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ABSTRACT
Ankle-foot orthoses (AFOs) can improve walking mobility in individuals with calf muscle weakness, but their impact on sit-to-walk, a common daily activity, is underexplored. Using predictive simulations, we tested the effects of AFO stiffness on sit-to-walk in case of different degrees of plantarflexor weakness. Results showed that AFO stiffness significantly affects sit-to-walk kinematics and kinetics, with an optimal stiffness for minimizing effort. This optimum stiffness depends on severity of weakness and seat height. These findings emphasize the need to assess and consider the effects of AFO stiffness in daily life activities besides walking.

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KEYWORDS
Ankle foot orthosis; plantar flexor weakness; sit to walk; predictive simulations; neuromuscular diseases

Background

People with calf muscle weakness, e.g. weakness of the Soleus and/or Gastrocnemius muscle, due to neuromuscular diseases (Vinci & Perelli, 2002; Don et al. 2007), are often provided ankle-foot orthoses (AFOs) to improve daily functioning (Ploeger et al. 2020). Calf muscle weakness results in an inability to control an external dorsiflexion moment as the muscles can generate a lower maximal moment (Perry et al. 1995; Neptune et al. 2001; Ploeger et al. 2017). As a consequence walking ability and stability is reduced (van Duijnhoven et al. 2024). To improve walking, AFOs that support the plantarflexors by generating an external moment are provided (Ploeger et al. 2014). The magnitude of the external moment, and hence the effect of these AFOs on walking depends on the AFO stiffness, for example, how easy the AFO bends (Totah et al. 2019; Waterval et al. 2019). By increasing AFO stiffness more support to the plantarflexors is provided at the expense of ankle range-of-motion (Totah et al. 2019). In case of calf muscle weakness, reducing the ankle dorsiflexion angle to reference values requires relatively stiff AFOs (Skigen et al. 2023; Waterval et al. 2021; Waterval et al. 2020), which may reduce the suitability and usability of the AFO during other activities of daily life (Zuccarino et al. 2021; Bashir et al. 2022).

Hindrance or restriction of AFOs during activities of daily living is an important reason for discontinued AFO use (Ploeger et al. 2020; Zuccarino et al. 2021). According to AFO users rising up from a chair is the most important activity after walking (Van Der Wilk et al. 2018), and is performed approximately 60 times a day (Dall & Kerr, 2010). For people with calf muscle weakness performing a so-called sit-to-walk (STW) requires 30–50% more muscular effort compared to healthy subjects (Caruthers et al. 2020). Ideally, AFOs would, besides walking, also make STW less burdensome.

The effect of AFOs on STW effort may, similar to walking, be highly depended on the AFO stiffness (Waterval et al. 2019). With increasing stiffness, initially STW effort may reduce as the AFO takes over the calf muscle function when the center-of-mass moves forward without imposing too much restriction on the ankle range of motion. However, the relatively stiff AFOs required to support walking may be too restrictive. In people with myelomeningocele with lower leg weakness, habitual ankle restricting AFO users demonstrate more difficulty standing up from a chair compared to people with myelomeningocele

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who do not use an AFO (Bartonek et al. 2024). The limited ankle range of motion in the AFO users hampered forward movement during the rising phase, thereby altering the movement pattern and effort to rise from a chair (Bartonek et al. 2024). Consequently, an U-shaped relation between AFO stiffness and effort during STW can be expected.

It is advised to acknowledge and take into account the effort to perform STW with AFOs in clinical practice (Bartonek et al. 2024). To adjust AFO stiffness provision or refer people to physiotherapy for learning altered STW movements (Van der Heijden et al. 2009), knowledge on the effect of AFO stiffness on effort and movement is needed. However, the expected U-shape relation between AFO stiffness and effort likely also depends on factors such as severity of weakness and seat height. With an increase in severity of weakness, more resistance of the AFO is needed to allow forward progression of the center-of-mass and, hence, the optimum stiffness for effort reduction will increase similar to walking (Waterval, van der Krogt, et al. 2023). Contrary, a lower seat height requires more ankle dorsiflexion to rise from a char likely reducing the optimum stiffness for STW (Mazzà et al. 2004).

Experimentally testing our hypothesis regarding how AFO stiffness effects STW is time-consuming, burdensome for the patients and challenged by inter-individual differences in impairment. With predictive neuromuscular simulations, variables can be studied independently and have previously been used to study the effects of AFO stiffness on walking (Kiss et al. 2024; Waterval, Brehm, et al. 2023). Recent research developed a validated planar neuromuscular controller to simulate adaptation strategies in STW in healthy individuals (van der Kruk & Geijtenbeek, 2023a, 2023b). Building on these studies, the aim of our study is to investigate (1) how AFO stiffness affects STW in terms of kinetics and kinematics, effort quantified as muscle activation and energy consumption and balance quantified by angular momentum and (2) how this differs between seat heights and severities of calf muscle weakness.

Methods

Simulation framework

We simulated the STW movement using a previously published model and controller (van der Kruk & Geijtenbeek, 2023a, 2023b). The used neuromuscular model had a height of 1.80 m and mass of 75 kg. The model featured 7 joints (ankle, knee and hip in each leg, and a lumber joint), 11 degrees of freedom (1 around ankle, knee, hip and lumbar joint, and 3 between pelvis and ground) and 10 Hill-type muscles per leg (Tibialis anterior (TA), soleus (SOL), gastrocnemius medialis (GAS), vasti (VAS), rectus femoris (RF), hamstring (HAM), biceps femoris short head (BFSH), gluteus maximus (GMAX), iliacus (ILIAC) and psoas (PSOAS)). Peak isometric are based on the lower limb model by Delp et al. (1990) (G2392) (Delp et al. 1990). To account for the reduction in number of muscles, we combined the peak isometric forces of muscles performing similar movements in the sagittal plane. The model was initially developed as an OpenSim3 model and translated into Hyfydy (Geijtenbeek, 2021). Hyfydy models are specifically developed for fast forward simulations. The lumbar and thoracic joint were controlled by motor torques, which mimic the lumbar and thorax muscle groups.

