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Technical note

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

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ABSTRACT

Mechanical properties of an articulated ankle–foot orthosis (AFO) are closely related to gait performance in individuals post-stroke. This paper presents a pilot study on the mechanical properties of a novel articulated AFO with adjustable plantarflexion resistance, dorsiflexion resistance and alignment, and its effect on ankle and knee joint kinematics and kinetics in an individual post-stroke during gait. The mechanical properties of the AFO were quantified. Gait analysis was performed using a 3D motion capture system with a split-belt instrumented treadmill under 12 different settings of the mechanical properties of the AFO [i.e. 4 plantarflexion resistances (P1 < P4), 4 dorsiflexion resistances (D1 < D4), 4 initial alignments (A1 < A4)]. The AFO demonstrated systematic changes in moment–angle relationship in response to changes in AFO joint settings. The gait analysis demonstrated that the ankle and knee angle and moment were responsive to changes in the AFO joint settings. Mean ankle angle at initial contact changed from −0.86° (P1) to 0.91° (P4) and from −1.48° (A1) to 4.45° (A4), while mean peak dorsiflexion angle changed from 12.01° (D1) to 6.40° (D4) at mid-stance. The novel articulated AFO appeared effective in influencing lower-limb joint kinematics and kinetics of gait in the individual post-stroke.

1. Introduction

Mechanical properties of an ankle–foot orthosis (AFO) affect gait in individuals post-stroke. The resistance of an AFO (resistance to plantarflexion and the resistance to dorsiflexion) is a key mechanical characteristic that influences sagittal motion of the ankle and knee joints during gait [1,2]. Alignment is another important characteristic of the AFO that determines the ankle joint angle where the resistances are initiated.

AFOs are generally categorized into two types: an articulated AFO that has a mechanical joint and a non-articulated AFO that does not have the joint. The importance of tuning the mechanical properties of an articulated AFO to improve gait in individuals with neuromuscular disorders has been increasingly recognized by researchers and clinicians [2–4]. In non-articulated AFOs, resistance and alignment are usually determined during the fabrication, fitting and adjustment processes which can affect stiffness and support in the planes of control; therefore, errors in the initial resistance or alignment are difficult to change once the AFO has been fabricated. A resistance and alignment adjustable AFO joint would minimize this trial-and-error process by simplifying the adjustment of sagittal plane mechanical characteristics post fabrication, and would thus save time for both clinicians and patients. Therefore, an articulated AFO with the adjustable joint would be beneficial.

The resistance and alignment of an articulated AFO affect gait during stance and swing in gait. Resistance functions like a gradual

Abbreviations: DF, dorsiflexion in angle and dorsiflexor in moment; EX, extensor in moment, FX, flexion in angle and flexor in moment; PF, planarflexion in angle and plantarflexor in moment.

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brake on the ankle joint while the AFO is plantarflexed or dorsiflexed. Traditionally the primary function of AFOs is considered to provide sufficient toe clearance during swing, but AFOs' functions during stance have been the focus of recent studies. Plantarflexion resistance plays an important role on the ankle joint mainly in the 1st rocker of stance to achieve heel strike and swing to enable foot clearance by complementing the function of impaired dorsiflexor muscles in individuals post-stroke. An articulated AFO generates more plantarflexion resistance at the 1st rocker than swing in a gait cycle [5]. Therefore, the plantarflexion resistance of the articulated AFO should be tuned to optimize the 1st rocker. Electromyography of the gastrocnemius muscle during the 1st rocker was also reported to be significantly reduced to avoid excessive stretch reflex in individuals post-stroke when using an AFO with plantarflexion resistance [6]. Plantarflexion resistance also affects on the knee joint mainly in the 1st and 2nd rocker of stance. Knee recurvatum was reduced in individuals post-stroke by increasing plantarflexion resistance of an articulated AFO [7]. While the effects of plantarflexion resistance of articulated AFOs have been investigated widely, there is paucity of research investigating the effects of dorsiflexion resistance and alignment of articulated AFOs. One study showed that the dorsiflexion resistance affects the amount of dorsiflexion angles at the 2nd rocker in individuals post-stroke [1]. Clinically, an increase in dorsiflexion resistance is also expected to reduce knee flexion if a patient is walking in a flexed-knee pattern. Alignment of a non-articulated AFO was shown to affect knee kinematics and kinetics in individuals post-stroke at the 1st rocker [8]. Generally, an AFO alignment in a dorsiflexed position increases knee flexion and decreases internal knee flexion moment. In gait energy consumption, it was suggested that individuals with neuromuscular disorders could potentially reduce gait energy cost when AFO's stiffness (i.e. change of resistance/change of angles) is tuned appropriately [9] and energy storing AFOs could assist their ankle work at the 3rd rocker [10]. Therefore, the mechanical properties of an articulated AFO should be tunable to optimize gait in individuals post-stroke.

There are some AFO joints that are commercially available which allow tuning of mechanical properties constructed with oil-damper [11] or springs [4]. The use of tunable articulated AFOs is expected to be more common in future clinical practice as the benefits of clinical optimization become more evident through research. However, to our knowledge, no studies have comprehensively investigated the effects of systematic changes in the three tunable parameters of an AFO joint in the sagittal plane on lower limb joint kinematics and kinetics in individuals post-stroke. These tunable parameters include plantarflexion resistance, dorsiflexion resistance and alignment of an articulated AFO. Data on how each joint parameter would affect gait of individuals post-stroke would benefit clinicians when they are tuning the AFO. Therefore, the aim of this pilot study was to quantify the mechanical properties (i.e. moment–angle relationships) of a novel articulated AFO that can independently adjust plantarflexion resistance, dorsiflexion resistance and alignment, and to explore the effect of changing its mechanical properties on ankle and knee joint kinematics and kinetics during gait in an individual post-stroke.

