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Energy cost optimized dorsal leaf ankle-foot-orthoses reduce impact forces on the contralateral leg in people with unilateral plantar flexor weakness

N.F.J. Waterval [a,*], M.A. Brehm [a], J. Harlaar [b,c], F. Nollet [a]

[a] Amsterdam UMC, University of Amsterdam, Rehabilitation Medicine, Amsterdam Movement Sciences, Meibergdreef 9, Amsterdam, The Netherlands
[b] Department of Biomechanical Engineering, Delft University of Technology, Delft, The Netherlands
[c] Department of Orthopaedics, Erasmus Medical Center, Rotterdam, Netherlands

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ABSTRACT

Background: In individuals with unilateral plantar flexor weakness, the second peak of the vertical ground reaction force (GRF) is decreased. This leads to a higher ground reaction force, e.g. impact, of the contralateral leg, potentially explaining quadriceps muscle and/or knee joint pain. Energy cost optimized dorsal leaf ankle-foot-orthoses (AFOs) may increase the push-off ground reaction force, which in turn could lead to lower impact forces on the contralateral leg.

Research questions: 1) Are impact forces increased in the contralateral leg of people with unilateral plantar flexor weakness compared to healthy subjects? 2) Do energy cost optimized AFOs reduce impact forces and improve leg impact symmetry compared to walking without AFO in people with unilateral plantar flexor weakness?

Methods: Nine subjects with unilateral plantar flexor weakness were provided a dorsal leaf AFO with a stiffness primarily optimized for energy cost. Using 3D gait analyses peak vertical GRF during loading response with and without AFO, and the symmetry between the legs in peak GRF were calculated. Peak GRF and symmetry were compared with reference data of 23 healthy subjects.

Results: The contralateral leg showed a significant higher peak vertical GRF (12.0 ± 0.9 vs 11.2 ± 0.6 N/kg, $p = 0.005$) compared to healthy reference data. When walking with AFO, the peak vertical GRF of the contralateral leg significantly reduced (from 12.0 ± 0.9 to 11.4 ± 0.7 N/kg, $p = 0.017$) and symmetry improved compared to no AFO (from 0.93 ± 0.06 to 1.01 ± 0.05, $p < 0.001$).

Conclusion: In subjects with unilateral plantar flexor weakness, impact force on the contralateral leg was increased when compared to healthy subjects and dorsal leaf AFOs optimized for energy cost substantially reduced this force and improved impact symmetry when compared to walking without AFO. This indicates that dorsal leaf AFOs may reduce pain resulting from increased impact forces during gait in the contralateral leg in people with unilateral plantar flexor weakness.

1. Introduction

Neuromuscular disorders, such as poliomyelitis or peripheral nerve damage, may cause unilateral plantar flexor weakness. During gait, plantar flexor weakness reduces the center of pressure progression during the 2nd rocker of the stance phase and impedes the 3rd rocker, also reducing the ankle push-off power [1,2]. This results in an increased and early weight shift to the contralateral side as also observed in amputee gait [3]. This early and increased loading may cause pain in the knee joint and/or quadriceps muscles of the contralateral leg [4]. In clinical practice, such pain is often reported to limit daily life activities by people with unilateral plantar flexor weakness [5].

In people with plantar flexor weakness, knee joint pain in the contralateral leg might emerge due to a higher and more abrupt first peak ground reaction force, e.g. impact force, as the body decelerates the increased downwards velocity when shifting weight to the contralateral leg [1,6–8]. In healthy gait, push-off force is essential for a smooth transition from one leg to the other by accelerating the center of mass upwards prior and during the double support phase [9]. In subjects with an impaired push-off force, this upward acceleration of the center-of-mass is reduced. Consequently, the leading leg's downward velocity is not decelerated, and the leg collides with the ground at a

* Corresponding author.
E-mail address: n.f.waterval@amsterdamumc.nl (N.F.J. Waterval).

