
Variations of ankle-foot orthosis-constrained movements increase ankle range of movement while maintaining power output of recumbent cycling

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Abstract: Previous research investigated recumbent cycle power output (PO) from the perspective of knee and hip joint biomechanics. However, ankle-foot biomechanics and, in particular, the effect of ankle-foot orthosis (AFO)-constrained movements on cycle PO has not been widely explored. Therefore, the purpose of this study was to determine whether AFOs of a fixed position (FP) and in dorsi-plantarflexion (DPF)-, dorsiflexion (DF)- and plantarflexion (PF)-constrained movements might influence PO during voluntary recumbent cycling exercises. Twenty-five healthy individuals participated in this study. All underwent 1-min cycling at a fixed cadence for each of the AFOs. The peak and average PO of each condition were analyzed. The peak and average PO were 27.2 ± 12.0 W (range 6–60) and 17.2 ± 9.0 W (range 2–36), respectively, during voluntary cycling. There were no significant differences in the peak PO generated by the AFOs (p = 0.083). There were also no significant differences in the average PO generated using different AFOs (p = 0.063). There were no significant differences in the changes of the hip and knee joint angles with different AFOs (p = 0.974 and p = 1.00, respectively). However, there was a significant difference in the changes of the ankle joint angle (p < 0.00). The present study observed that AFO-constrained movements did not have an influence in altering PO during voluntary recumbent cycling in healthy individuals. This finding might serve as a reference for future rehabilitative cycling protocols.

Keywords: ankle-foot orthosis; ankle movement; pedal power output; rehabilitation exercise; voluntary recumbent cycling.

Introduction

Recumbent cycling is a popular exercise modality for both healthy individuals and patients with musculoskeletal injury or neurological disability. Persons with musculoskeletal injury or neurological disability may require gait training, involving the use of a foot orthosis [15, 20] and an ankle-foot orthosis (AFO) [1, 2, 16, 17]. The prescribed AFOs may be of several types, such as solid AFO, hinged AFO, floor-reaction AFO, posterior leaf spring AFO, anterior elastic AFO, dorsi-assist/dorsi-stop AFO (DA-DS) and plantar stop/free dorsi-flexion AFO (PS/DF AFO) [1, 2, 16, 17], which aim to promote recovery or improve functional outcomes. However, weight bearing may be an issue, thus requiring them to train using cycling ergometers. The general goal of recumbent cycling exercise rehabilitation is to produce the highest possible mechanical power to maximize the merit of health benefits while using the AFOs [19, 23, 24]. A fixed AFO or a fixed pedal boot is often deployed to affix the foot to the pedal, and this has been widely used to also provide shank stability, thus restricting leg movements in the sagittal plane during recumbent cycling [3, 12, 26, 28]. During electrically stimulated cycling, for example, in individuals with neurological impairments, the ankle is uncontrolled by voluntary muscle contractions due to a lack of muscle strength or ankle instability, and constraining the ankle range of motion (ROM) might produce different power output (PO) outcomes. Only a few studies have been reported focusing on the recumbent bicycle [11] and the effect of orthoses provided during cycling [31]. Therefore, comparative data from able-bodied (AB) individuals performing voluntary recumbent cycling with AFO-constrained ankle movements are needed to fairly infer its benefits for the musculoskeletal or neurologically impaired population.
to more efficiently use the recumbent bicycle as a rehabilitation tool.

Researchers have previously sought to elicit maximum PO during recumbent cycling in order to increase the benefits of cycling during rehabilitation. Szecsi et al. [25], Sinclair et al. [22], Berkelmans [4] and Duffell et al. [6, 7] have stated that the mechanical PO produced during electrically stimulated cycling in patients with spinal cord injury (SCI) is very low compared to the PO produced during voluntary cycling in AB individuals. On the other hand, Gregor et al. [11] reported that the power requirement for recumbent cycling is less than that for upright cycling for a given overground velocity. Koch et al. [13] reported that there was no significant difference in power production a given overground velocity. Consequently, several studies investigated the origins of PO during recumbent cycling in AB individuals. On the other hand, Gregor et al. [11] reported that the power requirement for recumbent cycling is less than that for upright cycling for a given overground velocity. Koch et al. [13] reported that there was no significant difference in power production during cycling with and without the use of a foot orthosis. Consequently, several studies investigated the origins of cycling PO during electrically stimulated cycling and voluntary exercise as a function of knee and hip joint biomechanics, muscle size and strength, crank angle, cadence, workload or resistance, foot positioning, positioning of the participants on the recumbent bicycle and appropriate duration of cycling [18, 22, 29, 32].