2. Methods

2.1. Ankle–foot orthosis (AFO)

A prototype of the Becker Triple Action ankle joint (Troy, MI, USA) attached to a custom polypropylene AFO was tested in this study (Fig. 1a). This orthotic ankle joint permits the independent adjustment and metering of AFO plantarflexion resistance, dorsiflexion resistance and null torque alignment. The preload and range of motion of the plantarflexion resistance are changed by turning the plantarflexion resistance adjustment screw (Fig. 1b). The ankle joint's plantarflexion resist mechanism is comprised of a clevis style hinge; the stirrup articulates about a pivot bushing through the component body. The domed shape of the stirrup head creates a normal contact between the ball bearing and the stirrup head to transfer the spring force along the axis of the coil spring (spring constant: 50 N/mm) to the stirrup. There is a pin inside the spring to limit its compression to increase the cycle life of the spring and to facilitate locking of the component. The dorsiflexion resistance mechanism employs two springs in a parallel configuration: a low torque spring (spring constant: 50 N/mm) with large range of motion, and a high torque spring (spring constant: 100 N/mm) with smaller range of motion (Fig. 1b). Turning the dorsiflexion resistance adjustment screw alters the recruitment angle of the high torque dorsiflexion resistance spring. Adjustment may also change the range of motion and alter the preload of the spring for some settings. The independent alignment adjustment changes the component null torque angle (i.e. initial alignment of the AFO; the neutral angle between the plantarflexion and dorsiflexion resistances) (Fig. 1c). The AFO was custom fabricated for a participant in the study using 4.8 mm thick polypropylene homopolymer.

2.2. Mechanical testing of the AFO

A custom motorized mechanical testing device was used to measure mechanical properties of the AFO [12]. This device was comprised of an optical encoder (Danaher motion Inc., USA) and an inline uniaxial torque sensor (TRT-500, measurement range: 500 in lb or 56.5 N m, Transducer Tech Inc., USA) to quantify the moment–angle characteristics of the AFO within a torque range of $\pm$ 30 N m or an angle range of $\pm 15°$. The dorsiflexion resistance and angle were defined positive in this study. Both motor speed and direction of rotation were controlled by a motor drive. Data acquisition was done using an NI PCI-6221 M Series DAQ, while a graphical user interface (GUI) that can set rotational speed, maximum torque as well as range of motion and display moment–angle curves was built using LabVIEW (National Instrument Inc., USA).

The mechanical properties of the AFO were quantified under 12 conditions (Table 1): (1) four plantarflexion resistances with constant dorsiflexion resistance setting at initial alignment of 2° of dorsiflexion ($P1 < P2 < P3 < P4$) (Fig. 2(a)), (2) four dorsiflexion resistances with constant plantarflexion resistance setting at initial alignment of 2° of dorsiflexion ($D1 < D2 < D3 < D4$) (Fig. 3(a)), and (3) four initial alignments (*A1*: 2° of plantarflexion, *A2*: 2° of dorsiflexion, *A3*: 4° of dorsiflexion and *A4*: 6° of dorsiflexion) with constant plantarflexion and dorsiflexion resistance setting (Fig. 4(a)). The rotational center of the AFO ankle joints was carefully aligned to the rotational center of the motor shaft of the testing device. The AFO was mounted to the device by clamping the AFO footplate to the flat articulating platform of the testing device using a C-clamp and attaching the tibial section of the AFO to a plaster surrogate leg using the AFO's tibial Velcro® strap. Subsequently, the AFO was automatically plantarflexed and dorsiflexed at a rotational speed of 10°/s for 60 s. The test was repeated three times per condition. The resistive moment and corresponding angular positions of the AFO were measured with the torque sensor and the encoder at a sampling frequency of 1000 Hz. A fourth-order zero-lag low pass Butterworth filter with a cutoff frequency of 5 Hz was used to filter the data. The mean of the hysteresis loop (i.e. the moment–angle relationship of the AFO around the ankle joint) was calculated and plotted for each condition.