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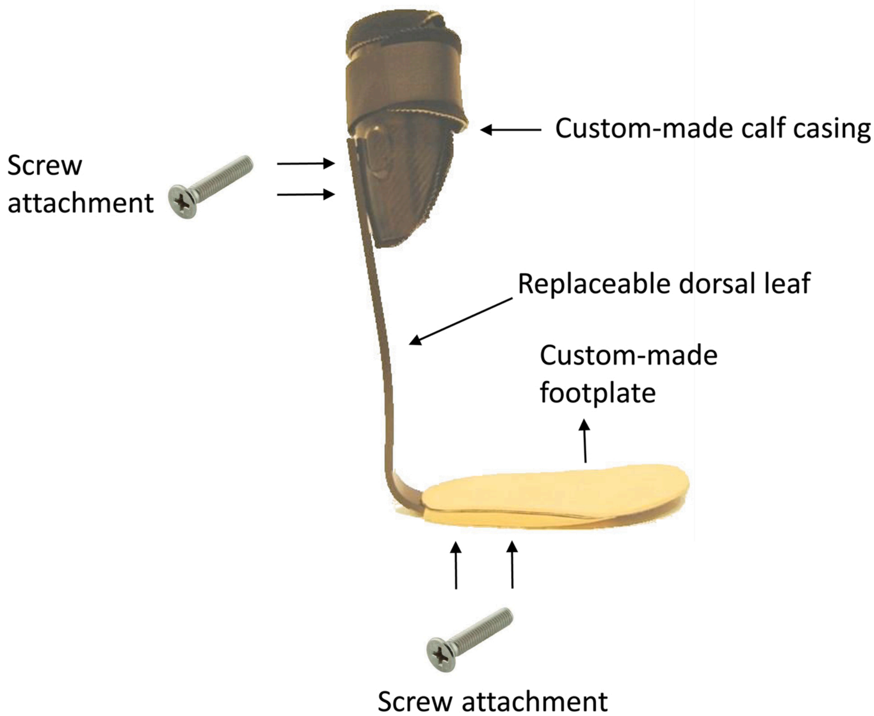

Fig. 1. Dorsal leaf AFO with an individually optimized stiffness.

relatively high velocity, causing higher impact forces [1,7]. In healthy subjects, higher impact forces are associated with the development of knee joint pain [6]. Additionally, higher impact forces potentially lead to a higher first peak in the external knee adduction moment, which has been shown to contribute to the development of knee joint pain and osteoarthritis in healthy subjects and people with a trans-tibia amputation [7,10,11]. Additionally, higher impact forces may increase the external knee flexion moment which might explain quadriceps overload often reported in people with plantar flexor weakness [4].

To improve gait in people with plantar flexor weakness, ankle-foot orthoses (AFOs) are commonly provided [12–14]. In general, AFOs successfully improve walking speed and restrain excessive ankle dorsiflexion, which reduces the downward velocity of the center of mass during the weight shift. However, most conventional AFOs negatively affect ankle push-off, which limits push-off force [12,13] and, hence their effect on lowering impact force of the contralateral leg [15]. Unlike conventional AFOs, dorsal leaf AFOs can support the push-off force by storing energy in the stance phase and returning this energy during push-off [16–18]. The increase in push-off force though depends on the stiffness of the dorsal leaf AFO, and we previously demonstrated the importance of individually optimizing the stiffness to maximally reduce energy cost and other gait outcomes [18–21]. The increase in push-off force with energy cost optimized dorsal leaf AFOs likely improves the upward acceleration of the center-of-mass during the double support phase [8,22] and may therefore lower the impact force and knee adduction and flexion moments on the contralateral leg.

Studies on the impact force and knee adduction and flexion moment of the contralateral leg in people with unilateral plantar flexor weakness and on the effectiveness of AFOs on reducing these forces are lacking so far. Consequently, it is unknown whether impact force and knee adduction and flexion moments in the contralateral leg can be reduced by applying energy cost optimized dorsal leaf AFOs. The aims of this study were to assess the impact force and knee adduction and flexion moments in the contralateral leg of people with unilateral calf muscle weakness compared to healthy subjects, and to determine whether dorsal leaf AFOs of which the stiffness was optimized for energy cost reduce the impact force, improve impact symmetry and reduce the knee adduction and flexion moments of the contralateral leg by increasing push-off force of the affected leg.