Contact forces between the feet and the ground and between the pelvis and the chair were modelled with a Hunt Crossley force spheres and box, respectively. The contact spheres sizes, strain modulus and friction coefficients were set as previously published (van der Kruk & Geijtenbeek, 2023b). Each foot had two contact spheres one at the heel and one at the toe, with a radius of 3 centimeter and strain modulus of 17,500 N/m^2. At the pelvis a contact sphere with a radius of 12 centimeter and strain modulus of 1000 N/m^2 represented the buttocks. The chair was modelled as a box ($40 \times 12 \times 5$ cm). The model is shown in Figure 1.

We simulated the STW movement for 2.5 s to ensure that the rising phase and initiation of the first step were within the simulation timeframe for all conditions despite potential differences in movement speed. The simulation of the STW movement used a stand-up controller and gait controller in sequence (van der Kruk & Geijtenbeek, 2023b). The stand-up controller consisted of 2 phases, while the gait controller consisted of 3 phases (stance, lift-off and swing). Both controllers operate on a reflex-control principle each with its own set of control parameters. The controllers contain both monosynaptic and antagonistic proprioceptive feedback (force, length, and velocity) from the muscles. Additionally, vestibular feedback loops controlling the pelvis tilt and velocity were applied to the muscles surrounding the hip. The gains of the reflexes were optimized separately per controller and phase. The timing of the transition between the phases of the controllers and between the controllers were optimized as part of the optimization problem. In total 551 control parameters were optimized, which includes the muscle reflex gains (force, length and velocity), offsets for muscle

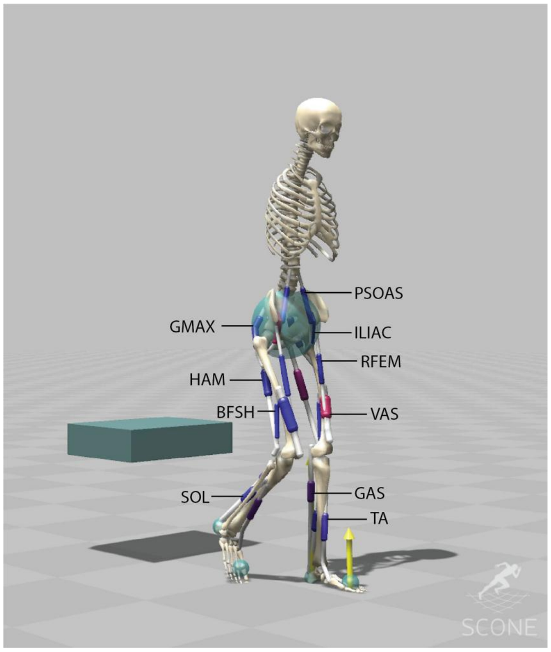

Figure 1. The planar model used for the simulations. The figure is based on van der Kruk and Geijtenbeek (2024). A planar neuromuscular controller to simulate compensation strategies in the sit-to-walk movement. *PLoS one*, *19*(6), e0305328. Abbreviations: GMAX = gluteus maximus, RFEM = rectus femoris, vas = vasti, HAM = hamstrings, BFSH = biceps femoris short head, SOL = soleus, GAS = gastrocnemius and TA = tibialis anterior.

length feedback (l_o), proportional feedback of the lumbar and thoracic joints regarding the vestibular feedback, transition times between the phases within the standing-up and gait controller, and the transition time between the controllers. A detailed description of the specific reflexes can be found in van der Kruk and Geijtenbeek (2023b).

The control parameters were optimized by minimizing a comprehensive objective measure, which included metabolic energy calculated based on (Umberger, 2010) (J_{mb}), cubed muscle activation(J_{act}) and torque from the thorax motors (J_T) as proxy for trunk muscles. We included cubed muscle activation to highly penalize maximal activation, necessary to avoid compensations. Besides these measures, joint range of motion, knee limit force and head acceleration were taken into account, but all reached zero during the optimization process. Consequently, the objective function is represented by the following formula ($J_{total} = \omega_{mb}J_{mb} + \omega_{act}J_{act} + \omega_T J_T$). Weight of the objective measures were set based on the previous work: Metabolic cost, $\omega_{mb} = 0.01$, Cubed muscle activation, $\omega_{act} = 0.1$ and Trunk torque $\omega_T = 0.0003$ for all conditions. Thes measure was the same as used in previous published STW controller (van der Kruk & Geijtenbeek, 2023b). reached zero during the optimization.

For the optimization we used the OpenSource software SCONE (Geijtenbeek 2019), using the Hyfydy simulation engine (Geijtenbeek, 2021). A Covariance Matrix Adaptation Evolutionary Strategy (CMA-ES), with a population size of 10 for each generation was applied (Hansen et al. 2003). The simulation was terminated when the progress became smaller than 0.01, indicating that the objective function outcome did not reduce anymore. For each simulation condition we executed three parallel optimizations with the same initial guess. Three parallel optimizations were used to minimize the effect of local minima. Of these three, the final simulations with the lowest objective function score was used for analysis.