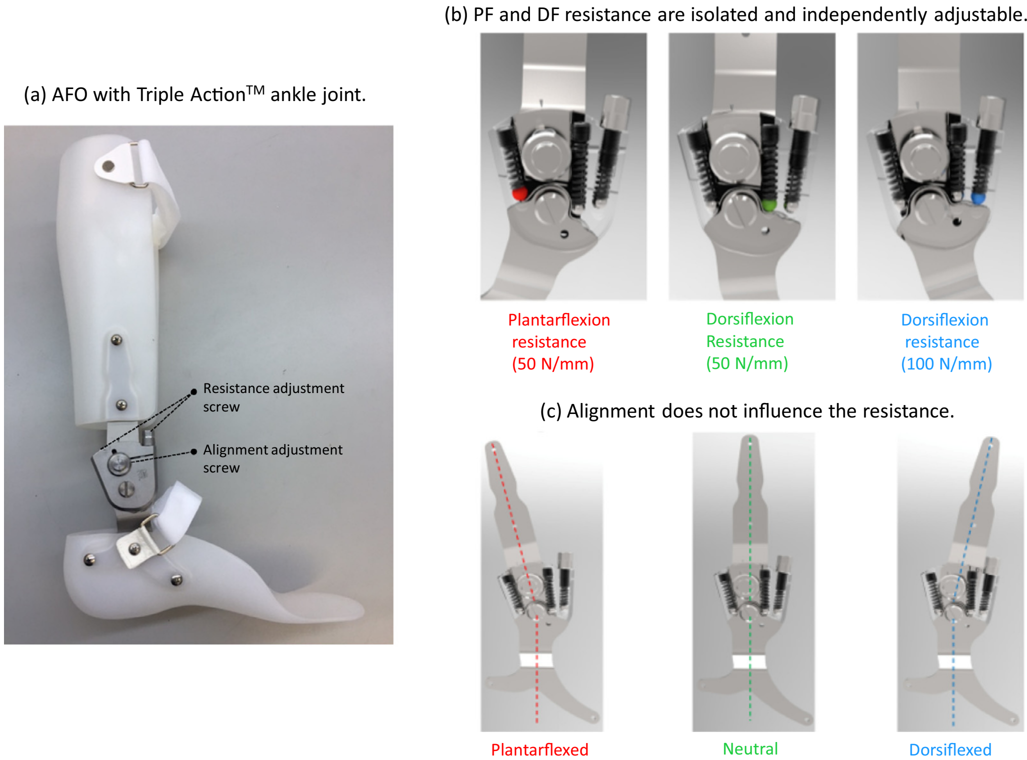

Fig. 1. (a) An articulated AFO with plantarflexion resistance, dorsiflexion resistance and alignment adjustable ankle joint (Triple Action). A hex key is used to adjust the plantarflexion and dorsiflexion resistance adjustment screws, while a hex spanner is used to adjust the hex head alignment adjustment bolt. (b) Mechanisms of isolated and independent adjustment of plantarflexion (PF) and dorsiflexion (DF) resistance of the joint. (c) Mechanisms of independent adjustment of alignment of the joint. Images are used by permission from Becker Orthopedic.

Table 1
Twelve settings of the articulated AFO joint tested in the study.

Setting	Plantarflexion resistance	Dorsiflexion resistance	Alignment
P1	Low	Medium high	2° of dorsiflexion
P2	Medium low	Medium high	2° of dorsiflexion
P3	Medium high	Medium high	2° of dorsiflexion
P4	High	Medium high	2° of dorsiflexion
D1	Medium low	Low	2° of dorsiflexion
D2	Medium low	Medium low	2° of dorsiflexion
D3	Medium low	Medium high	2° of dorsiflexion
D4	Medium low	High	2° of dorsiflexion
A1	Medium low	Medium high	2° of plantarflexion
A2	Medium low	Medium high	2° of dorsiflexion
A3	Medium low	Medium high	4° of dorsiflexion
A4	Medium low	Medium high	6° of dorsiflexion

Note: "Low", "Medium low", "Medium high" and "High" are only indicators of distinctive levels of resistance generated by the AFO. Therefore, the amount of resistance generated under "High" in plantarflexion resistance and "High" in dorsiflexion resistance is not equal.

2.3. Gait analysis

A 50 year-old female subject (body height: 1.73 m; body mass: 77 kg), who was post-stroke for 5.5 years, consented to participate in this Institutional Review Board approved study. She presented with left-sided hemiparesis due to an ischemic stroke. Manual muscle testing on the affected limb using medical research council (MRC) scale revealed that her ankle dorsiflexion strength was 1+/5 and her ankle plantarflexion strength was 4/5. Both knee extension and flexion strength was 5/5. She could not actively dorsiflex when the ankle joint was in more than 10° of plantarflexion and thus walked with foot-drop in a flexed-knee pattern. The subject was fit with the AFO and the orthosis was kinematically optimized using observational gait analysis and subjective feedback to determine the optimal AFO settings. The optimized settings (i.e. condition P2 shown in Fig. 2(a)) served as baseline setting during the motion trials. During the trials, the ankle joint setting of interest was changed while the other two settings were held at their optimized settings. Gait analysis was performed by placing reflective markers 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)]. On the shank and foot of the affected limb, the markers were placed directly on the AFO. A rigid cluster was secured to the upright of the AFO and used for dynamic tracking. The participant was secured in a safety harness and asked to walk on a split-belt instrumented treadmill (Bertec corporation, Columbus, OH, USA) while wearing the AFO on the affected leg (left) at a self-selected speed of 0.36 m/s. The participant walked with the AFO under the 12 conditions: P1–P4, D1–D4 and A1–A4. The participant was given