2. Methods

For this study, we used data of the PROOF-AFO trial that studied the effect of stiffness-optimized dorsal leaf AFOs on gait in subjects with plantar flexor weakness [23]. The medical ethics committee of the Academic Medical Center (AMC) in Amsterdam, The Netherlands, approved the study protocol and the trial was registered in the Dutch trial register, NTR5170.

2.1. Participants

In the PROOF-AFO trial, both subjects with uni- and bilateral calf muscle weakness were included. For the purpose of this study, only data of subjects with unilateral calf muscle weakness who walked without an assistive device, such as a walking stick or crutch, were used to ascertain complete registration of the ground reaction forces. Inclusion criteria were; the presence of calf muscle weakness due to a neuromuscular disorder determined by a Medical Research Council (MRC) score below 5 and/or not able to perform a single leg heel-rise [4,24], aged 18 years or older, using an AFO or orthopedic shoes in daily life, able to walk for at least 6 min and weight below 120 kg. Subjects were excluded in case of knee extensor weakness requiring a knee-ankle-foot orthosis and presence of a pes equinus (i.e. not able to reach ankle dorsiflexion during weight bearing). As healthy subjects, we used the reference data of our gait laboratory from 23 subjects without a history of lower limb injuries.

2.2. Intervention: energy cost optimized dorsal leaf AFO

Participants of the PROOF-AFO trial were provided with a new dorsal leaf AFO (Fig. 1), consisting of a Carbon Ankle7 dorsal leaf (Ottobock, Duderstadt, Germany) and custom-made calf casing and footplate. The stiffness of the AFO could be varied by manually changing the dorsal leaf and was measured using BRUCE [25]. As part of the PROOF-AFO trial, the AFO-stiffness was experimentally optimized for each individual based primarily on maximally reducing energy cost. In case no clear optimal stiffness for energy cost reduction was found, a difference less than 5% compared to all other stiffness configurations, the ankle angle, ankle power, knee angle and knee moment curves were judged by three independent assessors to determine the optimal stiffness. A consensus meeting was held when the assessors did not agree on the optimal stiffness based on 3D gait data. The protocol, algorithm and effects of the stiffness-optimization on walking have been published before [18,20,23]. For the purpose of this study, only the data of the optimal stiffness AFO condition were considered, referred to as energy cost optimized dorsal leaf AFO, as well as data of the shoes only condition (i.e. without AFO).

2.3. Measurements

2.3.1. Gait biomechanics

Gait kinematics and kinetics at comfortable speed while walking without AFO and with the stiffness-optimized AFO were assessed using 3D-gait analysis. Markers were placed according to the PlugInGait model, which is a common marker model consisting of 7 trunk markers, 3 pelvis markers and 6 markers on each lower limb (Vicon Motion Systems, Oxford, UK; [26]). A knee alignment device and additional marker on the medial malleolus was used to calibrate the knee and ankle-axis during a static pose. After calibration participants walked over ground on a 12-m walkway, while gait was recorded with a 12-camera 100 Hz Vicon MX 1.3 system (VICON, Oxford, UK). Ground reaction forces were measured for one stride in the middle of the walkway using two force plates (1000 Hz, OR6-7, AMTI, Watertown, USA). Measurements continued until we recorded three trials where both feet were placed completely within a separate force plate and all markers visible for the complete stride.

2.3.2. Muscle strength testing

A trained physician manually assessed muscle strength of the plantar flexors of both the affected and contralateral leg according to the Medical Research Council (MRC) scale (score range 0–5) [24]. Additionally, the maximal isometric plantar flexor strength of both legs was tested using a fixed dynamometer (Biodex, Corp., Shirley, NY). The ankle of the participants was fixed in 15° plantarflexion, while their shank was placed horizontally. We recorded three maximal voluntary contractions (MVCs) with an interval of 30 s. The highest recorded peak value (in Nm) was considered the maximal plantar flexor strength.