Ankle position during cycling is one of the more important factors for effective pedaling [11, 18], yet this has not received much research attention previously. Some studies have examined the ankle angle and the ankle general muscle moment (GMM) producing PO during cycling. For example, Pierson-Carey et al. [18] reported less power production during cycling with the ankle locked in a neutral position. The study was theoretically supported by Van Soest et al. [30] based on a forward dynamic modeling/simulation. Ferrante et al. [8] reported that the limited knee flexion generated by the calf muscle due to the presence of an orthosis that fixed the ankle angle might have reduced the maximum PO. A slight increase in PO could be achieved by releasing the ankle joint with the rotation point between the foot and crank at plantarflexion (PF) [30]. On the other hand, Gregor et al. [11] reported that the ankle GMM of AB individuals during free ankle voluntary recumbent cycling remained plantarflexed throughout the pedaling cycle at a low workload. As the workload increased, the ankle GMM became more plantarflexed especially during the power phase of cycling at top dead center (TDC), \(0^\circ\) to bottom dead center (BDC) and \(180^\circ\) [11]. Trumbower and Faghri [28] also reported the same result with a fixed pedal boot as there was passive PF during the mid-to-late recovery phase induced by the inertial boot pedal forces in two AB participants at zero resistance level. The plantarflexor GMM is important to efficiently transmit the forces from the hip and knee to the crank and pedal, respectively [11]. The force transmission from the muscles to the pedal during cycling is then utilized to characterize the pedal forces [28]. Therefore, the inability to plantarflex the ankle could cause reduced force transmission from the hip and knee to the pedal and, thus, reduces the PO during cycling [18]. Taken together, these studies have further shown the importance of investigating maximum PO as a function of ankle movements during cycling.

In contrast, a limited number of studies have investigated AFO-constrained ankle movements on the power production during voluntary recumbent cycling in AB participants [11, 28]. However, it is not clear whether the different AFO-constrained ankle movements would influence the PO during voluntary recumbent cycling, given that all other cycling parameters are constant. It is an important concern in the rehabilitation systems to elicit maximum PO during recumbent cycling. Therefore, the purpose of this study was to investigate whether a solid AFO [fixed position (FP) AFO]-, articulated AFO (DPF AFO), AFO with PF stop at neutral position (DF AFO) and an AFO with DF stop at neutral position (PF AFO)-constrained ankle movements might influence cycle PO during voluntary recumbent cycling. Our hypothesis is that differently constrained ankle movements might alter the production of PO during recumbent cycling, as the biomechanics are affected by the ankle movement patterns [10]. The findings of the study might serve as a reference for future rehabilitative cycling protocols where both ankle muscle stretching and strength training are the simultaneous aims.

Materials and methods

Participants

Twenty-five healthy participants, 6 males (22.7 ± 1.9 years and 71.2 ± 14.1 kg) and 19 females (21.6 ± 0.9 years and 58.7 ± 13.8 kg), participated in this study. Participants were invited as volunteers and were screened for their health conditions prior to the experiment. All participants provided their written informed consent before taking part in the study. Individuals without any previous or ongoing record of neurological, musculoskeletal, rheumatological and cardiovascular disorders or orthopedic lower limb injuries were included. All the participants were untrained and unfamiliar with recumbent cycling. This study was approved by the UMMC Medical Ethics Committee, University Malaya Medical Centre, Kuala Lumpur, Malaysia [Ref 100314(1)].