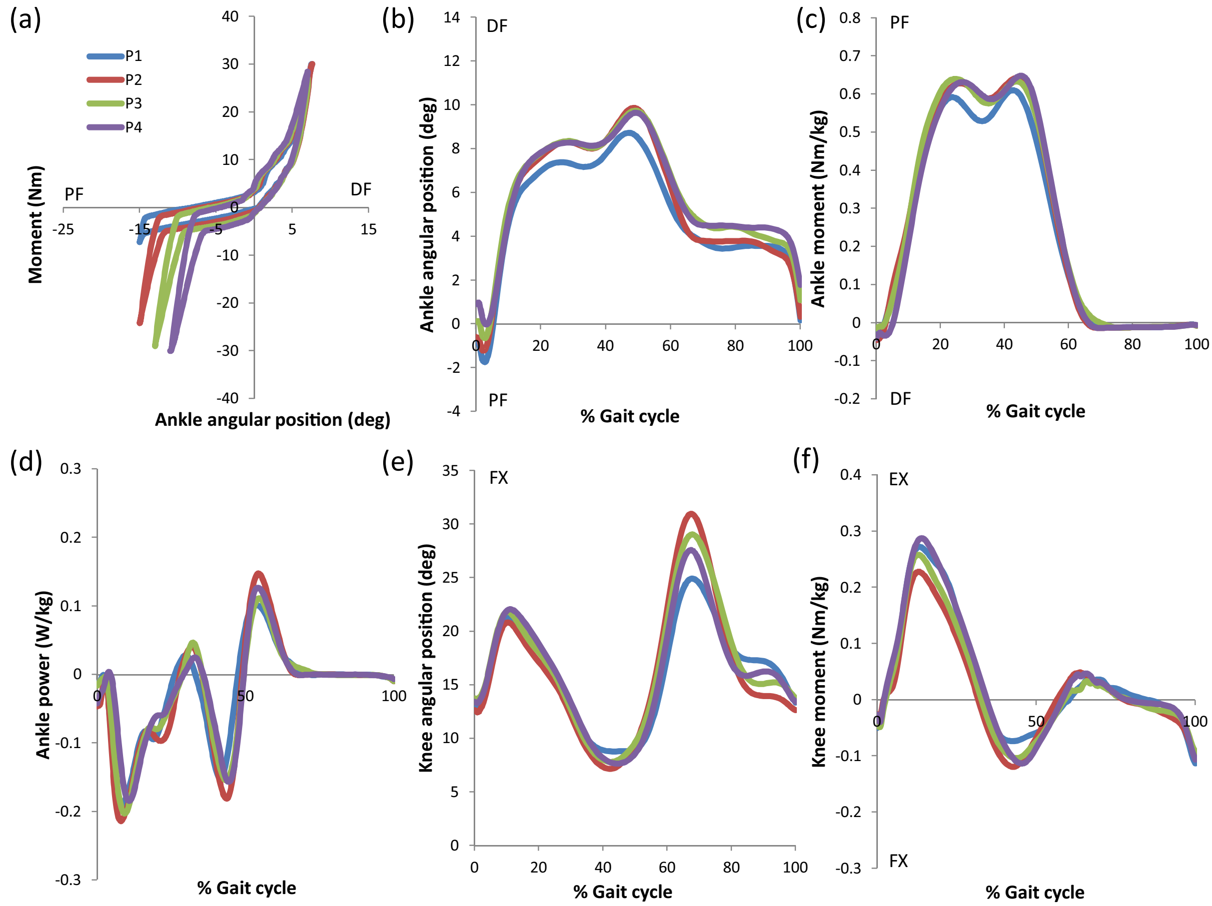

Fig. 2. (a) Moment–angle relationship of the AFO when planarflexion resistance was changed at 4 levels (P1 < P2 < P3 < P4). (b) Effect of changing plantarflexion resistance of the AFO on the mean ankle angular positions in the sagittal plane. (c) Effect of changing plantarflexion resistance of the AFO on the mean ankle moment in the sagittal plane. (d) Effect of changing plantarflexion resistance of the AFO on the mean ankle power in the sagittal plane. (e) Effect of changing plantarflexion resistance of the AFO on the mean knee angular positions in the sagittal plane. (f) Effect of changing plantarflexion resistance of the AFO on the mean knee moment in the sagittal plane.

a short acclimatization period to practice walking on the treadmill before data collection. Gait data were collected for 10 gait cycles of the affected limb wearing the AFO using a Vicon 10-camera motion analysis system (Vicon Motion Systems, Oxford, UK), and the Bertec instrumented treadmill at a rate of 200 Hz.

Data were recorded and synchronized using Vicon Nexus (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 and force platform data, respectively. The ankle joint angles, moments and power as well as knee joint angles and moments of the affected limb were averaged and normalized over 10 gait cycles for each of the 12 test condition of the orthosis.

3. Results

3.1. Mechanical properties of the AFO

Moment–angle relationships of the AFO when the settings of plantarflexion resistance, dorsiflexion resistance and alignment were altered are shown in Figs. 2(a), 3(a) and 4(a), respectively. The AFO demonstrated systematic changes in moment–angle relationship in response to changes in these settings.

3.2. Effect of changing plantarflexion resistance of the AFO

Effects of changing AFO plantarflexion resistance on the ankle and knee joint kinematics and kinetics are shown in Fig. 2. These data showed some systematic effects on ankle angle at initial contact (P1: $-0.86 \pm 0.61°$, P2: $-0.59 \pm 0.68°$, P3: $0.10 \pm 0.40°$, P4: $0.91 \pm 0.41°$ at 0% gait cycle) and moment (i.e. showing a more normal dorsiflexor moment pattern as plantarflexion resistance increased) from initial contact to loading response during gait. Peak knee flexion angles during stance were: P1: $21.33 \pm 1.10°$, P2: $20.81 \pm 0.86°$, P3: $21.81 \pm 1.36°$, P4: $22.06 \pm 1.16°$, while minimum knee flexion angles during stance were: P1: $8.79 \pm 1.45°$, P2: $7.16 \pm 1.81°$, P3: $7.82 \pm 2.21°$, P4: $7.66 \pm 1.27°$. Peak knee extensor moments were: P1: 0.27 ± 0.03 N m/kg, P2: 0.23 ± 0.03 N m/kg, P3: 0.26 ± 0.06 N m/kg, P4, 0.29 ± 0.04 N m/kg, while peak flexor moments were: P1: -0.07 ± 0.04 N m/kg, P2: -0.12 ± 0.03 N m/kg, P3: -0.10 ± 0.04 N m/kg, P4: -0.11 ± 0.04 N m/kg.

