International Journal of Cardiology 407 (2024) 131992

Contents lists available at ScienceDirect

International Journal of Cardiology

journal homepage: www.elsevier.com/locate/ijcard

Enhancing walking performance in patients with peripheral arterial disease: An intervention with ankle-foot orthosis

Farahnaz Fallahtafti [a], Kaeli Samson [b], Zahra Salamifar [a], Jason Johanning [c,d], Iraklis Pipinos [c,d], Sara A. Myers [a,d,*]

[a] Department of Biomechanics, University of Nebraska at Omaha, 6160 University Drive South, Omaha, NE 68182, USA
[b] Department of Biostatistics, University of Nebraska Medical Center, 984375 Nebraska Medical Center, Omaha, NE 68198-4375, USA
[c] Department of Surgery, University of Nebraska Medical Center, Omaha, NE 68105, USA
[d] Department of Surgery and VA Research Service, VA Nebraska-Western Iowa Health Care System, Omaha, NE 68105, USA

ARTICLE INFO

Keywords:
Peripheral artery disease
Variability
Intervention
Ankle foot orthoses

ABSTRACT

Lower extremity peripheral artery disease (PAD) is a cardiovascular condition manifesting from narrowed or blocked arteries supplying the legs. Gait is impaired in patients with PAD. Recent evidence suggests that walking with carbon fiber ankle foot orthoses (AFOs) can improve patient mobility and delay claudication time. This study aimed to employ advanced biomechanical gait analysis to evaluate the impact of AFO intervention on gait performance among patients with PAD. Patients with claudication had hip, knee, and ankle joint kinetics and kinematics assessed using a cross-over intervention design. Participants walked over the force platforms with and without AFOs while kinematic data was recorded with motion analysis cameras. Kinetics and kinematics were combined to quantify torques and powers during the stance period of the gait cycle. The AFOs effectively reduced the excessive ankle plantar flexion and knee extension angles, bringing the patients' joint motions closer to those observed in healthy individuals. After 3 months of the AFO intervention, the hip range of motion decreased, likely due to changes occurring within the ankle chain. With the assistance of the AFOs, the biological power generation required from the ankle and hip during the push-off phase of walking decreased. Wearing AFOs resulted in increased knee flexor torque during the loading response phase of the gait. Based on this study, AFOs may allow patients with PAD to maintain or improve gait performance. More investigation is needed to fully understand and improve the potential benefits of ankle assistive devices.

1. Introduction

The narrowing or blockages of the leg arteries resulting from atherosclerosis are responsible for developing lower extremity peripheral artery disease (PAD) [1]. As a result of PAD, the working leg muscles do not get an adequate oxygen supply during physical activity, and claudication occurs. Claudication is a pain in the legs that is brought on by physical activity and relieved with rest [2]. In the United States, approximately 25% of the population will experience PAD during their life [3].

PAD has a detrimental effect on walking performance even before the onset of claudication. During walking, reduced lower extremity joint powers [4], abnormal ankle and hip joint torques [5,6], and decrements in spatiotemporal gait parameters [7,8] were observed in patients with PAD compared to healthy counterparts. Changes in gait performance are likely driven by changes in muscle structure and function caused by PAD and its associated ischemia/reperfusion cycles. Smaller muscle area, increased muscle fat, and impaired energy production and utilization [9,10] are some of the ischemia-related adverse changes found in the calf skeletal muscles of patients with PAD [11]. Muscle changes manifest as reduced plantar flexor muscle strength in patients with PAD [12,13] and reduced torque and power during walking. We estimate that average reductions in lower limb joint torques during each phase of stance from all of our previous studies are as follows: a 24% decrease in hip power generation during early stance, a 39% decrease in knee power generation during midstance, and a 26% reduction in ankle plantarflexor power generation during late stance, compared with healthy controls [4,14,15]. Because of these reductions, individuals with PAD walk slower and can walk a shorter distance compared to individuals without PAD.

* Corresponding author at: University of Nebraska at Omaha, 6160 University Drive, Omaha, NE, USA.
E-mail addresses: ffallahtafti@unomaha.edu (F. Fallahtafti), samyers@unomaha.edu (S.A. Myers).

https://doi.org/10.1016/j.ijcard.2024.131992
Received 8 December 2023; Received in revised form 20 March 2024; Accepted 22 March 2024
Available online 23 March 2024

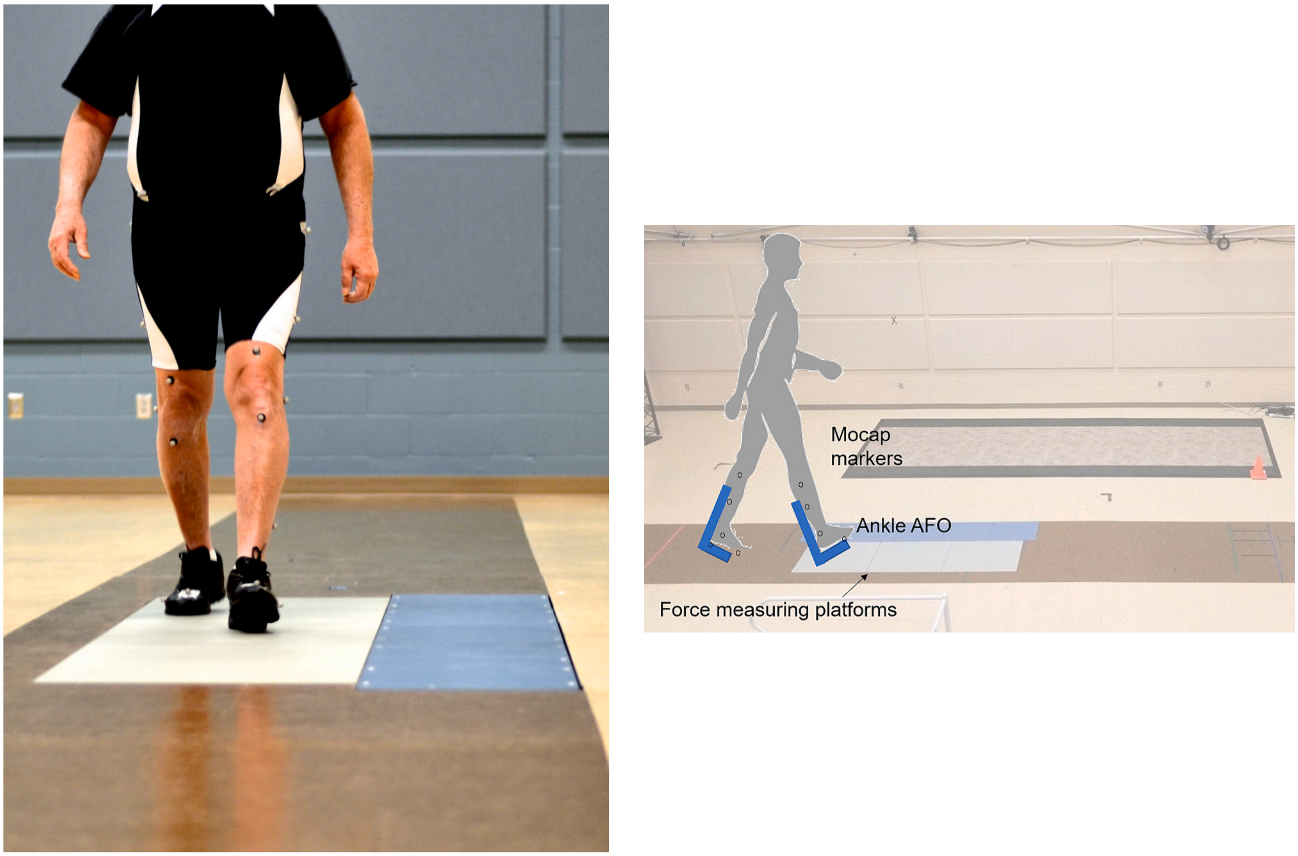

Fig. 1. In Fig. 1a, a patient with PAD is depicted walking on the force platforms with attached reflective markers. In Fig. 1b, a schematic view of the laboratory setup and the AFO utilized in this study is presented.