2.4. Data analysis

2.4.1. Gait biomechanics

Using the video-recordings and force-plate data, the timing of heel-strike and toe-off were determined. Impact force, defined as the peak ground reaction force during loading response, and push-off force, defined as the peak ground reaction force during push-off, the maximal ankle push-off power, maximal external knee adduction and external knee flexion moment during the loading response and the knee adduction and knee flexion impulse were calculated for affected and contralateral leg in subjects with plantar flexor weakness and for the right leg in healthy subjects. Variables were averaged for each leg across the three recorded trials using Matlab (The Mathworks, Natick, USA). Leg impact symmetry was calculated as the ratio of the impact force of the affected leg and the contralateral leg. To account for differences in stride length, we also calculated the knee adduction and knee flexion impulse per meter by dividing the impulse by stride length. Additionally, walking speed was determined by dividing stride length by stride time.

2.5. Statistical analysis

To test whether impact and push-off force, maximal ankle power, maximal external knee adduction and external flexion moment and impulse of the affected and contralateral leg of subjects with unilateral plantar flexor weakness differed from those in healthy subjects, independent *t*-tests were used. To test the effect of the energy cost optimized dorsal leaf AFO in subjects with unilateral plantar flexor weakness on impact and push-off force, impact symmetry, maximal ankle power and maximal external knee adduction and external flexion moment and impulse for affected and contralateral legs, paired samples *t*-tests were conducted.

Table 1
Participant and AFO characteristics.

	Subjects with plantar flexor weakness	Healthy subjects
Male / female	5/4	11/12
Age in years*	51.2 ± 15.7	38.7 ± 13.9
Weight in kg*	86.6 ± 20.0	71.4 ± 10.6
MRC plantar flexors (contralateral leg / affected leg) median [IQR]	5 [5–5] / 4 [3–4.5]	5 [5–5]
Maximal strength plantar flexors Nm (contralateral / affected leg) Median [IQR]	43 [35–54] / 18 [13–21]	–
Optimal AFO stiffness in Nm/degree	3.8 ± 1.1	–

MRC = Medical Research Council score, IQR = inter quartile range.
* Data presented as mean ± SD.

3. Results

From the PROOF-AFO trial, data of nine subjects with unilateral calf muscle weakness were used for analysis. Subjects were diagnosed with polio (n = 3), nerve damage (n = 3), spinal stenosis (n = 2) and radiculopathy (n = 1). Subjects with plantar flexor weakness were significantly older ($p = 0.037$) and heavier ($p = 0.008$) compared to the reference group of healthy subjects (Table 1).

3.1. Impact forces in subjects with plantar flexor weakness versus healthy subjects

When walking without AFO, subjects with plantar flexor weakness had a significant 23.3% lower walking speed compared to healthy subjects (1.05 ± 0.17 vs 1.37 ± 0.14 m/s, $p < 0.001$). The contralateral leg of subjects with plantar flexor weakness showed a significantly higher impact force compared to healthy subjects and significant lower push-off force, while no difference in maximal ankle power was found. For the affected leg of subjects with plantar flexor weakness, impact force did not differ significantly from healthy subjects, while the push-off force and maximal ankle power were significantly lower (see Table 2).

The contralateral leg of subjects with plantar flexor weakness showed a significantly higher knee adduction moment during loading response, knee adduction impulse and knee adduction impulse corrected for stride length compared to healthy subjects. Knee flexion impulse and impulse corrected for stride length were also higher compared to healthy subjects, while maximal knee flexion moment during loading response was not. For the affected leg of subjects with plantar flexor weakness, maximal knee adduction moment or impulse did not differ significantly from healthy subjects, while the knee adduction impulse corrected for stride length was significantly higher for the affected leg. Also for the affected leg, knee flexion impulse and impulse corrected for stride length were higher compared to healthy subjects, while maximal knee flexion moment during loading response was not (Table 2).