Conditions

We simulated STW without AFO and with bilateral AFOs with ten stiffness levels. The AFO was modelled as a massless linear rotational spring around the ankle, in line with previous studies (Waterval, Brehm, et al. 2023; Waterval, van der Krogt, et al. 2023). Modelling the AFO this way replicates a conventional AFO with a fixed stiffness that is the same towards both plantar- and dorsiflexion. Stiffness was varied between 0.9 (50 Nm/rad) and 8.8 Nm/degree (500 Nm/rad), in steps of 0.9 Nm/degree (50 Nm/rad), which represents the clinically applied AFO stiffness range (Shuman & Russell Esposito, 2022; Waterval et al. 2020). We had two different seat conditions: normal seat (45 cm) and low seat (35 cm), which is the range of commercially available seats (Weiner et al. 1993). To include plantarflexor muscle weakness, we created a model with severe and moderate calf muscle weakness. Severe plantar flexor weakness was imposed by reducing the maximal isometric force of SOL and GAS bilaterally from 3549 N and 2241 N to 1000 N (71% reduction) and 500 N (77% reduction) respectively. A higher reduction in GAS was chosen as fat infiltration is higher in GAS compared to SOL (Stilwell et al. 1995). For moderate calf muscle weakness the maximal isometric forces of SOL and GAS were doubled compared to severe and set to 2000 N and 1000 N respectively.

All conditions were simulated with the standard initial foot position where both feet were symmetrically on the ground with the ankle in 13.7 degrees dorsiflexion and the knee in 85 degrees flexion which was based on experimental data (van der Kruk et al. 2022).

Analysis

To adjust for differences in STW speed, and hence number of steps taken in the 2.5 simulated seconds, we

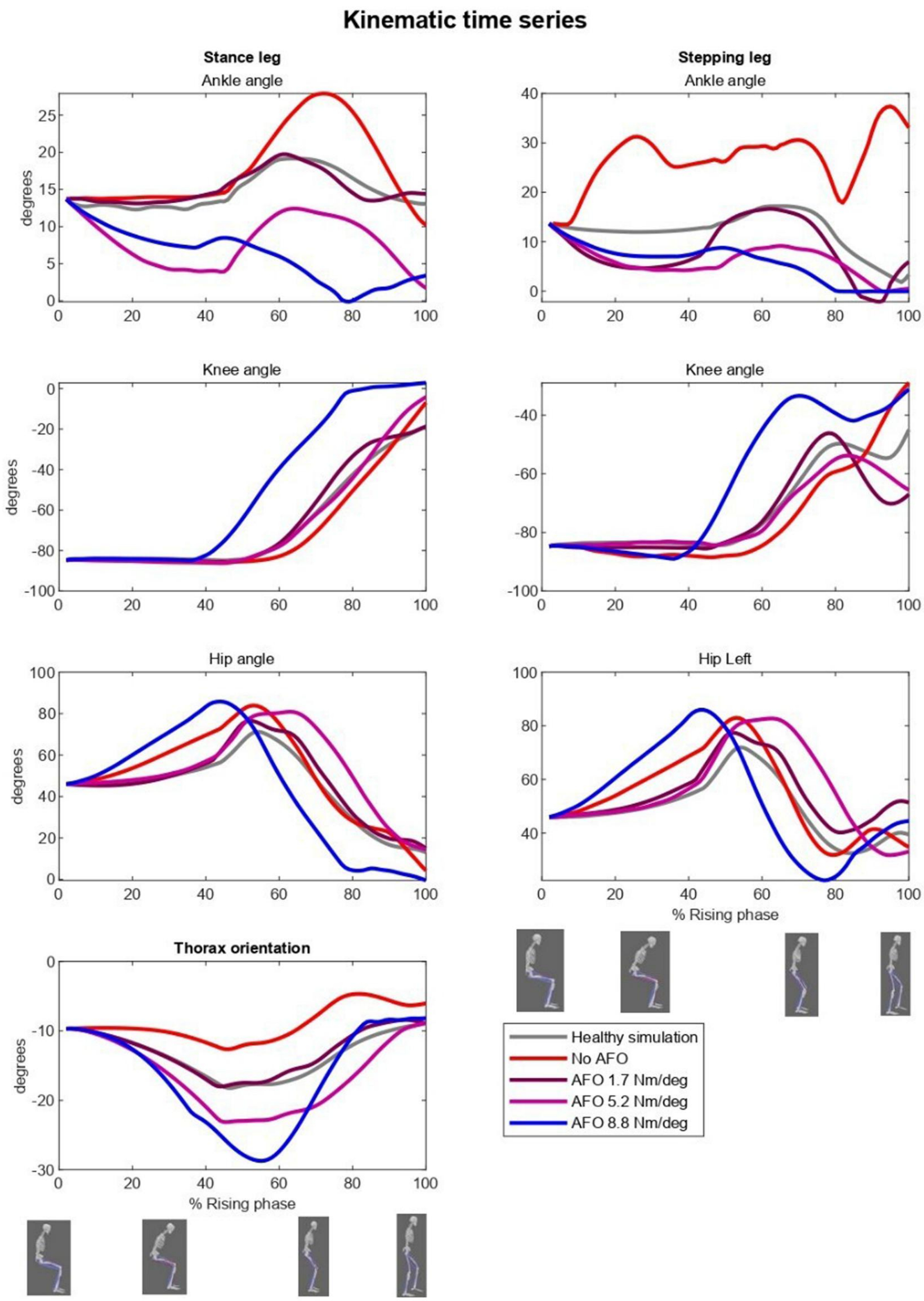

Figure 2. The kinematics time-series of different ankle-foot orthosis stiffness levels during the sit-to-walk task. For visibility reasons only the conditions with no AFO, and AFO with a stiffness of 1.7, 5.2 and 8.8 Nm/degree are shown.

analysed all simulations from the start of the simulation until the highest point of the Center-of-Mass before the first step. The leg that swings forward to take the first step is considered the stepping leg, the other the standing leg. From the start till the highest point of the Center-of-Mass, we determined the peak lower limb joint angles and moments, as well as the minimal trunk flexion compared to the global axis (e.g. the forward lean of the model). Additionally, during the same timeframe, the integral of energy consumption, muscle