Supervised exercise and revascularization increase the distances that patients with PAD can walk [16], but these interventions do not lead to significant improvements in spatiotemporal or lower extremity torque and power gait characteristics [8,17]. Many patients encounter obstacles such as lack of motivation, time constraints, limited access to resources, financial constraints, and physical limitations, which impede their ability to adhere effectively to supervised exercise therapy. Existing interventions are designed to improve blood flow or cardiovascular efficiency but are not designed to support the weak calf muscles of patients with PAD. The preliminary data suggesting that an ankle foot orthosis immediately increased patients' walking distance, similar to the effects of six months of medication, holds great promise. Additionally, the AFO appears to address both the deficits in ankle biomechanics identified in previous studies and the blood flow demand issues that contribute to muscular stress in the leg muscles of these patients. This project aims to expand upon the positive results observed in the pilot study and investigate additional improvements in walking performance after three months of using AFOs. AFOs can fill this gap by providing mechanical compensation through energy storage and return. The spring-like properties of carbon-composite AFOs allow energy storage at weight acceptance and return at the toe-off point when the ankle plantarflexes, providing power to propel the body into the next step. AFO devices have stimulated excitement regarding their potential for improving walking performance in individuals with movement disabilities. AFOs have been shown to restore walking performance in individuals with stroke, hemiplegia, and other motor disorders [18,19] that have similar, inadequate lower extremity torque and power generation ability during walking [20,21]. Off-the-shelf carbon composite AFOs are adjustable, affordable, and could be prescribed to overcome reduced propulsion [18,20,22,23].

Limited evidence exists regarding the specific benefits of various assistive devices for patients with PAD, with research often lacking focus on their unique challenges. More targeted studies are needed to assess the effectiveness of different devices, such as passive and powered orthotics, in improving walking distance, reducing pain, and enhancing overall mobility for patients with PAD. Moreover, existing studies primarily offer short-term insights into assistive device usage, lacking longitudinal exploration of their extended benefits and potential complications in patients with PAD. This study aimed to determine how a three-month AFO intervention improved gait performance, as assessed through lower extremity torques and powers in patients with PAD. We also compared gait performance between patients who completed an intervention with bilateral AFOs with a control group of patients with PAD who did not complete an intervention. We hypothesized that the AFO intervention would improve the gait performance of patients with PAD.

2. Methods

This study involves a secondary analysis of a primary dataset. The sample size determination was based on a variable (initial claudication walking distance) that is not present in this study. Forty-three patients with PAD were recruited from the claudication clinic at the Nebraska and Western Iowa Veterans Affairs Medical Center (VAMC) in Omaha, Nebraska. Participants were screened with detailed medical history questions, physical examinations, computerized tomographic angiography, hemodynamic assessment, and direct evaluation and observational analysis of walking impairments.

Patients were included if they had an ankle-brachial index (ABI) of less than 0.90 and claudication pain as a primary limiting factor during walking. Participants also had to demonstrate a positive history of chronic and exercise-limiting claudication, as determined through

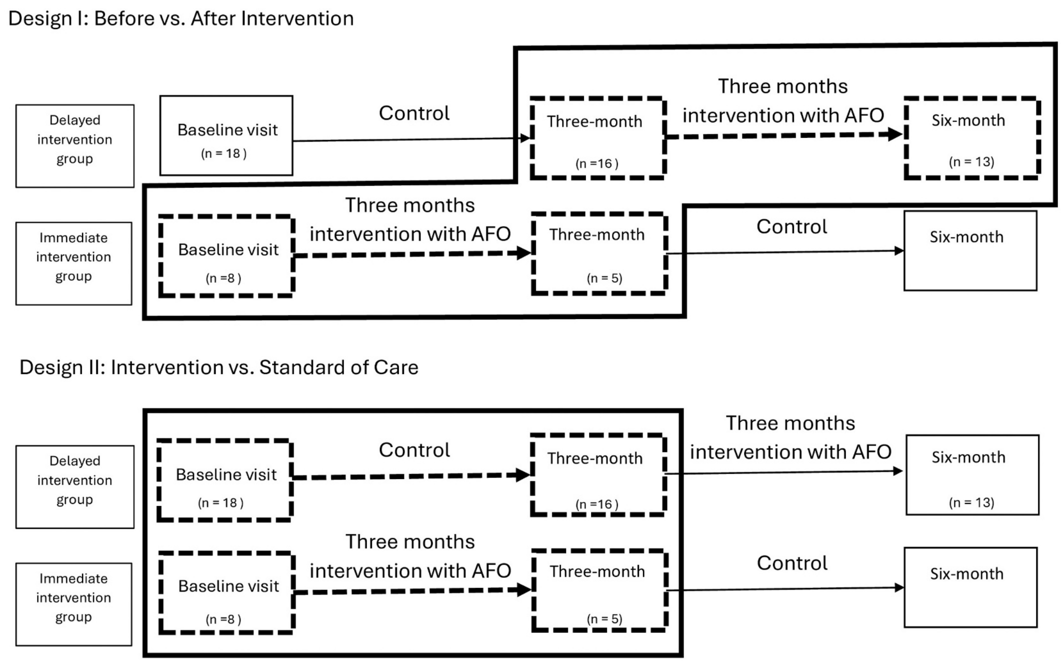

Fig. 2. Study design, group interpretation based on the cross over design of the study. The delayed intervention group had a baseline visit, followed by a three-month control period, then an AFO intervention for three months, ending with a six-month assessment. The immediate intervention group started with a baseline visit, followed by a three-month AFO intervention, then a three-month visit, and concluded with a six-month assessment after a three-month period with no intervention. Design (I) included the comparisons between the dependent variables at pre- vs. post-intervention. This design also included interactions for testing (Ankle foot orthosis (AFO) and no ankle foot orthosis (NAF) and group (delayed versus immediate) factors. This comparison showed changes as a result of the intervention. Design (II) included the comparisons between the dependent variables for at baseline vs. three months. This design also included interactions for testing (Ankle foot orthosis (AFO) and no ankle foot orthosis (NAF) and group (delayed versus immediate) factors. This comparison showed differences as a result of time regardless of control or intervention during that time period.

history and direct observation during a walking test screening. All tests were administered and/or reviewed by the evaluating vascular surgeon.