3.3. Effect of changing dorsiflexion resistance of the AFO

Effects of changing AFO dorsiflexion resistance on the ankle and knee joint kinematics and kinetics are shown in Fig. 3. These data showed some systematic effects on ankle angle at mid-stance (peak dorsiflexion angle: D1: $12.01 \pm 0.68°$, D2: $10.79 \pm 0.73°$, D3: $9.9 \pm 0.62°$, D4: $6.40 \pm 0.42°$) and ankle power at terminal stance (peak released power: D1: 0.21 ± 0.05 W/kg, D2: 0.14 ± 0.09 W/kg, D3: 0.15 ± 0.04 W/kg, D4: 0.06 ± 0.02 W/kg) during gait. Peak knee flexion angles during stance were: D1: $20.45 \pm 1.32°$, D2: $21.26 \pm 1.12°$, D3: $20.81 \pm 0.86°$, D4: $20.37 \pm 0.95°$, while minimum knee flexion angles during stance were: D1: $10.21 \pm 1.53°$, D2: $9.13 \pm 1.01°$, D3: $7.16 \pm 1.81°$, D4: $4.28 \pm 1.37°$. Peak knee extensor mo-

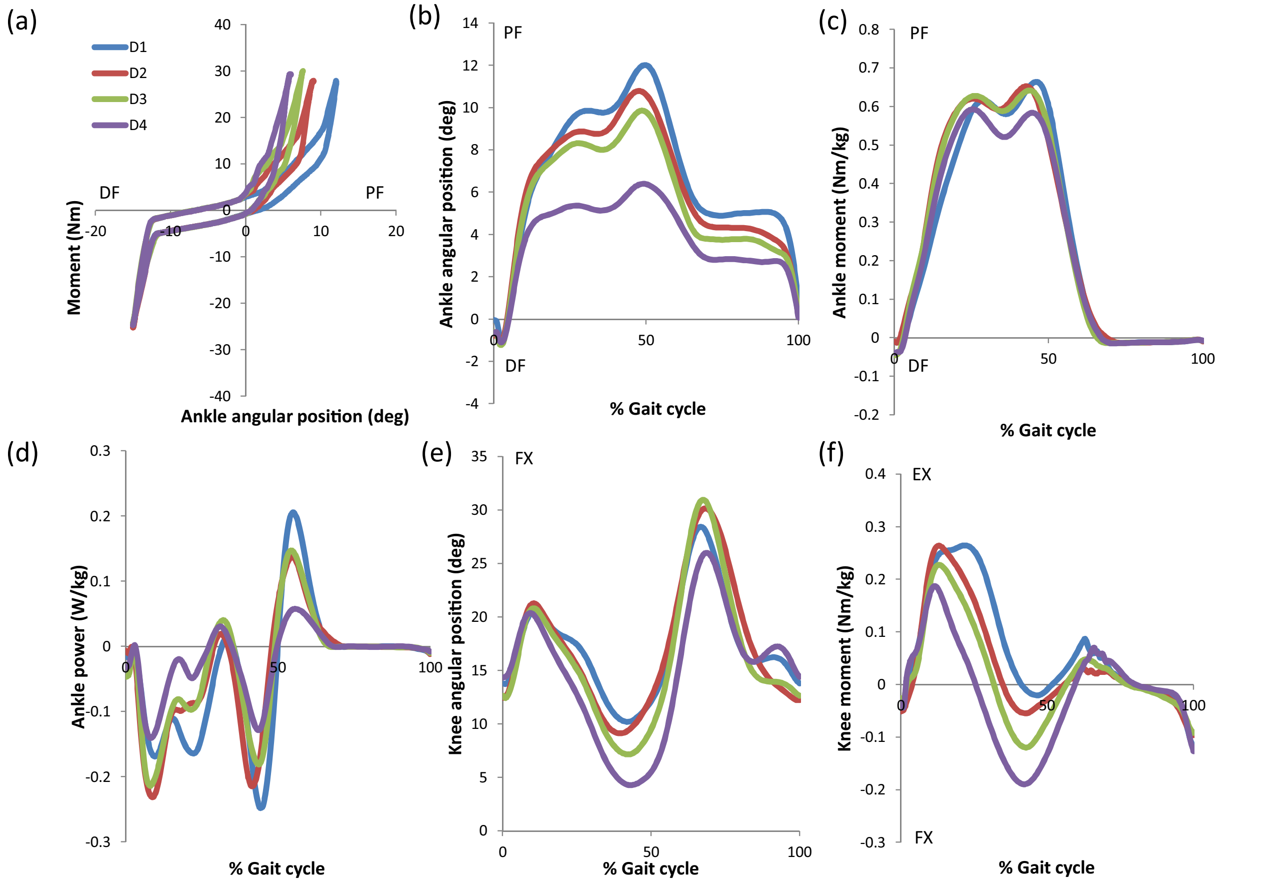

Fig. 3. (a) Moment–angle relationship of the AFO when dorsiflexion resistance was changed at 4 levels (D1 < D2 < D3 < D4). (b) Effect of changing dorsiflexion resistance of the AFO on the mean ankle angular positions in the sagittal plane. (c) Effect of changing dorsiflexion resistance of the AFO on the mean ankle moment in the sagittal plane. (d) Effect of changing dorsiflexion resistance of the AFO on the mean ankle power in the sagittal plane. (e) Effect of changing dorsiflexion resistance of the AFO on the mean knee angular positions in the sagittal plane. (f) Effect of changing dorsiflexion resistance of the AFO on the mean knee moment in the sagittal plane.

