a practical standpoint, these data suggest that increasing ankle dorsiflexion in an AFO can reduce knee hyperextension by decreasing the knee extensor moment in single limb stance in individuals post-stroke with knee recurvatum. No significant differences were found in hip joint kinematics and kinetics, which suggests that the effect of the ankle alignment of an articulated AFO with constant resistance to dorsiflexion and plantarflexion is limited to ankle joint kinematics and knee joint kinematics and kinetics. However, a previous study demonstrated that ankle dorsiflexor moments were significantly affected in individuals post-stroke when plantarflexion resistance of an articulated AFO was incrementally changed (Kobayashi et al., 2015). It is clear that the effects of AFO alignment and resistance on ankle joint kinetics and the concomitant effects on the knee and hip kinematics require further study. Moreover, the potential effects of the AFO on hip, pelvis, trunk and center of mass have been suggested in previous studies (Cruz and Dhaher, 2009; Kobayashi et al., 2012a; Wang et al., 2007). The hypothesis that the alignment of an articulated AFO would influence knee joint kinematics and kinetics was supported by the results of this study. Hyperextension of the knee may be alleviated through improvement of ankle joint kinematics with the use of a properly aligned AFO while walking in individuals post-stroke.

All subjects used the same type and model of shoes in this study. Therefore, the potential influence of different shoe designs on gait was well controlled. A previous study that manipulated the alignment of an articulated AFO by adjusting shoe heel-height (i.e. shank-to-vertical angles of 5–7°) did not demonstrate significant differences in knee angles and moments due to alignment changes when compared to an articulated AFO whose plantarflexion stop angle was set at the ankle joint neutral position (i.e. 0° position) in individuals post-stroke (Fatone et al., 2009). The authors suggested that alignment changes used in their study based on shoe heel-height were not sufficient enough to influence knee joint kinematics and kinetics. In our study, the adjustment of ankle alignment was determined by the adjustment of the Triple Action® ankle joint. Mechanical testing of the AFO demonstrated a clear shift of the angle-moment hysteresis curves as the alignment of the AFO was incrementally changed toward dorsiflexion, qualifying the effect of alignment changes of the AFO (Fig. 2). These changes in AFO alignment significantly influenced knee joint angles and moments in this study for individuals post-stroke (Fig. 4, Table 3). An alternative means to clinically alter ankle alignment is to wedge the AFO extrinsically at the heel inside the shoe. A future study should investigate the effects of AFO joint alignment in combination with wedging at the heel on gait in individuals post-stroke.

This study investigated the effects of incremental changes in ankle joint alignment of an articulated AFO on gait, and it was not the aim of this study to investigate optimal alignment of the AFO. Optimal alignment of the AFO depends on what we aim to optimize, such as minimizing energy expenditure, maximizing patient preference, maximizing gait stability, or maximizing gait speed. However, the influence on ankle and knee joint gait parameters in response to the incremental changes in the alignment of the articulated AFO suggests that these gait parameters could be used as one of the potential parameters for optimization of the AFO. This study also investigated the effect of the sagittal alignment changes of the AFO on the sagittal gait parameters. A previous study in transtibial prosthetic alignment demonstrated that kinetic interactions occur between the sagittal and coronal planes when sagittal alignment of a prosthesis is incrementally changed (Kobayashi et al., 2012b). Triplanar control of the AFO is important because pathologic gait of individuals post-stroke is generally influenced in three planes (Loke, 2006). Therefore, the effects of an AFO should be explored not only in the sagittal plane but also in other planes in a future study.

The outcome of the clinical assessments (i.e. manual muscle testing and range of motion) suggests that the group of subjects who participated in this study were not very homogeneous (Table 1). Considering that stroke is a very heterogeneous insult, this group of subjects may represent the general population of individuals post-stroke. The clearly demonstrated influence of AFO alignment on ankle and knee joints in this non-homogeneous group is encouraging. However, only ten subjects participated in this study. A larger scale study is warranted to understand the general effects of the AFO alignment on gait in individuals post-stroke.

The immediate effects of the changes in the ankle alignment of an articulated AFO for individuals post-stroke were investigated in this study. Ankle foot orthoses are generally used by patients who benefit from them for long periods of time. Therefore, the long-term effects of AFOs need to be clarified. These effects could be investigated by comparing gait immediately after fitting and after extended use of the AFO for a period of time in the community (Nikamp et al., 2017). Clarifying the long-term effects of tuned AFO may provide more clinically relevant information on the impact of the AFO for individuals post-stroke.

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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.