Participants were excluded if they had rest pain or tissue loss due to PAD, an acute lower extremity ischemic event secondary to thromboembolic disease or acute trauma, and/or walking capacity limited by anything other than claudication. The study was carried out in compliance with the principles stated in the Declaration of Helsinki, and the research protocol received approval from the Ethics Committee of IRBs # 0485–16 and # 1576199. Written informed consent was obtained from each patient before any study-related procedures.

2.1. Data collection

Gait kinematics was measured using a seventeen-high speed-camera motion capture system (Motion Analysis Corp, Santa Rosa, CA; 100 Hz). Participants wore a form-fitting suit, and 33 retro-reflective markers were placed on specific anatomical landmarks on the subject's pelvis, thighs, shanks, and feet according to the marker set protocol [24]. Participants wore bilateral AFOs and were instructed to walk with (AFO) and without AFOs (NAF) at a self-selected speed across a walkway with force plates embedded in the floor (AMTI, Watertown, MA, 1000 Hz). Participants repeated each walking trial until five successful trials were obtained for each foot in which heel-strike and toe-off events were within the boundaries of the force plate. Participants were instructed to walk forward naturally, so they did not intentionally target the force plates. Rest was required for a minimum of one minute between each trial or for as long as was required for the pain to completely subside. Fig. 1 illustrates the experimental setup. In Fig. 1A, a patient with PAD is depicted walking on the force platforms with attached reflective markers. In Fig. 1B, a schematic view of the laboratory setup and the AFO utilized in this study is presented.

2.2. Study groups

Participants were randomly divided into two groups. The first group was instructed to wear AFOs for the first three months, while the second group, referred to as the delayed intervention group, continued usual care without using AFOs during the same three-month period. For the subsequent three months, the participants crossed over to the other group. There were three total visits: baseline, three months, and six months after the initial visit (see Fig. 2).

The intervention included using the AFO at home. During the three-month intervention period, the patients were asked to always wear the AFOs except when they were in bed or showering/bathing. Patients wore either Matrix Truelife or Ottobock Walk-on Reaction AFO models based on the collaborating Orthotist's determination of best fit. These are carbon-composite designs that have been used in previous studies because of the potential to absorb energy in the strut with weight acceptance and return energy at push-off [20]. There was a risk of patients experiencing discomfort with the AFO and developing sores from the device. To reduce this risk, the device was properly fitted and adjusted during the baseline data collection period. Research personnel inspected patients' feet after the data collection to ensure there was no evidence of the device rubbing the foot. Additionally, the nurse coordinator educated patients on how to inspect their feet daily for signs of new sores developing. If subjects began to develop a sore, they were instructed to stop wearing the AFO immediately and to follow up with the research team to adjust the AFO. The experimental protocol was similar across the three visits.

2.3. Data analysis

We filtered marker trajectories using a low pass, fourth-order, zero lag Butterworth filter with cutoff frequencies of 6 Hz [25]. We calculated joint kinetics and kinematics during the stance phase of overground

Table 1
A. Demographic data for the participants used in the pre- and post-intervention analysis (Design I as shown in Fig. 2).

	Group		P-value
	A ($N = 14$)	B ($N = 5$)	
AGE			0.40
Median (IQR)	71.5 (70.0, 77.0)	70.0 (68.0, 72.0)	
BMI			0.43
Median (IQR)	29.7 (26.4, 32.2)	28.6 (22.1, 31.8)	
Height(cm)			0.58
Median (IQR)	175.3 (171.5, 180.3)	177.2 (167.6, 177.8)	
ABI			0.40
Median (IQR)	0.7 (0.5, 0.9)	0.6 (0.5, 0.7)	

B. Demographic data for the participants used at baseline and three-months for the analysis (Design II as shown in Fig. 2).

	Group		P-value
	A ($N = 17$)	B (N = 5)	
AGE			0.22
Median (IQR)	72.0 (70.0, 79.0)	70.0 (68.0, 72.0)	
BMI			0.39
Median (IQR)	30.0 (25.7, 32.8)	28.6 (22.1, 31.8)	
Height(cm)			0.66
Median (IQR)	172.7 (171.5, 180.3)	177.2 (167.6, 177.8)	
ABI			0.42
Median (IQR)	0.7 (0.5, 0.8)	0.6 (0.5, 0.7)	

P-values from Wilcoxon rank sum tests.

walking. We used custom MATLAB (Mathworks, Inc., Natick, Mass) and Visual 3D (C-Motion, Inc., Germantown, Md) software to calculate joint angles, torques, and powers for the ankle, knee, and hip [26]. We analyzed various joint angle variables, including peak flexion and extension angles, during the stance phase of walking. The net result of all forces acting around a joint was defined as joint torque.

We identified the peak values of torque for the ankle, knee, and hip joints. We defined the negative and positive torques as extensor and flexor torques, respectively. In weight acceptance, ankle dorsiflexor, knee flexor, and hip flexor torques were measured, while ankle plantar flexor, knee extensor, and hip extensor torques were observed during propulsion.

Joint power, reflecting muscle work was considered the product of the angular velocity and the net torque at a joint [4]. Peak power values were observed at the ankle, knee, and hip joints. Positive and negative powers were defined as generation and absorption, respectively. During weight acceptance, power absorption occurred at the ankle and knee, and power generation at the hip. During single leg support, power absorption was observed at the ankle and hip, and power generation occurred at the knee. In the propulsion phase, power generation was seen at the ankle and hip, and power absorption occurred at the knee.

2.4. Statistical analysis

Baseline data for the sample were summarized using medians and interquartile ranges (IQRs) and differences in baseline data between groups were assessed using Wilcoxon Rank Sum tests. The statistical analysis of biomechanics outcomes comprised two designs, illustrated in Figs. 2, respectively. The first design involved comparing the delayed intervention and immediate intervention groups before and after the intervention (i.e., three vs. six months for the delayed intervention group, and baseline vs. three months for the immediate intervention group). The second design compared the delayed intervention and immediate intervention groups between baseline and three months, evaluating differences in the intervention versus the control standard of care periods.

For each design analysis, only patients with measurements at both time points were included. Descriptive statistics for outcomes are presented as means and standard deviations based on the raw data. For statistical testing, linear mixed models were used to account for correlated data and separate models were built for each outcome. The models incorporated factors for **time** (before vs. after-intervention in the first design, or baseline vs. three months in the second design), test **condition** (with AFOs vs. without AFOs), and **group** (delayed intervention vs. immediate intervention).

The analysis also included the triple interaction between the three factors, along with all two-way interactions. Backward selection was employed to remove interactions one by one, starting with the highest ordered interaction and then those with the highest p-values, until only interactions with p-values less than 0.05 remained. Main effects were always retained in the model. Significant main effects were interpreted using model-adjusted means and associated standard errors (SEs).

To assess significant three-way interactions, two separate models were generated for each group, investigating whether the two-way interaction between condition and time was significant. Subsequently, significant two-way interactions were examined using post hoc pairwise comparisons with Bonferroni adjustments, and model-adjusted means were presented. All the statistical analyses were performed using SAS software, version 9.4 (SAS Institute Inc., Cary, NC).