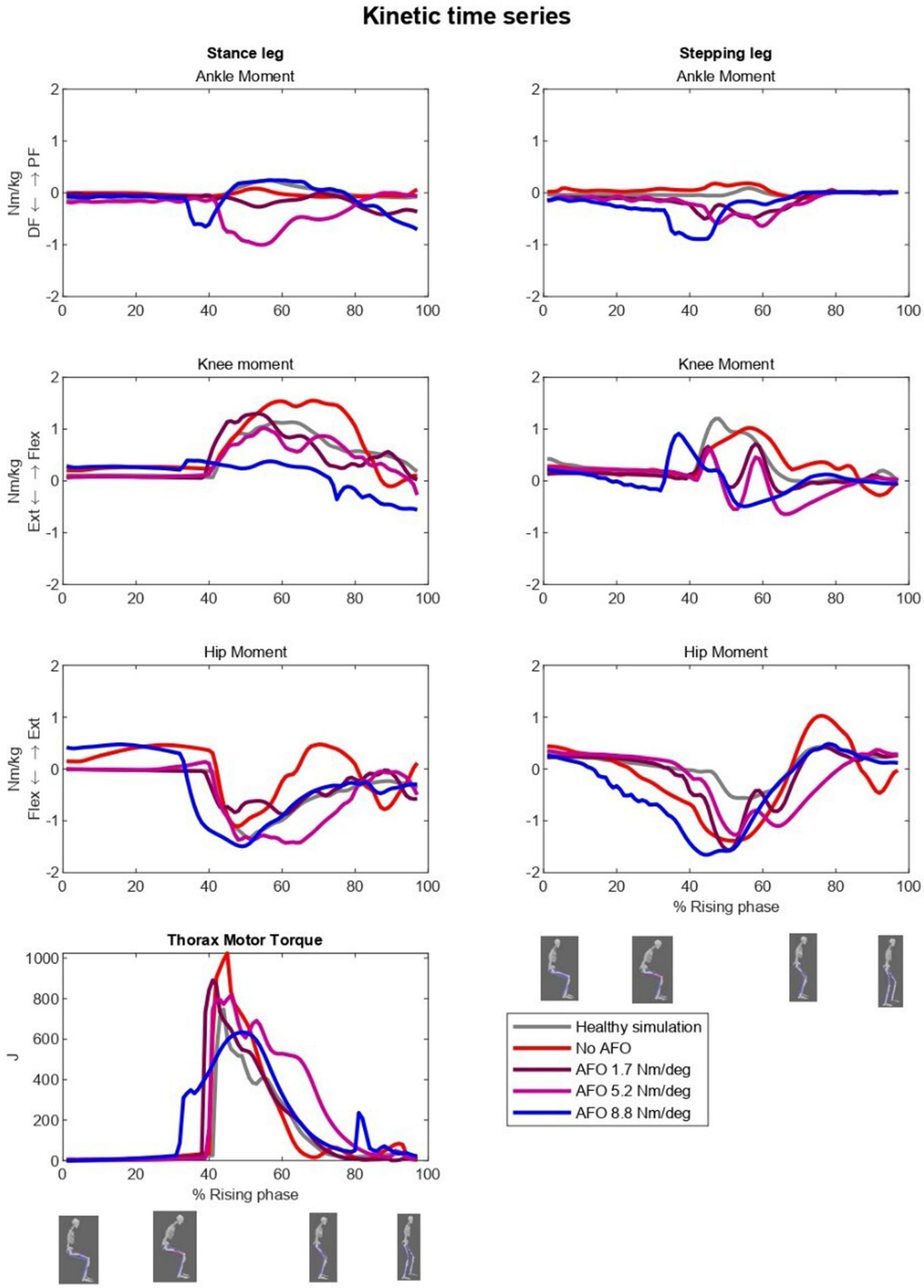

Figure 3. The kinetics time-series of different ankle-foot orthosis stiffness levels during the sit-to-walk task. For visibility reasons only the conditions with no AFO, and AFO with a stiffness of 1.7, 5.2 and 8.8 Nm/degree are shown.

activation (of whole body and separate muscles) and thorax motor torque was calculated.

To determine the relation between AFO stiffness and the outcome measures a regression was applied. For joint kinetics and kinematics a linear fit was applied, while for energy consumption and muscle activation a quadratic fit was used. This choice was based on the relation between these outcomes and AFO stiffness during walking (Kiss et al. 2024). The lowest point of the

Table 1. R values for the regression analysis.

	Stance leg		Stepping leg	
	R	Equation	*R*	Equation
Kinetics and kinematics (Linear)				
Max ankle dorsiflexion (degree)	0.92	−1.66x + 13.57	0.55	−1.22x + 13.01
Max knee extension (degree)	0.75	−2.16x + 17.6	0.35	−1.28x + 44.68
Max hip extension (degree)	0.49	−1.09x + 14.26	0.69	−2.02x + 141.19
Trunk flexion (degree)	0.94	−1.55x +15.55		
Ankle moment (Nm)	0.95	5.84x + 15.05	0.96	5.79x + 21.82
Knee moment (Nm)	0.90	−8.21x + 109.72	0.05	0.31x + 70.29
Hip moment (Nm)	0.90	4.11x + 88.25	0.92	5.47x + 87.97
Trunk Torque (J)	0.89	1.18E05x + 1.03E06		
Muscle metabolics (Quadratic)				
Gastrocnemius	0.78	16.8 × 2 − 133.6x + 254	0.42	6.8x 2 − 98.3x + 650
Soleus	0.49	4.1x 2 − 25.0x + 86	0.49	−6.0x 2 + 82.0x + 14
Tibialis Anterior	0.76	13.2x 2 + 600.9x + 2500	0.65	138.0x 2 − 895.3x + 3554
Vasti	0.21	65.2x 2 − 750.2x + 22898	0.63	151.2x 2 − 757.1x + 8228
Rectus femoris	0.84	245.3x 2 − 2050x + 4904	0.83	77.8x 2 − 943.3x + 3692
Hamstrings	0.79	56.7x 2 + 33.2x + 5182	0.76	93.5x 2 − 467.3x + 4600
Biceps Femoris	0.68	49.9x 2 − 410.9x + 1281	0.28	−7.4x 2 + 21.4x + 891
Gluteus Maximus	0.53	197.8x 2 − 2034.0x + 7269	0.96	512x 2 − 3542.8x + 7028
Iliacus	0.21	3.7x 2 − 60.0x + 753	0.37	6.6x 2 − 77.7x + 1489
Psoas	0.31	8.6x 2 − 101.6x + 512	0.76	39.0x 2 − 372.8x + 1125