References

Baumbach, S.F., Brumann, M., Binder, J., Mutschler, W., Regauer, M., Polzer, H., 2014. The influence of knee position on ankle dorsiflexion – a biometric study. BMC Musculoskeletal Disorders 15, 246.
Boone, D.A., Kobayashi, T., Chou, T.G., Arabian, A.K., Coleman, K.L., Orendurff, M.S., Zhang, M., 2013. Influence of malalignment on socket reaction moments during gait in amputees with transtibial prostheses. Gait Posture 37, 620–626.
Cruz, T.H., Dhaher, Y.Y., 2009. Impact of ankle-foot-orthosis on frontal plane behaviors post-stroke. Gait Posture 30, 312–316.
Fatone, S., Gard, S.A., Malas, B.S., 2009. Effect of ankle-foot orthosis alignment and foot-plate length on the gait of adults with poststroke hemiplegia. Arch. Phys. Med. Rehabil. 90, 810–818.
Gao, F., Carlton, W., Kapp, S., 2011. Effects of joint alignment and type on mechanical properties of thermoplastic articulated ankle-foot orthosis. Prosthet. Orthot. Int. 35, 181–189.

Kendall, F.P., McCreary, E.K., Provance, P.G., 1993. Muscles Testing and Function with Posture and Pain. Williams & Wilkins, Baltimore, Maryland.
Kerkum, Y.L., Buizer, A.I., van den Noort, J.C., Becher, J.G., Harlaar, J., Brehm, M.A., 2015. The effects of varying ankle foot orthosis stiffness on gait in children with spastic cerebral palsy who walk with excessive knee flexion. PloS one 10, e0142878.
Kobayashi, T., Leung, A.K., Akazawa, Y., Hutchins, S.W., 2011. Design of a stiffness-adjustable ankle-foot orthosis and its effect on ankle joint kinematics in patients with stroke. Gait Posture 33, 721–723.
Kobayashi, T., Leung, A.K., Akazawa, Y., Hutchins, S.W., 2012a. Effect of ankle-foot orthoses on the sagittal plane displacement of the center of mass in patients with stroke hemiplegia: a pilot study. Top. Stroke Rehab. 19, 338–344.
Kobayashi, T., Orendurff, M.S., Hunt, G., Lincoln, L.S., Gao, F., LeCursi, N., Foreman, K. B., 2017a. An articulated ankle-foot orthosis with adjustable plantarflexion resistance, dorsiflexion resistance and alignment: a pilot study on mechanical properties and effects on stroke hemiparetic gait. Med. Eng. Phys. 44, 94–101.
Kobayashi, T., Orendurff, M.S., Singer, M.L., Gao, F., Foreman, K.B., 2017b. Contribution of ankle-foot orthosis moment in regulating ankle and knee motions during gait in individuals post-stroke. Clin. Biomech. (Bristol, Avon) 45, 9–13.
Kobayashi, T., Orendurff, M.S., Zhang, M., Boone, D.A., 2012b. Effect of transtibial prosthesis alignment changes on out-of-plane socket reaction moments during walking in amputees. J. Biomech. 45, 2603–2609.
Kobayashi, T., Singer, M.L., Orendurff, M.S., Gao, F., Daly, W.K., Foreman, K.B., 2015. The effect of changing plantarflexion resistive moment of an articulated ankle-foot orthosis on ankle and knee joint angles and moments while walking in patients post stroke. Clin. Biomech. (Bristol, Avon) 30, 775–780.
Loke, M.D., 2006. Triplanar control dynamic response orthoses based on new concepts in lower limb orthotics. Phys. Med. Rehab. Clin. N. Am. 17, 181–202.
Nikamp, C.D., Buurke, J.H., van der Palen, J., Hermens, H.J., Rietman, J.S., 2017. Six-month effects of early or delayed provision of an ankle-foot orthosis in patients with (sub)acute stroke: a randomized controlled trial. Clin. Rehab. 31, 1616–1624.
Owen, E., 2010. The importance of being earnest about shank and thigh kinematics especially when using ankle-foot orthoses. Prosthet. Ortho. Int. 34, 254–269.
Schmalz, T., Blumentritt, S., Jarasch, R., 2002. Energy expenditure and biomechanical characteristics of lower limb amputee gait: the influence of prosthetic alignment and different prosthetic components. Gait Posture 16, 255–263.
Tyson, S.F., Sadeghi-Demneh, E., Nester, C.J., 2013. A systematic review and meta-analysis of the effect of an ankle-foot orthosis on gait biomechanics after stroke. Clin. Rehab. 27, 879–891.
Wang, R.Y., Lin, P.Y., Lee, C.C., Yang, Y.R., 2007. Gait and balance performance improvements attributable to ankle-foot orthosis in subjects with hemiparesis. Am. J. Phys. Med. Rehab. 86, 556–562.
Yamamoto, S., Hagiwara, A., Mizobe, T., Yokoyama, O., Yasui, T., 2005. Development of an ankle-foot orthosis with an oil damper. Prosthet. Ortho. Int. 29, 209–219.