3. Results

A description of the demographic characteristics of the patients who participated in the study and completed all the required visits is provided in Table 1A and B. All participants included in this study were male, as recruitment was exclusively conducted at the claudication clinic at the Nebraska and Western Iowa Veterans Affairs Medical Center (VAMC) in Omaha. We performed two separate analysis designs, one comparing the pre and post intervention visits and the other comparing baseline visit to the three-month visit (Figs. 2). The results of the first design and then the second design are described below for each category of outcomes.

3.1. Kinematics outcomes

Pre vs. post intervention (Design I): A significant effect of condition was found for ankle joint angles. After adjusting for group and time, patients had a significantly greater magnitude of peak plantar flexion and dorsiflexion angles in the NAF (−10.55, and 12.42, SE = 0.54 and 0.57) vs. the AFO condition (−8.26, and 9.34, SE = 0.57 and 0.56), p's $<$ 0.01. A significant effect of intervention was found for hip peak joint flexion and extension angles. After three months of intervention with AFOs, both hip flexion and extension angles (36.59, and 0.70, SE = 0.85 and 0.71) were significantly reduced compared to pre intervention (38.19, and 2.40, SE = 0.92 and 0.91), $p = 0.02$ and 0.04, respectively.

Baseline vs. three months (Design II): Patients had a significantly greater magnitude of peak plantarflexion and dorsiflexion angles when walking in the NAF condition (−10.43, and 12.42, SE = 0.56 and 0.49) vs. the AFO condition (−8.37, and 9.45, SE = 0.56 and 0.49), p's < 0.01. Moreover, walking with AFO led to reduced knee extension angle (0.29, SE = 0.71), compared to the NAF condition (1.39, SE = 0.71), $p = 0.04$.

3.2. Kinetics outcomes

Pre vs. post intervention (Design I): We observed reduced knee flexor torque in the NAF (−0.05, SE = 0.04) vs. the AFO condition (−0.12, SE = 0.04), $p = 0.02$. After AFO intervention, the hip flexor torque significantly increased (−0.50, SE = 0.05) compared to pre intervention (−0.41, SE = 0.04), $p = 0.03$. Ankle power absorption during the weight acceptance decreased in the AFO (−0.33, SE = 0.04) vs. NAF condition (−0.43, SE = 0.04), $p = 0.01$. Ankle power absorption during single support decreased in the AFO (−0.78, SE = 0.03), compared to the NAF

Table 2

Dependent variables from design I. Estimates are based on the raw data. *P*-values derived from linear mixed models, where a separate model was run for each measure of interest, and included the variables group, time, and condition, as well as interactions between variables; interactions which were not significant were ultimately removed from the model, starting with the triple interaction. For measures in this table, none of the models included significant interactions. All values are written as mean (standard deviation).

Joint and/or variable	Pre intervention		Pre intervention		Post intervention		Post intervention		P-Values for main effects not included in a significant interaction			Significant Interaction	
	Immediate Start intervention group		Delayed Start intervention group		Immediate Start intervention group		Delayed Start intervention group		Group (Delayed vs. Immediate Start)	Time (Pre vs. Post Intervention)	Condition (AFO vs NAF)	Variables Involved	p-value
Ankle	AFO	NAF	AFO	NAF	AFO	NAF	AFO	NAF					
Plantar flexion angle (Degrees)	−7.87 (2.64)	−9.78 (3.22)	−8.92 (2.35)	−10.29 (2.80)	−8.66 (2.59)	−10.87 (2.47)	−7.77 (2.35)	−10.69 (2.50)	0.95	0.95	0.001		
Dorsiflexion angle (Degrees)	9.95 (0.97)	12.36 (1.53)	9.27 (1.81)	13.30 (3.45)	9.16 (1.80)	11.08 (2.09)	9.72 (3.50)	12.11 (2.68)	0.68	0.29	<0.0001		
Dorsiflexor torque (N*m/kg)	−0.14 (0.04)	−0.16 (0.06)	−0.19 (0.09)	−0.18 (0.06)	−0.19 (0.05)	−0.21 (0.01)	−0.18 (0.06)	−0.20 (0.06)	0.73	0.43	0.37		
Plantar flexor torque (N*m/kg)	1.30 (0.14)	1.33 (0.18)	1.23 (0.12)	1.25 (0.12)	1.26 (0.20)	1.24 (0.21)	1.20 (0.14)	1.24 (0.15)	0.34	0.35	0.34		
Power absorption early stance (W/kg)	−0.23 (0.05)	−0.47 (0.18)	−0.36 (0.14)	−0.41 (0.19)	−0.36 (0.13)	−0.43 (0.15)	−0.32 (0.12)	−0.44 (0.23)	0.84	0.86	0.01		
Power absorption mid stance (W/kg)	−0.86 (0.10)	−1.14 (0.23)	−0.73 (0.11)	−1.02 (0.24)	−0.81 (0.17)	−0.94 (0.16)	−0.72 (0.10)	−0.97 (0.14)	0.34	0.09	<0.0001		
Power generation late stance (W/kg)	1.33 (0.26)	2.03 (0.40)	1.45 (0.35)	1.98 (0.51)	1.55 (0.29)	1.89 (0.51)	1.50 (0.35)	2.13 (0.56)	0.72	0.36	<0.0001		
Knee													
Extension angle (Degrees)	0.59 (3.98)	1.32 (2.59)	0.90 (3.76)	2.28 (3.84)	−0.43 (2.69)	0.45 (2.81)	0.04 (4.95)	1.43 (3.37)	0.65	0.23	0.06		
Flexion angle (Degrees)	11.61 (3.65)	12.62 (3.51)	10.81 (5.46)	10.80 (4.84)	10.60 (4.64)	10.52 (3.35)	9.86 (5.47)	10.69 (6.19)	0.63	0.28	0.65		
Extensor torque (N*m/kg)	0.34 (0.03)	0.44 (0.06)	0.55 (0.20)	0.51 (0.27)	0.49 (0.06)	0.52 (0.15)	0.57 (0.24)	0.59 (0.21)	0.24	0.13	0.62		
Flexor torque(N*m/kg)	−0.29 (0.19)	−0.16 (0.17)	−0.05 (0.16)	5.37 (0.17)	−0.12 (0.24)	−0.59 (0.11)	−0.007 (0.19)	0.05 (0.16)	0.03	0.08	0.02		
Power absorption early stance (W/kg)	−0.33 (0.16)	−0.060 (0.19)	−0.62 (0.41)	−0.57 (0.47)	−0.51 (0.16)	−0.51 (0.26)	−0.64 (0.35)	−0.67 (0.41)	0.37	0.41	0.61		
Power generation early stance (W/kg)	0.15 (0.33)	0.26 (0.07)	0.35 (0.21)	0.34 (0.24)	0.27 (0.14)	0.31 (0.22)	0.34 (0.23)	0.38 (0.22)	0.21	0.43	0.31		
Power absorption late stance (W/kg)	−0.64 (0.65)	−0.68 (0.36)	−0.79 (0.39)	−0.88 (0.54)	−0.80 (0.59)	−0.85 (0.25)	−0.82 (0.24)	−0.95 (0.33)	0.58	0.35	0.08		
Hip													
Extension angle (Degrees)	4.64 (2.41)	3.04 (3.22)	1.53 (4.75)	1.51 (3.95)	0.61 (1.62)	1.41 (1.87)	0.17 (3.65)	0.28 (3.65)	0.37	0.02	0.29		
Flexion angle (Degrees)	39.04 (2.67)	39.90 (2.75)	37.24 (5.07)	37.22 (4.44)	36.73 (1.91)	37.57 (2.89)	35.22 (4.84)	36.55 (4.55)	0.42	0.04	0.86		
Extensor torque (N*m/kg)	0.74 (0.23)	0.75 (0.21)	0.69 (0.18)	0.71 (0.20)	0.65 (0.09)	0.59 (0.05)	0.65 (0.16)	−0.56 (0.21)	0.87	0.15	0.85		
Flexor torque (N*m/kg)	−0.33 (0.16)	−0.40 (0.13)	−0.43 (0.18)	5.37 (0.17)	−0.45 (0.20)	−0.49 (0.18)	−0.50 (0.18)	0.64 (0.16)	0.36	0.03	0.11		
Power absorption mid stance (W/kg)	−0.18 (0.10)	−0.17 (0.16)	−0.33 (0.24)	−0.29 (0.23)	−0.29 (0.21)	−0.27 (0.17)	−0.40 (0.27)	−0.38 (0.28)	0.18	0.21	0.22		
Power generation early stance (W/kg)	0.74 (0.37)	0.75 (0.31)	0.65 (0.26)	0.64 (0.28)	0.064 (0.23)	0.53 (0.11)	0.54 (0.21)	0.50 (0.28)	0.42	0.08	0.42		
Power generation late stance (W/kg)	0.55 (0.23)	0.65 (0.17)	0.70 (0.25)	0.78 (0.31)	0.67 (0.11)	0.71 (0.13)	0.72 (0.21)	0.87 (0.36)	0.27	0.16	0.02		