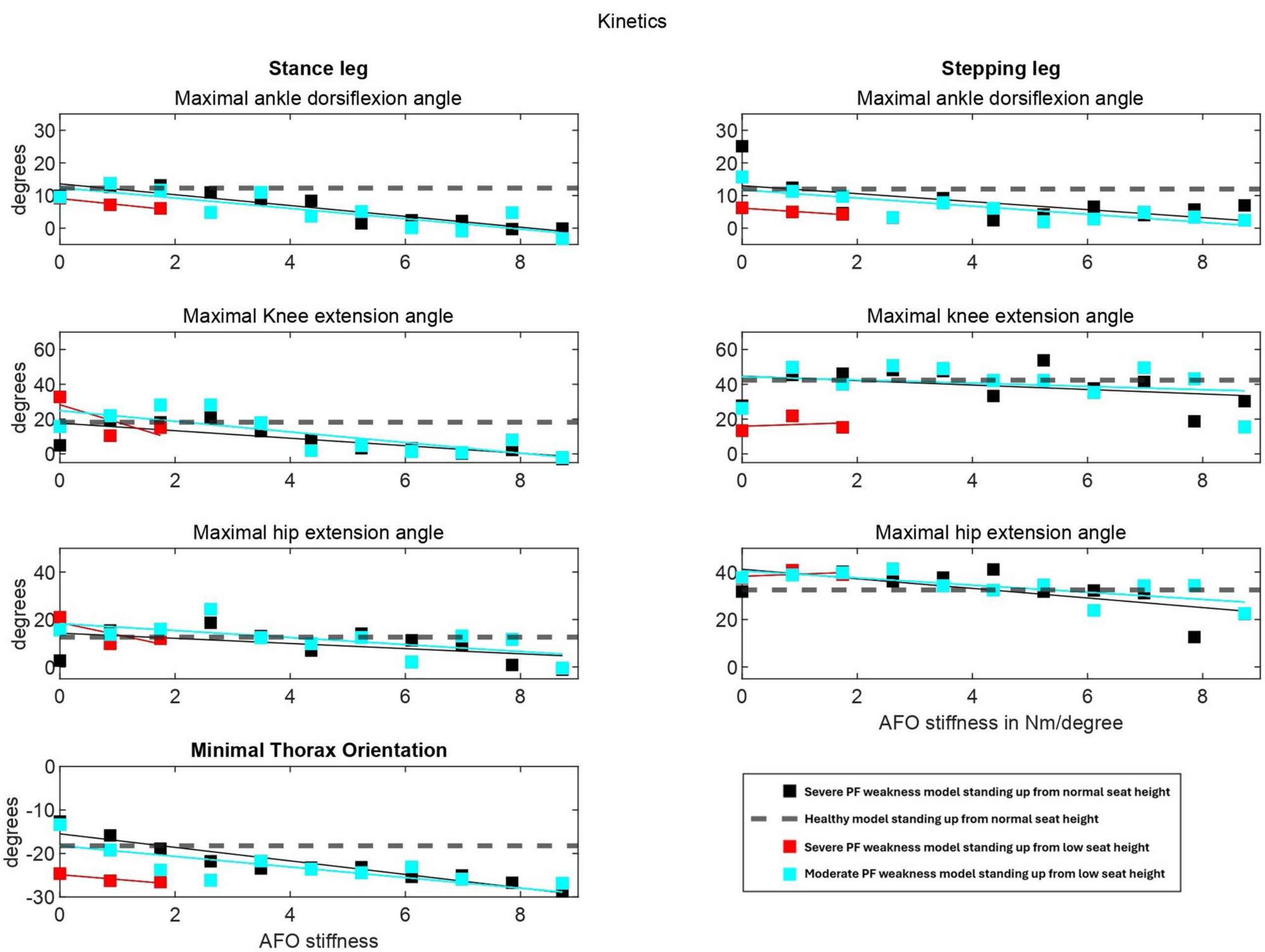

Figure 4. The maximal joint angles during the sit-to-walk task.