3.2. Effect of the AFO on impact forces and symmetry

When using the AFO, walking speed of subjects with plantar flexor weakness increased significantly by 10.5% to 1.16 ± 0.15 m/s ($p = 0.021$) compared to walking without AFO. Impact force of the

Table 2
Comparison between healthy subjects and contralateral and affected legs of subjects with plantar flexor weakness.

	Healthy subjects	Contralateral leg subjects with plantar flexor weakness	*P value* Contralateral leg versus healthy subjects	affected leg subjects with plantar flexor weakness	*P value* affected leg versus healthy subjects
Impact force in N/kg	11.2 ± 0.6	12.0 ± 0.9*	*0.005*	11.2 ± 0.8	*0.841*
Push-off force in N/kg	11.4 ± 0.7	10.2 ± 0.5*	*< 0.001*	9.3 ± 0.4*	*< 0.001*
Maximal ankle power W/kg	3.48 ± 1.02	3.17 ± 1.57	*0.519*	1.42 ± 0.74	*< 0.001*
Maximal knee adduction moment during loading response in Nm/kg	0.65 ± 0.12	0.79 ± 0.22*	*0.026*	0.67 ± 0.27	*0.771*
Knee adduction impulse in Nm/s	0.18 ± 0.06	0.26 ± 0.05*	*0.002*	0.21 ± 0.08	*0.315*
Knee adduction impulse corrected for stride length in Nm/s	0.12 ± 0.04	0.22 ± 0.04*	*< 0.001*	0.17 ± 0.06*	*0.008*
Maximal knee flexion moment during loading response in Nm/kg	0.57 ± 0.21	0.64 ± 0.25	*0.492*	0.56 ± 0.28	*0.890*
Knee flexion impulse in Nm/s	0.12 ± 0.05	0.16 ± 0.05*	*0.021*	0.17 ± 0.08*	*0.021*
Knee flexion impulse corrected for stride length in Nm/s	0.08 ± 0.03	0.14 ± 0.04*	*< 0.001*	0.14 ± 0.06*	*< 0.001*

Data are presented as mean±standard deviation.
* Significant different from healthy subjects.

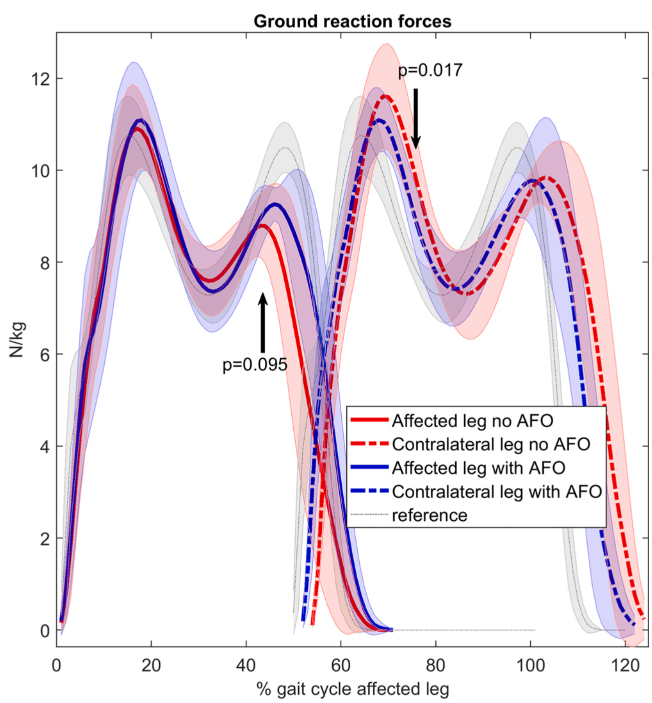

Fig. 2. Effect of energy cost optimized dorsal leaf AFOs on ground reaction force.

contralateral leg decreased significantly when using the AFO compared to without AFO, while no effect of the AFO on push-off force or maximal ankle power of the contralateral leg was found. For the affected leg, no significant effect of the AFO on impact, push-off force or ankle power was found (Figs. 1 and 2, Table 3). Use of the AFO improved the impact symmetry between the legs from 0.93 ± 0.06 without AFO to 1.01 ± 0.05 with AFO ($p < 0.001$).