condition (−1.01, SE = 0.03), $p < 0.0001$. Ankle power generation in late stance was also reduced in the AFO condition (1.46, SE = 0.09) compared to the NAF condition (2.02, SE = 0.09), $p < 0.0001$. Hip power generation in late stance was reduced in the AFO condition (0.66, SE = 0.06) compared with the NAF condition (0.76, SE = 0.06), $p = 0.02$ (Table 2).

Baseline vs. three months (Design II): We observed reduced knee flexor torque in the NAF condition (−0.06, SE = 0.04) compared to the AFO condition (−0.12, SE = 0.04), $p = 0.03$. We observed a two-way interaction between time and condition for ankle power absorption (p = 0.02; Table 3). Specifically, when walking with AFOs, ankle power absorption was significantly greater (−0.29, SE = 0.04) than in the NAF condition (−0.43, SE = 0.04) at baseline ($p = 0.002$), whereas there was no statistically significant difference in ankle power absorption between the NAF and AFO conditions at 3 months ($p = 0.30$). We observed a three-way interaction for ankle power generation ($p = 0.01$; Table 3). We ran two separate models, one for each group, investigating the interaction between time and condition. The no intervention (i.e. delayed) group had a significant main effect of condition, where ankle power generation was reduced when walking with AFOs (1.40, SE = 0.10) compared to the NAF condition (1.87, SE = 0.10), $p < 0.001$. However, for the intervention (i.e. immediate) group, there were no significant effects of time or condition ($p > 0.05$). We observed a two-way interaction between time and condition for knee power absorption in early stance ($p = 0.02$), however, post hoc comparisons were not significant.

4. Discussion

The current study evaluated the effect of a three-month intervention with AFOs, used during daily activities, on gait biomechanics in patients with PAD. We also conducted a comparison of gait performance among patients who underwent an AFO intervention and those who did not, during their initial assessment and three-month follow-up visits. The utilization of AFOs, as opposed to not wearing them, resulted in a decreased ankle plantar flexion angle and diminished ankle power generation and absorption. This, in turn, would alleviate stress on the ankle muscles during plantar flexion. Patients with PAD showed improved ankle kinematics, as evidenced by a decrease in excessive ankle motion; however, the results for other lower limb joint kinetics were mixed.

Prior literature has shown that patients with PAD demonstrate a lack of ability to control downward foot movement after a heel strike, which leads to the foot dropping quickly [27]. This disrupts the roll-over shape that leverage's the body's ability to roll forward during the single-limb support phase of walking, which in turn decreases the mechanical efficiency of walking [28]. Using AFOs was effective in decreasing this excessive and rapid ankle plantar flexion following heel strike. This finding aligns with previous research, emphasizing the positive impact of AFOs [29]. In our prior research, we demonstrated that patients with PAD displayed significantly increased ankle plantar flexion in early stance and a more extensive range of motion in the ankle during stance compared to other cohorts ($p < 0.05$) [27]. Reducing the range of motion in ankle movement could potentially align the ankle motion more closely with the patterns observed in healthy controls. Excessive ankle plantar flexion is often attributed to ankle dorsiflexor muscles or the plantar flexor muscles contracting. These factors contribute to an inability to control foot lowering after a heel strike [15]. In our study, the AFO seemed to support normal foot lowering in these patients.

Using AFOs resulted in decreased joint power generation in the ankle during the late stance phase. The reported restriction of ankle range of motion when walking with AFOs has been linked to a diminished ability to generate ankle power [20,30]. Ankle plantar flexors generate as much as 50% of the total mechanical power during late stance [31]. Therefore, it is important that any assistive device contributes to late stance kinetics if a goal is improving the efficiency of walking. The spring-like properties of carbon-composite AFOs allowed energy storage at weight acceptance and return at the point of toe-off when the ankle plantar flexors are supposed to propel into the next step. The reduced ankle power absorption observed during single support and weight acceptance phases when using AFO can be attributed to decreased energy delivery by the plantar flexor muscles. This decrease in energy generation during stance limits the energy available for absorption and redistribution, leading to diminished ankle power absorption. Thus, employing AFOs might offer compensation for weakened muscles, albeit at the expense of reduced power generation during push-off [32–34].

At the knee, walking with AFOs reduced the peak knee extension angle. During the stance phase of walking, the knee typically undergoes extension (straightening) during single-limb support as weight is transferred from one leg to the other. Although AFOs are designed to provide support and control to the ankle and foot, they can limit ankle movement, leading to altered movement at other joints [35]. With limited ankle movement, the knee did not fully extend during the stance phase of walking. Specifically, the limited ankle movement tends to keep the foot in a neutral position and keeps the foot from achieving typical dorsiflexion, which led to an earlier toe-off, prior to the knee straightening [36]. Decreased dorsiflexion, leading to earlier toe-off, may yield varied outcomes depending on individual factors and walking mechanics. While it can assist in foot clearance for individuals with foot drop and enhance energy efficiency, it may also prompt compensatory movements and potentially adverse impacts on the muscles and joints involved. The constrained knee range of motion linked with AFOs could result in adverse long-term consequences, such as the risk of knee extensor muscle atrophy due to disuse. Nevertheless, AFOs that permit a degree of knee movement may improve the muscle activity balance of the vastus medialis [37]. Additionally, extended AFO usage may impact joint stiffness and tendon shortening [38]. The muscle activity levels in users are intricately influenced by compensatory walking patterns and the positioning of ground reaction forces relative to lower limb joints, all of which are contingent upon the characteristics of the AFO.