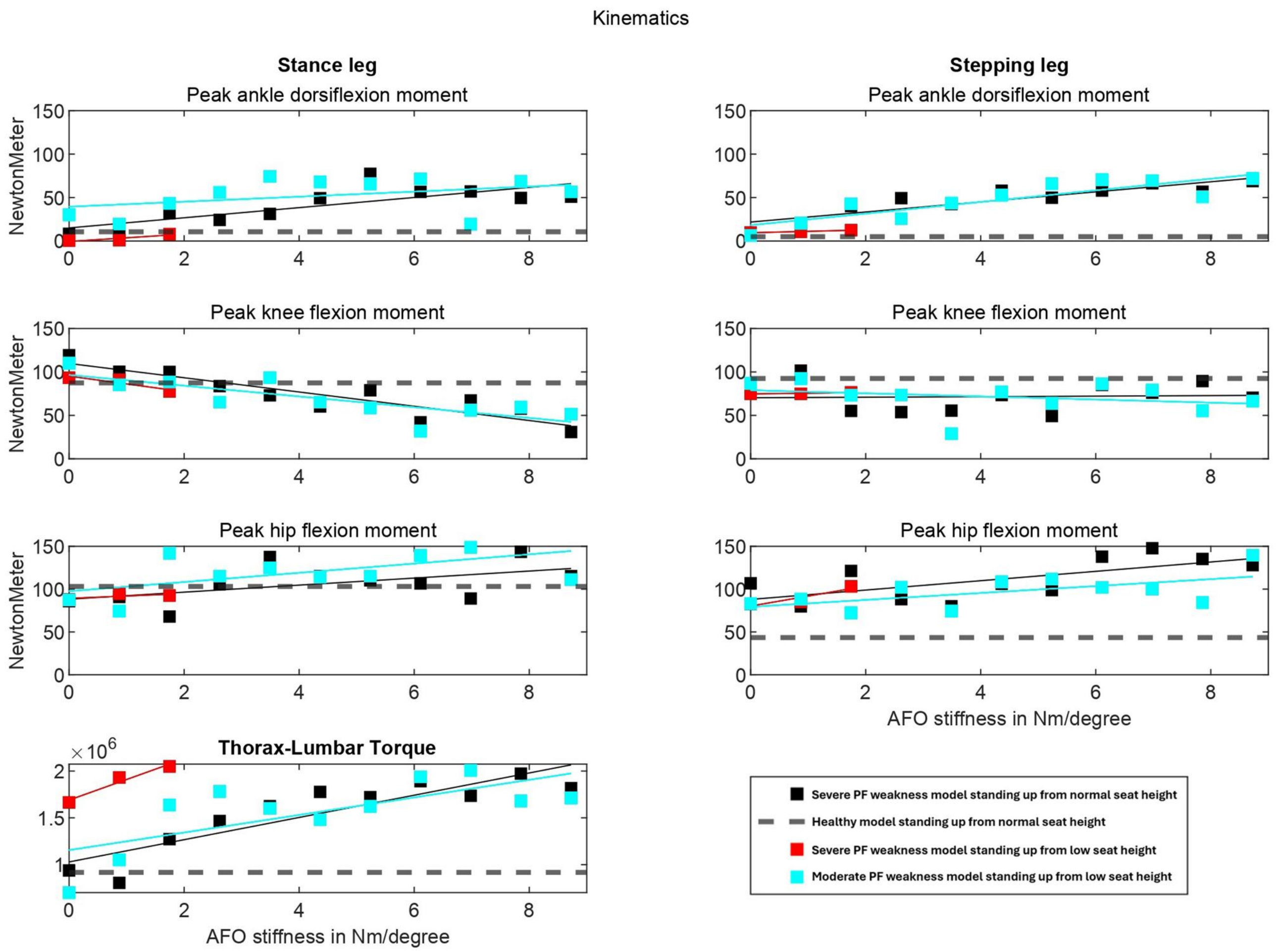

Figure 5. The maximal joint moments during the sit-to-walk task.

quadratic fit was considered the optimum AFO stiffness for the respective outcome measure.

Results

Effect of AFO stiffness

The simulated kinematics and kinetics of the conditions without AFO, stiffness of 1.7, 5.2 and 8.8 Nm/degree are shown in Figures 2 and 3. For visibility reasons, the other conditions are not shown.

The bilateral maximal ankle dorsiflexion, knee extension, and hip extension angle during STW were linearly related with increasing AFO stiffness ($r > 0.35$). With an increase in AFO stiffness from 0.9 to 8.8 Nm/degree, the maximal ankle dorsiflexion decreased by 12 degrees, the knee flexion by 22 degrees and hip flexion decreased by 16 degrees. Additionally, trunk flexion increased from 15 degrees to 29 degrees ($r = 0.94$). Regarding maximal joint moments, with increasing AFO stiffness, maximal external ankle dorsiflexion, knee flexion, and hip flexion moment increased linearly ($r > 0.9$), except for the knee flexion moment of the stepping leg which was not related to AFO stiffness ($r = 0.05$). Motor torque at the lumbar and thorax joint increased linearly with increasing stiffness, and was 2.25 times at large with a stiffness of 8.8 Nm/degree compared to 0.9 Nm/degree ($r = 0.88$) (Table 1; Figures 4 and 5).

AFO stiffness showed a strong quadratic relation with energy consumption during STW ($r = 0.90$) and summed muscle activation ($r = 0.82$). Without AFO, the model with calf muscle weakness consumed 44% more energy compared to the healthy condition. With increasing stiffness energy consumption decreased, with an optimum at 3.4 Nm/degree. At 3.4 Nm/degree energy consumption was 21.5% lower compared to without AFO. Regarding muscle activation the optimum stiffness was at 3.6 Nm/degree, during which summed muscle activation was 24.3% lower compared to without AFO (Figure 6).

At the individual muscle level, large effects of stiffness were found for the energy consumed by the vasti of the stepping leg and the hamstrings, gluteus maximus, rectus femoris, and tibialis anterior of both legs