ments were: D1: 0.26 ± 0.04 N m/kg, D2: 0.26 ± 0.06 N m/kg, D3: 0.23 ± 0.03 N m/kg, D4, 0.19 ± 0.02 N m/kg, while peak knee flexor moments were: D1: -0.02 ± 0.04 N m/kg, D2: -0.05 ± 0.03 N m/kg, D3: -0.12 ± 0.03 N m/kg, D4: -0.19 ± 0.03 N m/kg.

3.4. Effect on changing alignment of the AFO

Effects of changing AFO alignment on the ankle angle, moment and power are shown in Fig. 4. These data showed some systematic effects on ankle angle at initial contact (A1: $-1.48 \pm 0.47°$, A2: $-0.59 \pm 0.68°$, A3: $2.74 \pm 0.27°$, A4: $4.45 \pm 0.27°$ at 0% gait cycle) and mid-stance (A1: $5.41 \pm 0.30°$, A2: $9.86 \pm 0.62°$, A3: $11.31 \pm 0.45°$, A4: $12.49 \pm 0.31°$ as peak dorsiflexion angle) and moment (i.e. showing a more normal dorsiflexor moment pattern as alignment increased) from initial contact to loading response during gait. Peak knee flexion angles during stance were: A1: $19.37 \pm 1.49°$, A2: $20.81 \pm 0.86°$, A3: $22.05 \pm 0.69°$, A4: $22.14 \pm 1.00°$, while minimum knee flexion angles during stance were: A1: $0.30 \pm 2.39°$, A2: $7.16 \pm 1.81°$, A3: $8.62 \pm 1.85°$, A4: $11.37 \pm 1.18°$. Peak knee extensor moments were: A1: 0.17 ± 0.04 N m/kg, A2: 0.23 ± 0.03 N m/kg, A3: 0.29 ± 0.04 N m/kg, A4, 0.35 ± 0.04 N m/kg, while peak flexor moments were: A1: -0.24 ± 0.04 N m/kg, A2: -0.12 ± 0.03 N m/kg, A3: -0.08 ± 0.03 N m/kg, A4: -0.00 ± 0.04 N m/kg.

4. Discussion

The aim of this pilot study was to investigate the mechanical properties of the novel articulated AFO that can independently adjust plantarflexion resistance, dorsiflexion resistance and alignment, and to explore the effect of changing AFO mechanical properties on ankle and knee joint angle, moment and power during gait in an individual post-stroke. The AFO showed a systematic change in moment–angle relationship when plantarflexion resistance, dorsiflexion resistance and alignment were changed. The ankle and knee joint kinematics and kinetics also showed some systematic changes in response to changes in mechanical properties of the AFO during gait. The knee moment appeared more responsive than the knee angle to the changes in AFO's mechanical properties.

Comparison of the data collected form the participant with the AFO to normal data in ankle and knee joint kinematics and kinetics during gait [13] is shown in Fig. 5. The ankle angle and moment were responsive to changes in plantarflexion resistance, dorsiflexion resistance and alignment. Improvement in heel rocker (1st rocker) was revealed by more normal dorsiflexor moment pattern when plantarflexion resistance and dorsiflexion angle in alignment were increased (Figs. 2(c), 4(c), 5(a)). Preservation of heel rocker using an AFO was shown to increase gait speed in individuals with hemiplegia [14]. Our study suggests that both plantarflexion resistance and alignment of the AFO would need to be adjusted in order to optimize the heel rocker during gait. The participant walked in a flexed knee pattern. Both knee angle and moment were systematically responsive to the dorsiflexion resistance and alignment changes of the AFO (Figs. 3(e)–(f), 4(e)–(f), and 5(b)).

Based on observational gait analysis using foot to floor contact angle and knee stability in early stance as visual indicators of ankle and knee kinematics, condition P2 was selected as optimal setting. The participant's preference for these settings was also a