The knee not extending fully during single-limb support would lead to increased knee flexor torque, as individuals seek to transfer weight to the next step. This is consistent with Kobayashi et al. (2016) [39], who observed that restricting ankle plantar flexion using steel springs led to an increase in knee flexor torque. This increased knee flexor torque is what we observed when walking with AFOs. Individuals may rely more on knee flexion as a compensatory strategy to generate forward propulsion or to compensate for limited ankle movement allowed by the AFOs. Moreover, AFO often possess a certain degree of stiffness to provide support and stability. The AFO stiffness can affect the biomechanics of the lower limb and influence muscle activation patterns. Increased knee flexor torque may be necessary to compensate for the resistance provided by the AFO's stiffness. When the AFOs are excessively rigid, it leads to an increased knee flexor torque during the weight acceptance phase, which can contribute to instability during walking [40].

Hip range of motion reduced after the AFO intervention. The reduction in excessive ankle plantar flexion likely led to the changes in the hip angle during walking with AFOs after the intervention period. Patients with PAD were able to preserve the natural foot rocker shape when walking with AFOs, facilitating efficient mechanical propulsion into the swing phase of the opposite leg. As a result, the hip's involvement in aiding limb swing was reduced, and hip range of motion also decreased [41].

In previous studies, there were no reported effects on hip kinematics and kinetics when using various types of AFOs [36,42]. In this study, hip flexor torque increased after the three-month intervention. The contrasting results could be attributed to previous studies using different patient populations than PAD, or because we investigated the long-term effect of AFOs. Over time, individuals wearing AFOs may undergo functional adaptations to the device. This adaptation process involves neuromuscular changes and motor learning to accommodate wearing AFOs. As the hip flexors play a crucial role in maintaining stability and

Table 3
Dependent variables from design II. Estimates are based on the raw data. *P*-values derived from linear mixed models, where a separate model was run for each measure of interest, and included the variables group, time, and condition, as well as interactions between variables; interactions which were not significant were ultimately removed from the model, starting with the triple interaction. Detailed interpretations of interactions can be found in the results section.

Joint and/or variable	Baseline		Baseline		Three months		Three months		P-Values for main effects not included in a significant interaction			Significant Interaction	
	Immediate intervention group		Delayed intervention group (i.e. No intervention)		Immediate intervention group		Delayed intervention group (i.e. No intervention)		Intervention Group (Intervention vs. No Intervention)	Time (Baseline vs. Three Months)	Condition (AFO vs NAF)	Variables Involved	p-value
Ankle	AFO	NAF	AFO	NAF	AFO	NAF	AFO	NAF					
Plantar flexion angle (Degree)	−7.87 (2.64)	−9.78 (3.22)	−7.71 (2.74)	−10.97 (3.19)	−8.66 (2.59)	−10.87 (2.47)	−8.70 (2.32)	−10.24 (2.77)	1.00	0.45	0.003		
Dorsiflexion angle (Degree)	9.95 (0.97)	12.36 (1.53)	9.65 (2.44)	12.46 (2.69)	9.16 (1.80)	11.00 (2.09)	9.45 (1.79)	13.18 (3.14)	0.71	0.82	<0.0001		
Dorsiflexor torque (N*m/kg)	−0.14 (0.04)	−0.16 (0.06)	−0.19 (0.08)	−0.18 (0.07)	−0.19 (0.05)	−0.21 (0.01)	−0.18 (0.08)	−0.17 (0.06)	0.95	0.80	0.96		
Plantar flexor torque (N*m/kg)	1.30 (0.14)	1.33 (0.18)	1.19 (0.17)	1.21 (0.16)	1.26 (0.20)	1.24 (0.21)	1.21 (0.14)	1.23 (0.13)	0.34	0.98	0.44		
Power absorption early stance (W/kg)	−0.23 (0.05)	−0.47 (0.18)	−0.29 (0.18)	−0.40 (0.19)	−0.36 (0.13)	−0.43 (0.15)	−0.33 (0.15)	−0.37 (0.19)	0.69	–	–	Time x Condition	0.02
Power absorption mid stance (W/kg)	−0.86 (0.10)	−1.14 (0.23)	−0.70 (0.12)	−0.97 (0.17)	−0.81 (0.17)	−0.94 (0.16)	−0.74 (0.09)	−0.01 (0.22)	–	–	<0.0001	Group x Time	0.04
Power generation late stance (W/kg)	1.33 (0.26)	2.03 (0.40)	1.39 (0.46)	1.79 (0.60)	1.55 (0.29)	1.89 (0.51)	1.43 (0.35)	1.92 (0.49)	–	–	–	Group x Time x Condition	0.01
Knee													
Extension angle (Degree)	0.59 (3.98)	1.32 (2.59)	0.82 (3.77)	1.63 (4.77)	−0.43 (2.69)	0.45 (2.81)	0.40 (3.31)	2.06 (3.57)	0.44	0.91	0.049		
Flexion angle (Degree)	11.61 (3.65)	12.62 (3.51)	9.78 (6.23)	9.84 (6.30)	10.60 (4.68)	10.52 (3.35)	10.19 (5.42)	10.44 (4.55)	0.62	0.98	0.76		
Extension torque (N*m/kg)	0.34 (0.03)	0.44 (0.06)	0.51 (0.20)	0.54 (0.26)	0.49 (0.06)	0.52 (0.15)	0.57 (0.17)	0.51 (0.25)	0.44	0.47	0.96		
Flexor torque (N*m/kg)	−0.29 (0.19)	−0.16 (0.17)	−0.04 (0.16)	0.01 (0.13)	−0.12 (0.24)	−0.05 (0.11)	−0.01 (0.17)	0.008 (0.16)	0.03	0.29	0.03		
Power absorption early stance (W/kg)	0.15 (0.03)	−0.60 (0.19)	−0.52 (0.34)	−0.58 (0.47)	−0.51 (0.16)	−0.51 (0.26)	−6.63 (0.37)	−0.55 (0.43)	0.65	–	–	Time x Condition	0.02
Power generation early stance (W/kg)	0.74 (0.37)	0.26 (0.07)	0.30 (0.20)	0.33 (0.23)	0.27 (0.14)	0.31 (0.22)	0.35 (0.19)	0.32 (0.21)	0.38	0.50	0.40		
Power absorption late stance (W/kg)	−0.64 (0.65)	−0.68 (0.36)	−0.76 (0.25)	−0.82 (0.23)	−0.80 (0.59)	−0.85 (0.25)	−0.83 (0.36)	−0.88 (0.48)	0.50	0.32	0.28		
Hip													
Extension angle (Degree)	4.64 (3.41)	3.04 (3.22)	0.69 (4.18)	1.83 (3.15)	0.61 (1.62)	1.41 (1.87)	1.31 (5,14)	1.57 (4.02)	0.08	0.98	0.16		
Flexion angle (Degree)	19.04 (2.67)	39.90 (2.75)	34.31 (5.53)	35.82 (3.21)	36.73 (1.91)	37.57 (2.89)	35. 82 (6.17)	35.95 (5.04)	0.32	0.48	0.44		
Extension torque (N*m/kg)	0.74 (0.23)	0.75 (0.21)	0.58 (0.20)	0.62 (0.22)	0.65 (0.09)	0.59 (0.05)	0.69 (0.16)	0.67 (0.20)	0.86	0.60	0.93		
Flexor torque (N*m/kg)	−0.33 (0.16)	−0.40 (0.13)	−0.50 (0.16)	−0.49 (0.11)	−0.45 (0.20)	−0.49 (0.18)	−0.44 (0.19)	−0.45 (0.17)	0.47	0.73	0.51		
Power absorption mid stance (W/kg)	−0.18 (0.10)	−0.17 (0.16)	−0.42 (0.22)	−0.38 (0.14)	−0.29 (0.21)	−0.27 (0.17)	−0.34 (0.25)	−0.30 (0.25)	0.08	0.55	0.10		
Power generation early stance (W/kg)	0.74 (0.37)	0.75 (0.31)	0.51 (0.32)	0.51 (0.22)	0.64 (0.23)	0.53 (0.11)	0.61 (0.27)	0.58 (0.29)	0.52	0.79	0.47		
Power generation late stance (W/kg)	0.55 (0.23)	0.65 (0.17)	0.68 (0.21)	0.71 (0.14)	0.67 (0.11)	0.71 (0.13)	0.69 (0.24)	0.74 (0.29)	0.48	0.51	0.06		