No significant effect of the AFO was found on the maximal knee adduction moment or impulse of the contralateral leg, while the knee adduction impulse corrected for stride length significantly decreased when walking with AFO compared to without AFO. Further, the AFO reduced the knee flexion impulse and impulse corrected for stride length while no significant effect on the maximal knee flexion moment was found. For the affected leg no effect of the AFO on knee adduction and knee flexion moment, impulse or corrected impulse was found (Fig. 2 and Table 3).

4. Discussion

This study demonstrated that the gait pattern of subjects with unilateral plantar flexor weakness contains risk factors for the development of pain in the contralateral leg due to a higher impact force and increased knee adduction and knee flexion impulses during body weight shift from the affected to the contralateral leg, when compared to healthy subjects. The provision of energy cost optimized dorsal leaf AFOs reduced these risk factors and normalized impact symmetry between legs despite an increase in walking speed. These findings indicate that energy cost optimized dorsal leaf AFOs may contribute to the prevention or reduction of pain in the contralateral leg.

Our finding that impact force of the contralateral leg was increased despite a lower walking speed in subjects with unilateral plantar flexor weakness compared to healthy subjects can explain the overuse and/or pain in the contralateral leg, often reported by subjects with calf muscle weakness [5,6]. As indicated by dynamic walking models, a reduced second peak of the vertical ground reaction force during push-off from the weakened leg leads to a higher velocity of the center-of-mass during foot-ground contact of the contralateral leg, and hence higher impact forces on this leg [6,9,27]. Additionally, as expected, also the knee adduction and knee flexion impulse were higher in the contralateral leg of subjects with plantar flexor weakness compared to healthy subjects. Both the higher impact force and higher knee adduction moment are risk factors for the development of knee osteoarthritis, as found in people with an unilateral trans-tibia amputation [7,10], and may contribute to increased knee valgus. Prevalence of knee osteoarthritis or excessive knee valgus in neuromuscular diseases have not been reported to our knowledge, but overuse symptoms of the knee and quadriceps muscle are common [4] which may be related to the increased knee flexion impulse.

Provision of energy cost optimized dorsal leaf AFOs improved impact symmetry between the legs by reducing impact forces of the contralateral leg during the loading response, despite increasing walking speed. Previously, in healthy subjects, it was shown that an increase in walking speed of 0.1 m/s increased the peak GRF during loading response with 0.05 N/kg [28]. In our participants, the GRF decreased by 0.6 N/kg (5%), while speed increased by 0.11 m/s, indicating that the effect is substantial. Additionally, the fact that impact symmetry normalized suggests that the increased risk of overload injuries due to impact stress in the contralateral leg may be diminished by providing the AFO.

We hypothesized that the reduction in impact force and normalization of impact symmetry would be the result of an increase in push-off forces of the affected leg [18] caused by the force of the spring-like properties of the AFO during the push-off. However, no significant effect of the AFO on push-off force or maximal ankle power was found,

Table 3
Effect of the AFO on impact force, push-off force and knee adduction and flexion moment.