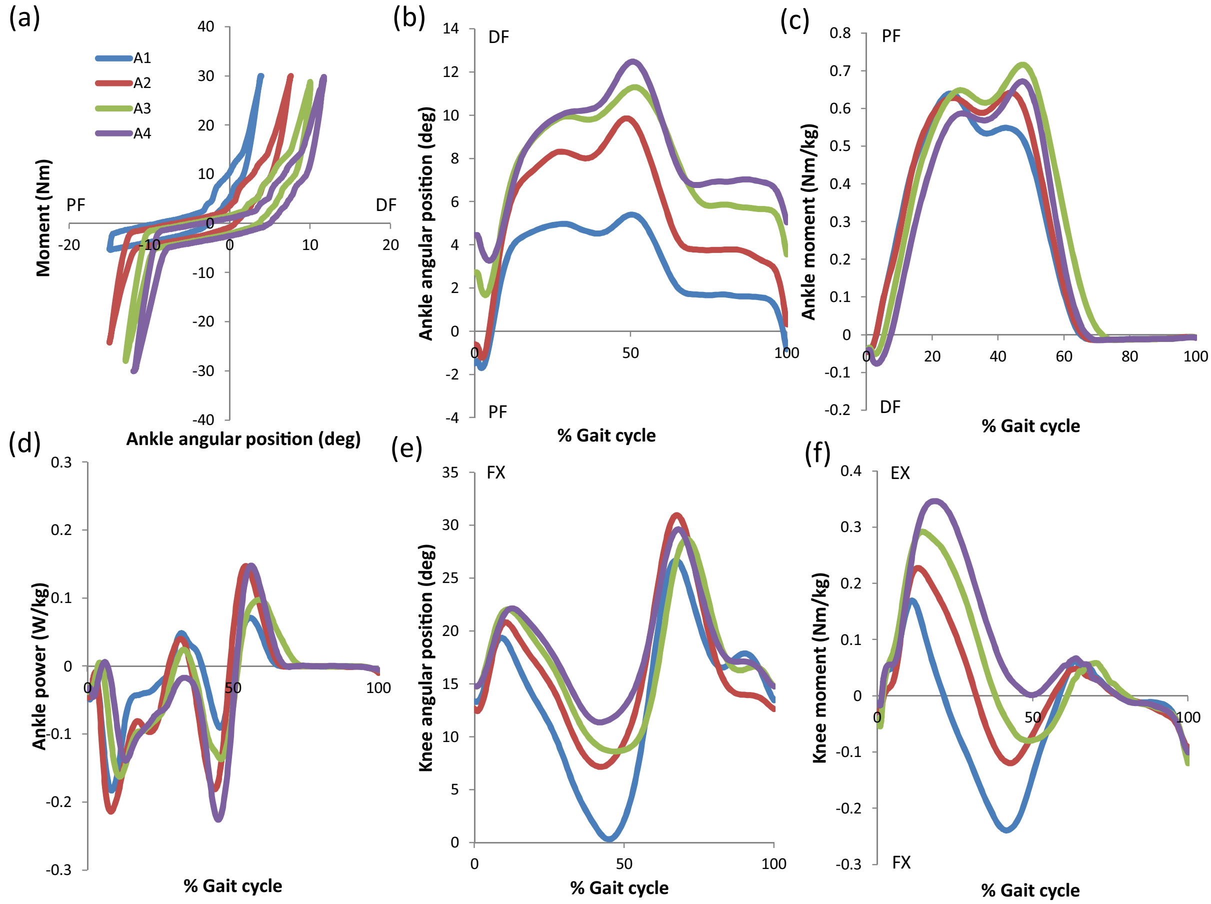

Fig. 4. (a) Moment–angle relationship of the AFO when initial alignment was changed at 4 levels [A1 (2° PF) < A2 (2° DF) < A3 (4° DF) < A4 (6° DF)]. (b) Effect of changing initial alignment of the AFO on the mean ankle angular positions in the sagittal plane. (c) Effect of changing initial alignment of the AFO on the mean ankle moment in the sagittal plane. (d) Effect of changing initial alignment of the AFO on the mean ankle power in the sagittal plane. (e) Effect of changing alignment of the AFO on the mean knee angular positions in the sagittal plane. (f) Effect of changing alignment of the AFO on the mean knee moment in the sagittal plane.

determining factor. Kinetic data showed that the minimum knee extensor moment was at condition P2, which suggested the knee was most stable under this condition (Fig. 2(f)). Kinematic and kinetic data showed that heel strike was in a dorsiflexed position with a more normal dorsiflexor moment in early stance under condition P4 (Fig. 2(c) and (d)). However, kinetic data also showed that the knee extensor moment was maximum at condition P4, which suggested the knee was most unstable under this condition (Fig. 2(f)). These outcomes suggest the benefit of measuring both kinematic and kinetic data in the AFO optimization process in addition to only performing observational gait analysis. However, patient preference and comfort may be a significant contributing factor in the selection of optimal AFO mechanical characteristics.

The peak released ankle power was reduced as the dorsiflexion resistance was increased (Fig. 3(d)), which is consistent with findings of a previous study [4]. This result suggests that toe rocker (3rd rocker) was somewhat impeded due to restriction in ankle rocker (2nd rocker), shown by reduction in range of motion toward dorsiflexion (Fig. 3(b)), resulting from increased dorsiflexion resistance. These data suggest that dorsiflexion resistance of the AFO might not be beneficial for this individual. However, dorsiflexion resistance may be beneficial for some individuals whose knee joints are unstable in stance during gait. It is generally known that the plantarflexion resistance of the AFO is beneficial for some individuals with a hyperextended knee [7]. Anecdotal evidence suggests that the influence of AFO dorsiflexion resistance may be functionally beneficial with significant gastrocnemius/soleus muscle weakness or quadriceps insufficiency; however, this requires further study.