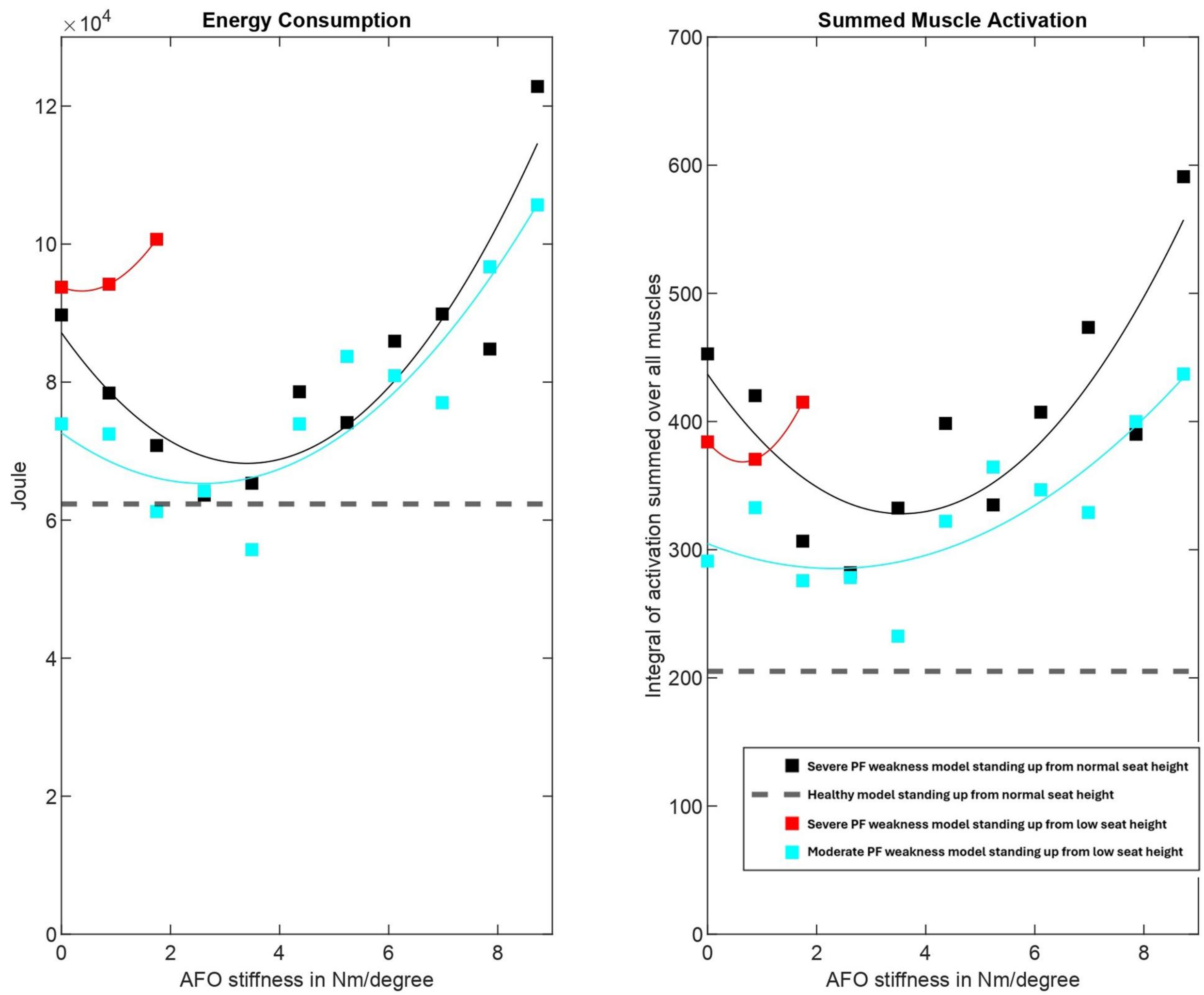

Figure 6. The relation between ankle-foot orthosis stiffness and muscle energy consumption and activation.

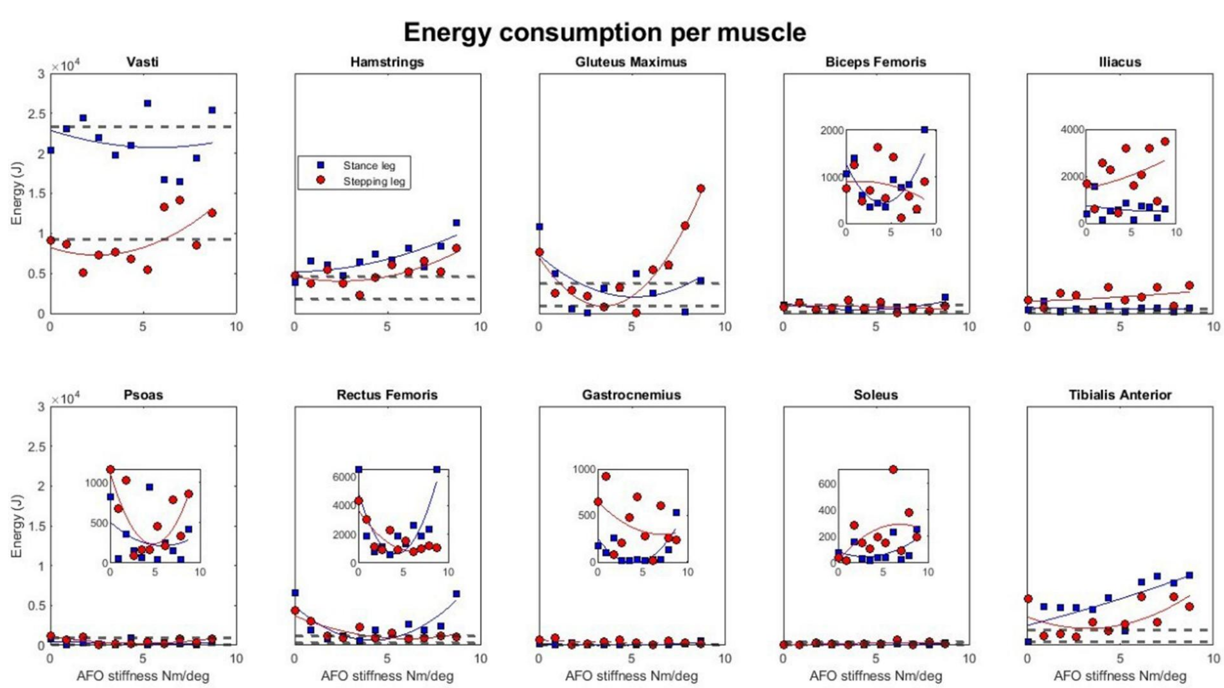

Figure 7. Relation between ankle-foot orthosis stiffness and muscular energy consumption.