	Contralateral leg without AFO	Contralateral leg with AFO	*P value*	Affected leg without AFO	Affected leg with AFO	*P value*
Impact force in N/kg	12.0 ± 0.9	11.4 ± 0.7*	*0.017*	11.2 ± 0.8	11.4 ± 1.0	*0.281*
Push-off force in N/kg	10.2 ± 0.5	10.3 ± 0.6	*0.649*	9.3 ± 0.4	9.5 ± 0.4	*0.095*
Maximal ankle power in W/kg	3.17 ± 1.57	3.11 ± 1.26	*0.827*	1.42 ± 0.73	1.64 ± 0.42	*0.233*
Knee adduction moment during loading response in Nm/kg	0.79 ± 0.22	0.75 ± 0.17	*0.257*	0.67 ± 0.27	0.61 ± 0.25	*0.199*
Knee adduction impulse in Nm/s	0.26 ± 0.05	0.24 ± 0.07	*0.091*	0.21 ± 0.08	0.20 ± 0.08	*0.180*
Knee adduction impulse corrected for stride length in Nm/s	0.22 ± 0.04	0.19 ± 0.05*	*0.009*	0.17 ± 0.06	0.16 ± 0.07	*0.380*
Maximal knee flexion moment during loading response in Nm/kg	0.64 ± 0.25	0.54 ± 0.17	*0.212*	0.56 ± 0.28	0.71 ± 0.41	*0.064*
Knee flexion impulse in Nm/s	0.16 ± 0.05	0.14 ± 0.06*	*0.030*	0.17 ± 0.08	0.18 ± 0.08	*0.964*
Knee flexion impulse corrected for stride length in Nm/s	0.14 ± 0.04	0.11 ± 0.05*	*0.006*	0.14 ± 0.06	0.14 ± 0.06	*0.902*

Data are presented as mean ± standard deviation.
* Significantly different from walking without AFO.

although in 6 out of 9 participants both push-off force and power improved. In two participants the push-off force did not increase and these participants had the largest push-off force without AFO with 9.71 and 9.75 N/kg, respectively, while maximal ankle power decreased substantially in the 2 subjects (in one also the force did not increase) with a maximal ankle power above 2 W/kg without AFO. Likely, in these subjects restriction of the ankle range of motion by the AFO impaired the push-off force generation of the muscles [12,18], which apparently outweighed the beneficial effect of the spring-like properties of the AFO. In the other participants, push-off forces and maximal ankle power increased, suggesting that in the majority of patients the assistance of the AFO led to a lower impact force on the contralateral leg and normalized impact symmetry. This assistance of the AFO also becomes clear from Fig. 2, where, when using the AFO, the ground reaction force of the contralateral leg started to rise earlier (contralateral heel-strike) while the ground reaction force on the affected leg was still high. Furthermore, the ground reaction forces cross each other at a higher force, demonstrating that the sum of the ground reaction forces of both legs was higher throughout the transition phase with the AFO, indicating less energy was lost during the transition phase [22,29].

Despite a lower impact force, no effects of the AFO on maximal knee adduction moment or knee impulse were found, likely due to the higher walking speed. An increase of 0.11 m/s is expected to increase the knee adduction moment with approximately 15% [30], while in our subjects knee adduction moment and impulse did not change despite such increase in walking speed. When correcting the knee adduction impulse for stride length we did find a decrease, indicating that the lack of effect indeed can be explained by differences in walking speed and stride length. This is supported by research at a fixed speed in trans-tibia amputees where a slight increase in push-off force already decreased the knee adduction moment [7]. Contrary, a reduction in knee flexion impulse was found when using the AFO, likely because the largest component of the ground reaction force is in this direction. Reducing knee flexion impulse likely reduces quadriceps activation and hence, the risk of overloading this muscle.

A strength of our study is that all participants used the same AFO with an optimized stiffness for energy cost reduction, thereby avoiding potential differences between subjects due to differences in AFO design or inadequately matched AFO stiffness levels. This implies that the reported effects are the maximal effects that can be achieved by current state-of-art passive dorsal-leaf AFOs. However, the current method to optimize the stiffness are extensive and time-consuming, making it