generating propulsion during walking, hip flexor activation and torque may have increased to compensate for the altered mechanics introduced by the AFOs. These findings align with previous research that suggests walking with AFOs can impede the smooth motion of plantarflexion before toe off [40] phase. As a result, patients may need to adjust rapidly hip flexion, leading to increased hip angular velocity while pulling the trailing leg forward. This compensatory action aims to offset the limited ankle plantarflexion during before toe off [43].

Using AFOs seemed to be beneficial overall in supporting patients with PAD, who have muscles that are weakened from myopathy due to chronic cycles of ischemia and reperfusion [13,15,44–46]. However, if the AFO is too stiff, it may hinder ankle function during stance, resulting in a restricted dorsiflexion, and a reduced ability to perform work around the ankle joint. The kinematics analysis revealed that wearing AFOs reduced ankle dorsiflexion during walking due to the device's limited ankle articulation. This restriction helped patients with PAD to use their foot-ankle complex more effectively as a rocker during gait, but it also constrained overall ankle motion. While AFOs can provide stability and improve gait efficiency, AFO design and prescription should be carefully considered in relation to specific functional needs and activities. This is discussed in more details in the limitations of this study [47].

Limitations of the study include baseline differences in biomechanics values between groups, a limited number of female participants, and unequal group sizes. Furthermore, healthy individuals were not included in this study. Further research is necessary to verify the influence of different AFO materials and stiffnesses on walking performance. Moreover, a significant number of participants did not return for follow-up visits, and research indicates poor intervention adoption rates are due to weaknesses in AFO design and function [48,49]. Participants who did not adopt the AFO tended to have negative perceptions early in the intervention. It is plausible that individuals who chose to withdraw might have more medical complexities and could potentially derive greater benefits from the intervention. Therefore, developing devices more compatible with completing daily activities warrants further research. Little is known about the mechanical contribution of AFOs with different levels of stiffness, which is another potential area of investigation.

Conclusion: In conclusion, the use of AFOs shows potential positive effects on gait kinematics and kinetics, with some variables approaching levels observed in healthy individuals. It is important to note that the effects may vary and cannot be universally categorized as improvement. These initial findings generate enthusiasm for further research on assistive devices aimed at improving the gait of patients with PAD.

CRediT authorship contribution statement

Farahnaz Fallahtafti: Writing – review & editing, Writing – original draft, Software, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Kaeli Samson:** Writing – review & editing, Visualization, Software, Methodology, Formal analysis. **Zahra Salamifar:** Software, Methodology, Formal analysis, Conceptualization. **Jason Johanning:** Methodology, Investigation, Funding acquisition, Conceptualization. **Iraklis Pipinos:** Methodology, Investigation, Conceptualization. **Sara A. Myers:** Writing – review & editing, Methodology, Investigation, Funding acquisition, Formal analysis, Conceptualization.

Declaration of competing interest

Some of the the ankle foot orthosis used by subjects in the study were donated by Ottobock.

Acknowledgments

The authors would like to thank the participants, and Mahdi Hassan and Hafizur Rahman for their help with data collection. This study was Supported by the National Institutes of Health (R01HD090333), Veteran Affairs (Merit grant no. I01 RX003266), United States Department of Veterans Affairs Rehabilitation Research and Development Service (va.gov). The authors declare no conflict of interest.