Experimental setup

A recumbent cycle ergometer (BerkelBike Pro, BerkelBike B.V., St. Michielsgestel, The Netherlands) with its front wheel fixed to rollers during cycling was utilized in this study (Figure 1). Four AFOs (FP, AFOs...
DPF, DF, and PF AFOs) with different ankle movements (Figure 2) were fabricated using the same measurement of original AFO provided with the BerkelBike, where all the participants’ legs could fit into it. The AFOs were fabricated using 5 mm polypropylene and the types of ankle joints used were Tamarack (DPF AFO) and Oklahoma (DF and PF AFOs) joints. The lower legs of each participant were placed in the AFO that was affixed to a force-sensing pedal (Garmin Vector, Garmin Ltd., Kansas City, MO, USA) through a custom-made footplate. The footplate, which was fixed to the bottom of the AFO, was connected to the pedal (Figure 3). It allowed the AFOs to be unscrewed from the pedal to change to another AFO. During cycling, FP AFO was used to fix the ankle angle at the neutral position (90°), DPF AFO allowed the ankle to move from the neutral position to both DF and PF, DF AFO allowed the ankle to move from the neutral position to DF and PF AFO allowed the ankle to move from the neutral position to PF. The distance between seat position and the crank axle was adjusted for each participant according to their height and leg length. The distance was adjusted until the knee joint angle of each participant reached 160° at the BDC (180° was where the knee joint was fully extended). The backrest was standardized to 45° as
this angular posture was reported to provide maximum PO during recumbent cycling [21]. The backrest angle was measured using an analog goniometer. To measure the joint angles, a two-dimensional approach using a single video camera was used to capture the markers placed at the right shoulder, hip, knee, ankle and fifth metatarsophalangeal joint [11]. The marker placements for the ankle and fifth metatarsophalangeal joints were on the AFO. The upper limb positions were standardized between participants as shown in Figure 1.

Data collection protocol

The participants performed minimum loaded cycling within the set cadence range using visual feedback. The cadence range was measured using the Garmin Edge 510 (Garmin Ltd., Kansas City, MO, USA) placed on the front part of the cycle. Each participant was required to perform cycling with the FP, DPF, DF and PF AFOs in random order for 1 min each followed by 5-min recovery periods. A 1-min cycling with instructed speed ranging of 60–80 revolutions per minute (rev/min) was set for each constrained ankle movements to extract maximum PO during cycling. Power phase was defined from TDC to BDC whereas recovery phase was defined from BDC to TDC [11].

All participants performed recumbent cycling with DF AFO, but only data of 20 participants who used FP AFO, and 17 used DPF AFO and PF AFO, were retained and analyzed, because not all were able to maintain their cadence within the set cycling cadence.

Data processing and analysis

The cycle PO and cadence of each 1-min cycling session was recorded wirelessly (ANT+ module, Garmin Ltd., Kansas City, MO, USA) using a commercial data acquisition unit (Garmin Edge 510) and software (GoldenCheetah, v3.1) (open source project) to store the data into a PC for offline analysis. The PO was obtained directly from the force-sensing pedal (Garmin Vector). The outcome measurement is in watts (W). Static and dynamic calibrations of the force-sensing pedal were done beforehand to maximize the accuracy of the sensor. The angles of the hips, knees and ankles were recorded at 120 Hz. The last 1 min of the event was synchronized, extracted and further analyzed using the software Kinovea (0.8.15) (open source project). The video was synchronized with the sensing pedal since the beginning of the experiment using a timer. The peak and average POs of each constrained ankle movement during the entire cycling period for each participant were used for further analysis. One-way repeated measures analysis of variance (ANOVA) was performed to analyze the effect of PO generated by the ankle movement at a significance level of $\alpha = 0.05$. All data were statistically analyzed using the software SPSS (IBM SPSS Statistics, v20, New York, NY, USA).

Results

There were no significant differences between POs generated from pedal forces at cadences ranging between 60 and 80 rpm. Figures 4 and 5 portray the peak and average PO values generated by the FP-, DPF-, DF- and PF-constrained ankle movements during recumbent cycling. The peak and average POs in all constrained ankle movements was $27.2 \pm 12.0$ W (range 6–60) and $17.2 \pm 9.0$ W (range 2–36), respectively, with only a 14.6% variance between mean data among AFOs. The present study observed that there were no significant differences in the peak pedal PO [$F(3, 75) = 2.31, p = 0.083, \eta^2 = 0.085$] and the average pedal PO [$F(3, 75) = 2.54, p = 0.063, \eta^2 = 0.0992$] during AFO-constrained ankle movements during recumbent cycling.

There were also no significant differences in the changes of the hip and knee joint angles with different AFOs ($p = 0.974$ and $p = 1.00$, respectively). However, there was significant difference in the changes of the ankle joint angles.
angle ($p < 0.01$). The hip, knee and ankle joint excursion angles are presented in Figure 6. It can be clearly seen that only ankle ROM are different among AFOs (min–max) (FP: $90.0–90.0^\circ$, DPF: $84.8\pm8.1–95.1\pm5.5^\circ$, DF: $82.7\pm7.8–91.8\pm10.7^\circ$, PF: $95.0\pm7.3–102.9\pm9.7^\circ$).