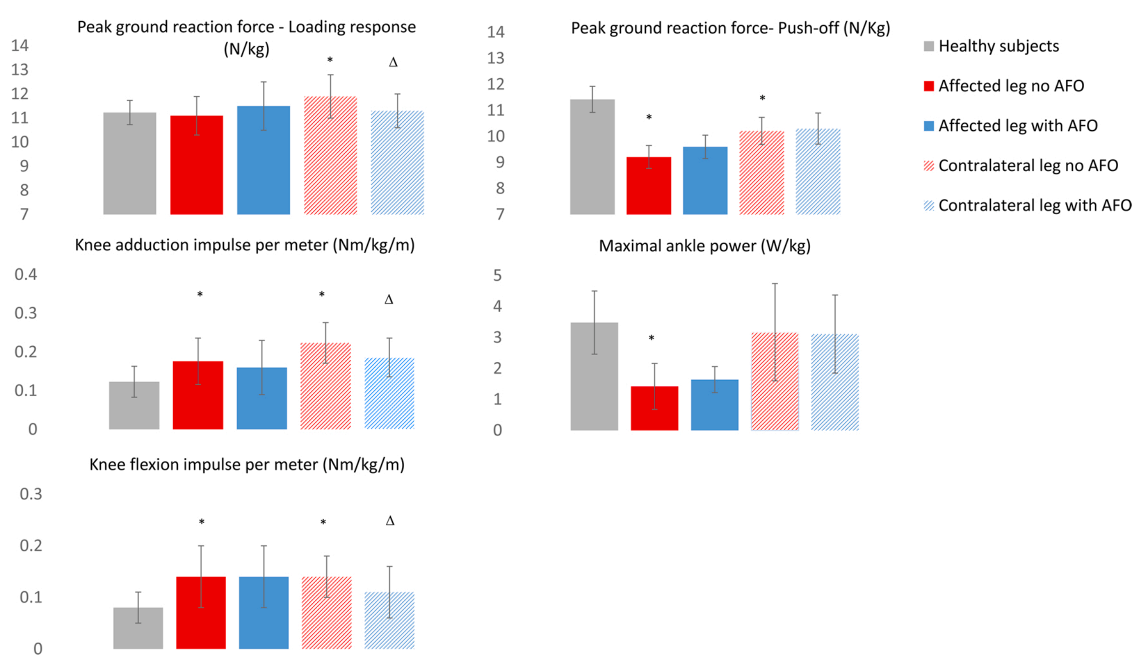

Fig. 3. Ground reaction force and knee adduction moments. * denotes a significant difference with healthy subjects. Δ denotes a significant effect of the AFO.

challenging to implement in usual-care. As for 8 out of 9 subjects the optimal stiffness was between 2.8 and 4.3 Nm/degree, the duration of the protocol might be reduced by testing less stiffness levels. Future research should focus on predicting the optimal stiffness using simulations or further speed up the optimizations using human-in-the-loop optimization [31].

We only measured the effect of AFOs at comfortable speed while it would have been insightful to also assess the effects at a fixed speed to account for its confounding effect [32]. The fact that healthy subjects were heavier might have confounded our effects as well. Additionally, we included only 9 patients with a substantial heterogeneity in plantar flexor muscle weakness (and underlying disease). Consequently, differences in baseline gait patterns as indicated by the large standard deviations in Fig. 3 were seen, which limits the power to identify significant differences and may have contributed to the lack of significant findings for push-off force and ankle power. Furthermore, our participants had varying degree of remaining muscle force and high MRC scores for patients diagnosed with a neuromuscular disease which may limit generalizability to subjects with full paralysis. However, none of the participants could perform a single heel-rise or provided a push-off force larger than 9.8 N/kg, indicating no functional push-off can be generated. Therefore, future research should match larger numbers of patients and healthy controls for important characteristics such as bodyweight, and perform measurements at multiple speeds. Future studies should furthermore assess whether impact forces and moments can also be reduced by other AFO designs.

In conclusion, in subjects with unilateral plantar flexor weakness, the contralateral leg is at risk of developing overload injuries due to higher impact forces and knee adduction and knee flexion impulses. Provision of energy cost optimized dorsal leaf AFOs substantially reduce these forces. In the majority of patients this could be explained by an increase in push-off force of the affected leg. Energy cost optimized dorsal leaf AFOs may contribute to the reduction of overload symptoms and prevention of long term joint degeneration of the contralateral leg in subjects with unilateral plantar flexor weakness due to neuromuscular disorders.

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Conflict of interest

None to declare.

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