There are some limitations in this study. First, this is a pilot study with a single individual post-stroke. Although the data presented in this study demonstrated clear effects of the mechanical properties of the novel AFO joint, it is necessary to confirm this effect in a larger scale study. Second, the participant walked on the split-belt instrumented treadmill. While walking on the treadmill enabled the participant to walk at the same gait speed under all the conditions, the gait on the treadmill might differ from the gait on the ground. Third, this study focused on the ankle and knee joint kinematics and kinetics. However, the AFO may also affect the hip, pelvis and trunk. Finally, the immediate effects of the mechanical properties of the AFO were tested in this study and long-term effect should also be investigated in a future study.

Convenient adjustment of an AFO's mechanical properties in a clinical setting is advantageous. Currently there are some AFO joints available on the market whose mechanical properties are tunable [4,11]. However, the evidence necessary to maximize the benefit of these AFO joints is still limited. This study showed that both resistance and alignment of the articulated AFO influenced kinematics and kinetics of the ankle and knee joints in different ways during gait in the individual post-stroke. The benefit of tuning mechanical properties of the AFO on gait needs further investigation in larger scale studies with individuals with neuromuscular disorders.

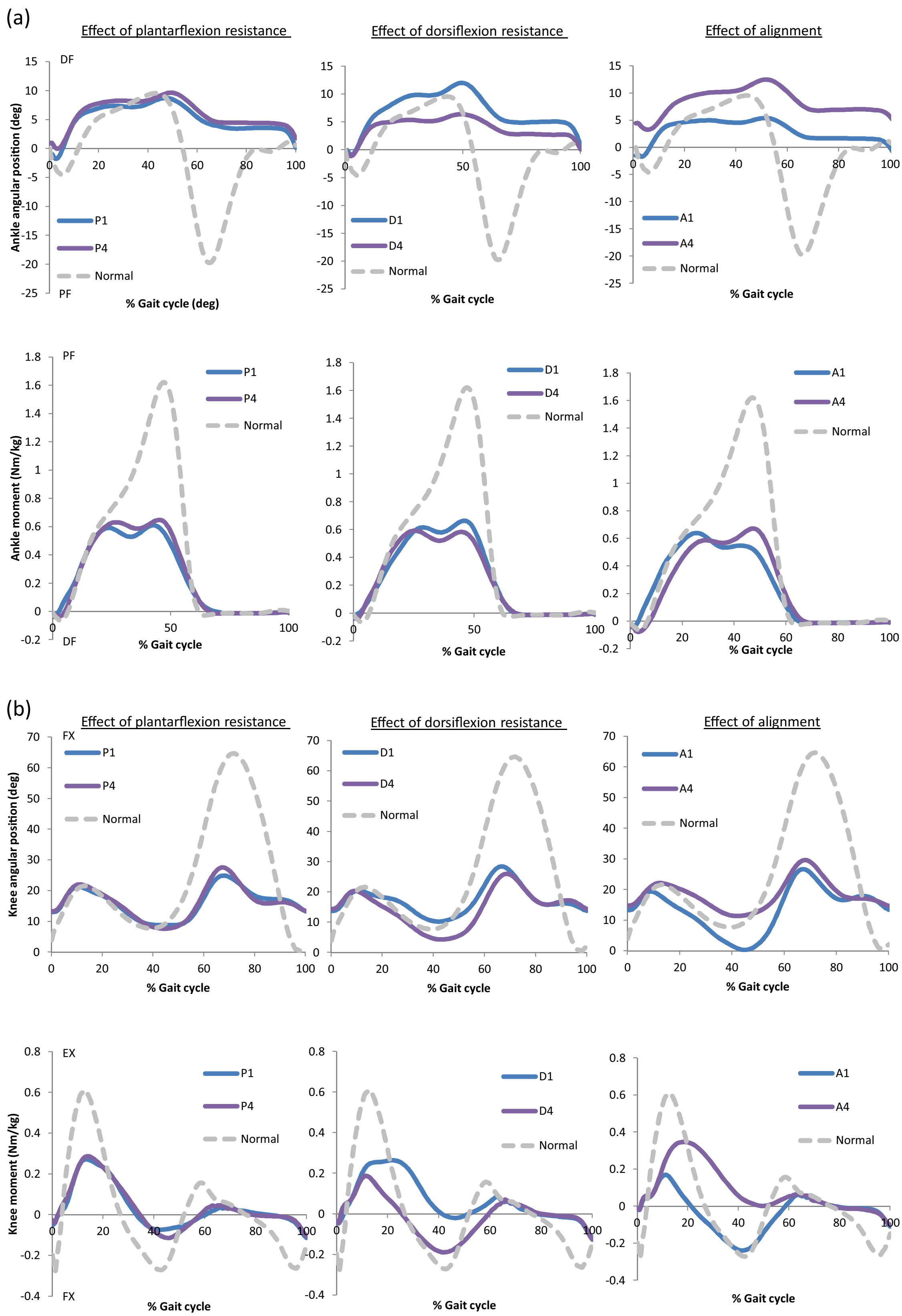

Fig. 5. Comparison of the gait data collected from the individual post-stroke to normal gait data in (a) ankle and (b) knee joint kinematics and kinetics. The normal data were taken from Winter [13]. Conditions 1 and 4 of plantarflexion resistance, dorsiflexion resistance and alignment were compared to the normal data.

Ethical approval

This study was approved by University of Utah Institutional Review Board (IRB_00062924).

Conflict of interest

T. Kobayashi, M.S. Orendurff and L.S. Lincoln are/were employees of Orthocare Innovations. N. LeCursi works for Becker Orthopedic, manufacture of the AFO joint (Triple Action^{TM}) used in this study.

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