**Discussion**

The present study sought to investigate possible differences in peak and average POs generated by the FP, DPF, DF and PF AFOs that constrained ankle movements in specific ranges during recumbent cycling. The peak and average pedal POs revealed in the current study was $27.2\pm12.0$ W and $17.2\pm9.0$ W, respectively. To our knowledge, no studies have as yet investigated the effect of ankle-constrained movements on the PO during voluntary recumbent cycling. However, a few studies had investigated the PO during voluntary recumbent cycling for 30 s. Duffell et al. [6] reported that the peak PO achieved by AB participants was $311.6\pm24.2$ W, while Martin and Brown [14] reported that the average PO achieved by well-trained AB cyclists was $540\pm31$ W. One of the reasons that the current study revealed a lower peak and average PO may be due to the ankle immobilization itself, as has been previously suggested [8]. However, a second explanation for the lack of differences between AFOs might be that only...
minimal power was required during the unloaded cycling [11] compared to a higher resistance [6], as changes in workload would necessarily alter PO [11, 22]. Our study selected minimal loaded cycling to allow a more direct comparison with data of clinical populations of interest [11]. Well-trained cyclists [14] might also have led to the contribution of higher PO compared to the untrained participants in this study. However, the previous studies on recumbent cycling focused on lower total POs (30–65 W) [11], which was almost similar with the peak pedal PO in the current study.

The present study also reported that there were no significant differences in the peak and average PO values among the different types of AFO-constrained ankle movements. The results refuted our initial hypothesis that the ankle movements might alter the cycling PO as ankle pattern affected movement kinematics [10]. This may be because the main components of PO are the knee and hip extensors and flexors but not the ankle movements [11, 29]. The ankle acts primarily to transmit force produced from the upper leg to the crank and less as a power generator [5, 11, 28]. Joint power was the product of joint moment and corresponding velocity, while pedal power was the product of the cadence and forces applied by the leg to the pedal and crank. Van Soest et al. [30] expected that there would be no large increase in PO during free ankle cycling in reality. In this study, the PO was a direct measure of the effective force which is perpendicular to the crank. As illustrated in Figure 3, the AFO is mounted directly onto the pedal; thus, the direction of propulsion is also mostly in line with the direction of the measured force.

To our knowledge, there are a limited number of studies that have investigated the influence of ankle angle on the PO production during cycling [11, 18]. It is important to note that the FP, DPF, DF and PF ankle movements did not have any influence on the hip and knee angle changes during cycling (Figure 6). Van Soest et al. [30] also reported the same finding as ours that the hip and knee joint angles were not significantly affected by releasing the ankle joint. This further justified that the changes in PO generated during the voluntary recumbent cycling, as found in this study, was more associated with ankle movements. In this study, however, it was revealed that the ankle PF for DFP, DF and PF ankle movements occurred earlier (early-to-mid-power phase), where the knee reached maximum extension at the mid-power phase (90° of crank angle). With ankle mobilization, Trumbower and Faghri [27] reported that ankle PF occurred at the mid-to-late power phase. This is an important finding where the constrained ankle movements can cause an alteration of the ankle PF during the power phase of recumbent cycling at a minimal workload, by assisting earlier PF and, thus, propulsion potential during power phase. In addition, Gregor et al. [11] also reported that the knee was extended during the first 90° of pedal revolution as the hip continued to extend until the end of the power phase; which was similar to the present study. In addition, the present study also revealed that all the participants were able to control their cadence within the set cycling cadence with the DF AFO compared to the other AFOs. Therefore, the DF AFO might be effective for cycling training if the goal of the training was speed-performance oriented.

Overall, our results may be useful in the field of rehabilitation therapy in eliciting increased PO during cycling training. For example, these data could promote the development of improved lower limb training for people with musculoskeletal or neuromuscular disorders such as stroke or SCI in order to benefit from therapy using functional electrical stimulation-based recumbent cycling [9, 23, 30]. Finally, the results could lead to a better understanding of AFO-assisted ankle joint biomechanics in producing maximum pedal PO during recumbent cycling.

In summary, different types of AFO-constrained ankle movements allowed increase in ankle ROM while maintaining PO productions during voluntary recumbent cycling. Further work is needed to investigate whether the FP, DPF, DF and PF AFO-constrained ankle movements might influence lower limb PO during recumbent cycling in individuals with pathological conditions.

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References