References

[1] M.H. Criqui, et al., Lower extremity peripheral artery disease: contemporary epidemiology, management gaps, and future directions: a scientific statement from the American Heart Association, Circulation 144 (9) (2021) e171–e191.
[2] A.T. Hirsch, et al., Peripheral arterial disease detection, awareness, and treatment in primary care, Jama 286 (11) (2001) 1317–1324.
[3] K. Matsushita, et al., Lifetime risk of lower-extremity peripheral artery disease defined by ankle-brachial index in the United States, J. Am. Heart Assoc. 8 (18) (2019) e012177.
[4] P. Koutakis, et al., Joint torques and powers are reduced during ambulation for both limbs in patients with unilateral claudication, J. Vasc. Surg. 51 (1) (2010) 80–88.
[5] J.D. McCamley, et al., Gait mechanics differences between healthy controls and patients with peripheral artery disease after adjusting for Gait velocity stride length and step width, J. Appl. Biomech. 35 (1) (2019), 19–24–19–24.
[6] P. Koutakis, et al., Abnormal joint powers before and after the onset of claudication symptoms, J. Vasc. Surg. 52 (2) (2010) 340–347.
[7] M. Szymczak, et al., Gait pattern in patients with peripheral artery disease, BMC Geriatr. 18 (1) (2018) 52.
[8] S. Myers, et al., Spatiotemporal changes posttreatment in peripheral arterial disease, Rehabil. Res. Pract. 2015 (2015) 1–10.
[9] I.I. Pipinos, et al., Phosphorus 31 nuclear magnetic resonance spectroscopy suggests a mitochondrial defect in claudicating skeletal muscle, J. Vasc. Surg. 31 (5) (2000) 944–952.
[10] M. AlGhatrif, et al., Lower mitochondrial energy production of the thigh muscles in patients with low-normal ankle-brachial index, J. Am. Heart Assoc. 6 (9) (2017).
[11] M.M. McDermott, et al., Lower extremity ischemia, calf skeletal muscle characteristics, and functional impairment in peripheral arterial disease, J. Am. Geriatr. Soc. 55 (3) (2007) 400–406.
[12] M.M. McDermott, et al., Calf muscle characteristics, strength measures, and mortality in peripheral arterial disease: a longitudinal study, J. Am. Coll. Cardiol. 59 (13) (2012) 1159–1167.
[13] M.N. Schieber, et al., Muscle strength and control characteristics are altered by peripheral artery disease, J. Vasc. Surg. 66 (1) (2017) 178–186.e12.
[14] S.R. Wurdeman, et al., Patients with peripheral arterial disease exhibit reduced joint powers compared to velocity-matched controls, Gait Posture 36 (3) (2012) 506–509.
[15] S.J. Chen, et al., Bilateral claudication results in alterations in the gait biomechanics at the hip and ankle joints, J. Biomech. 41 (11) (2008) 2506–2514.
[16] J. Regensteiner, W. Hiatt, Exercise in the management of peripheral arterial disease ACSM's Resource Manual for Guidelines for Exercise Testing and Prescription, Lippincott Willaims and Wilkins, Philadelphia, PA, 2001.
[17] S.A. Myers, et al., Spatiotemporal changes posttreatment in peripheral arterial disease, Rehabil. Res. Pract. 2015 (2015) 124023.
[18] A. Danielsson, K.S. Sunnerhagen, Energy expenditure in stroke subjects walking with a carbon composite ankle foot orthosis, J. Rehabil. Med. 36 (4) (2004) 165–168.
[19] S.I. Wolf, et al., Dynamic assist by carbon fiber spring AFOs for patients with myelomeningocele, Gait Posture 28 (1) (2008) 175–177.
[20] D.J. Bregman, et al., Spring-like Ankle Foot Orthoses reduce the energy cost of walking by taking over ankle work, Gait Posture 35 (1) (2012) 148–153.
[21] E.S. Arch, S.J. Stanhope, Passive-dynamic ankle-foot orthoses substitute for ankle strength while causing adaptive gait strategies: a feasibility study, Ann. Biomed. Eng. 43 (2) (2015) 442–450.
[22] K. Desloovere, et al., How can push-off be preserved during use of an ankle foot orthosis in children with hemiplegia? A prospective controlled study, Gait Posture 24 (2) (2006) 142–151.
[23] A. Bartonek, M. Eriksson, E.M. Gutierrez-Farewik, Effects of carbon fibre spring orthoses on gait in ambulatory children with motor disorders and plantarflexor weakness, Dev. Med. Child Neurol. 49 (8) (2007) 615–620.
[24] B.M. Nigg, G.K. Cole, W. Nachbauer, Effects of arch height of the foot on angular motion of the lower extremities in running, J. Biomech. 26 (8) (1993) 909–916.
[25] K.M. Jackson, Fitting of mathematical functions to biomechanical data, IEEE Trans. Biomed. Eng. 26 (2) (1979) 122–124.
[26] D.A. Winter, The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological, University of Waterloo Press, 1991.
[27] R. Celis, et al., Peripheral arterial disease affects kinematics during walking, J. Vasc. Surg. 49 (1) (2009) 127–132.
[28] G.M. Bapat, S.A. Myers, A robust technique for optimal fitting of roll-over shapes of human locomotor systems, Med. Eng. Phys. 100 (2022) 103756.
[29] R.L. Mizner, A.A. Mays, R.J. Mays, Mechanical adaptations in walking using carbon fiber ankle foot orthoses for patients with peripheral artery disease, Gait Posture 101 (2023) 14–20.
[30] E. Russell Esposito, et al., How does ankle-foot orthosis stiffness affect gait in patients with lower limb salvage? Clin. Orthop. Relat. Res. 472 (10) (2014) 3026–3035.
[31] S.N. Fickey, M.G. Browne, J.R. Franz, Biomechanical effects of augmented ankle power output during human walking, J. Exp. Biol. 221 (22) (2018).

[32] R. Woo, Spasticity: orthopedic perspective, J. Child Neurol. 16 (1) (2001) 47–53.
[33] C.E. Buckon, et al., Comparison of three ankle-foot orthosis configurations for children with spastic hemiplegia, Dev. Med. Child Neurol. 43 (6) (2001) 371–378.
[34] S. Ounpuu, et al., An evaluation of the posterior leaf spring orthosis using joint kinematics and kinetics, J. Pediatr. Orthop. 16 (3) (1996) 378–384.
[35] M. Nevisipour, C.F. Honeycutt, The impact of ankle-foot-orthosis (AFO) use on the compensatory stepping response required to avoid a fall during trip-like perturbations in young adults: implications for AFO prescription and design, J. Biomech. 103 (2020) 109703.
[36] A. Daryabor, M. Arazpour, G. Aminian, Effect of different designs of ankle-foot orthoses on gait in patients with stroke: A systematic review, Gait Posture 62 (2018) 268–279.
[37] M. Murayama, Knee joint movement and muscle activity changes in stroke hemiplegic patients on continuous use of knee-ankle-foot orthosis with adjustable knee joint, J. Phys. Ther. Sci. 33 (4) (2021) 322–328.
[38] K. Ghoseiri, A. Zucker-Levin, Long-term locked knee ankle foot orthosis use: a perspective overview of iatrogenic biomechanical and physiological perils, Front Rehabil Sci 4 (2023) 1138792.
[39] T. Kobayashi, et al., Reduction of genu recurvatum through adjustment of plantarflexion resistance of an articulated ankle-foot orthosis in individuals post-stroke, Clin. Biomech. 35 (2016) 81–85.
[40] M. Alam, I.A. Choudhury, A. Bin Mamat, Mechanism and design analysis of articulated ankle foot orthoses for drop-foot, ScientificWorldJournal **2014** (2014) 867869.
[41] E. Genda, et al., A new walking orthosis for paraplegics: hip and ankle linkage system, Prosthetics Orthot. Int. 28 (1) (2004) 69–74.
[42] F. Farmani, et al., The influence of rocker bar ankle foot orthosis on gait in patients with chronic hemiplegia, J. Stroke Cerebrovasc. Dis. 25 (8) (2016) 2078–2082.
[43] A. Daryabor, et al., Therapeutic effect of gait training with two types of ankle-foot orthoses on the gait of the stroke patients in the recovery phase, Turk J Phys Med Rehabil 68 (2) (2022) 175–183.
[44] J.M. Perkins, et al., Exercise training versus angioplasty for stable claudication. Long and medium term results of a prospective, randomised trial, Eur. J. Vasc. Endovasc. Surg. 11 (4) (1996) 409–413.
[45] P. Koutakis, et al., Abnormal accumulation of desmin in gastrocnemius myofibers of patients with peripheral artery disease: associations with altered myofiber morphology and density, mitochondrial dysfunction and impaired limb function, J. Histochem. Cytochem. 63 (4) (2015) 256–269.
[46] P. Koutakis, et al., Abnormal myofiber morphology and limb dysfunction in claudication, J. Surg. Res. 196 (1) (2015) 172–179.
[47] D.J. Bregman, et al., Polypropylene ankle foot orthoses to overcome drop-foot gait in central neurological patients: a mechanical and functional evaluation, Prosthetics Orthot. Int. 34 (3) (2010) 293–304.
[48] A.Z. Bashir, et al., Long-term use of an ankle-foot orthosis intervention in patients with peripheral artery disease using the integrated promoting action on research implementation in health services (i-PARIHS) framework, Int. J. Cardiol. 372 (2023) 23–32.
[49] A.Z. Bashir, et al., Considerations for implementation of an ankle-foot orthosis to improve mobility in peripheral artery disease, Arch Rehabil Res Clin Transl 3 (1) (2021) 100092.