($r > 0.53$) (Table 1; Figure 7). With increasing stiffness, energy consumed by the vasti of the stepping leg, and gluteus maximus and rectus femoris muscles of both legs first decreased before an increase was seen, demonstrating an optimum. For the hamstrings and tibilias anterior of both legs energy consumption increased from the first stiffness onwards. The energy consumption of the vasti of the stance leg showed a poor relation with AFO stiffness ($r = 0.22$), while the other muscles contributed little to whole body energy consumption.

Effect of seat height and severity of weakness

When reducing the seat height from 45 centimetres to 35 centimetres, the simulation was not able to find a solution to perform the STW for an AFO stiffness above 1.7 Nm/degree. When standing up from the lower seat height the trunk was more in flexion and a higher lumbar-thorax motor torque was used (Figures 3 and 4). When standing up from the low seat height, for all AFO conditions the energy consumption was higher compared to normal seat height (Figure 6).

Increasing muscle strength of the plantarflexors did not substantially alter the slope of the regression analysis regarding maximal joint kinematics and kinetics (Figures 3 and 4). The optimum stiffness for energy consumption decreased from 3.4 Nm/degree in case of severe weakness to 2.6 Nm/degree in the case of moderate weakness (Figure 6).

Discussion

The results of our simulations indicate that in people with bilateral calf muscle weakness, an U-shape relation exists between AFO stiffness and effort, demonstrating an optimum AFO stiffness for minimizing effort required from the lower limb muscles. As expected, increasing severity of weakness increased the optimal stiffness. However, lowering the seat height to 35 centimetres made it impossible for the neuromusculoskeletal model to complete the simulation when AFO stiffness exceeded 1.7 Nm/degree. The differences in optimum stiffness can be explained by the substantial effect of AFO stiffness on STW kinematics and kinetics.

In agreement with our hypothesis, an optimal stiffness regarding effort, measured by muscle activation and energy consumption, was found. The optimal AFO stiffness for minimization of muscular forces during STW was 3.6 Nm/degree. This is lower compared to the optimal stiffness for walking (between 4 and 5 Nm/degree) found in experimental and simulation studies (Waterval, Brehm, et al. 2023; Waterval, van der Krogt, et al. 2023) (Skigen et al. 2023; Waterval et al. 2020). In clinical care, AFO stiffness optimization is performed solely during level ground walking, while our simulations now demonstrate that the optimal stiffness can differ considerably between STW and walking. To take this discrepancy of required stiffness into account during the prescription process, future research should focus of how to incorporate AFO users value of certain daily life activities in AFO stiffness optimization algorithms. An adjacent research area is the development of adaptive AFOs. Orthosis and prosthesis using hydraulic or microprocessor systems to adapt to sloped environments or stairs have been recently developed, which, based on our findings, could benefit from expansion of STW possibilities as well.

The existence of an optimum AFO stiffness for STW effect can be explained by the effect of stiffness on kinematics and kinetics. Our model showed that increasing the AFO stiffness primarily lead to a restriction of the ankle range-of-motion, similar to what is found during walking (Totah et al. 2019). This resulted in a substantially altered STW movement pattern, where the trunk flexion keeps increasing with increasing stiffness. These compensations are needed to move the center-of-mass in front of the ankle (Bartonek et al. 2024). As a consequence of the increased trunk flexion, the moments generated around the ankle and hip and required thorax-lumbar torque increase almost twofold from no AFO to using an AFO with a high stiffness. This is similar to what is found in amputees, where trunk load is higher compared to healthy individuals (Actis et al. 2018). As STW is performed dozens of times a day (Dall & Kerr, 2010), the higher loads around the hips and trunk may contribute to overload complaints, especially as hip forces during walking are also increased (Waterval et al. 2018, 2022). Noteworthy is that severity of weakness did not seem to have any effect on the compensations with an AFO. The model with moderate weakness showed the same increase in trunk flexion and thorax-lumbar torque as the model with severe weakness, indicating that regardless of severity of weakness, the AFO induced substantial compensations and individuals may be at risk of overuse problems. On the other hand, lowering the seat-height substantially increased the required compensations when using an AFO. With a low seat height, trunk flexion and thorax-lumbar torque at 1.7 Nm/degree were similar to a stiffness of 8 Nm/

degree at normal seat height (see Figure 3&4). Potentially, the required flexion and thorax-lumbar torque became too high at higher stiffness levels, explaining why the model could not find a solution. In daily life, AFO users might switch strategy beforehand and start to rely more on their arms.

Our findings can improve an informed and individualized AFO stiffness prescription. As STW is an important activity for AFO users (Van Der Wilk et al. 2018), and the associated effort is substantial, minimizing the effort is important to reduce fatigue (Bartonek et al. 2024). Our simulations demonstrate that regardless of severity of weakness, the optimal stiffness for STW is lower compared to walking (Waterval, Brehm, et al. 2023). This indicates that especially in patients that are not community-dwelling and for whom walking long distances is less important a lower stiffness than optimal for walking may be provided. Furthermore, AFO users should be informed about the effect of seat height on effort as chairs that are too low put a high effort and load on the muscles potentially reducing the effect of AFOs on fatigue.

To simulate STW, we utilized a controller previously validated on experimental data of healthy young adults, older adults, and knee osteoarthritis (van der Kruk & Geijtenbeek, 2023a, 2023b), and added an AFO model previously validated specifically in people with calf muscle weakness (Kiss et al. 2024). With these relatively simple predictive simulations, we have demonstrated that there is likely an optimum regarding minimizing effort during STW. This provides a good reason and motivation to pursue further research. A future research direction should be on extending the model to 3D, incorporate arms and validate the simulations against patient data. Although these improvements are needed to make the models clinically applicable, in its current state the model can capture relevant general working principles. Our simulation framework has several assumptions that should be taken into account when interpreting the results. The most important is the fact that a planar model without arms was used. Although the largest joint movements are in the sagittal plane during STW, balance disturbances occur often in 3D which were not taken into account. In addition, increased arm push-off during STW is an important compensation strategy often used by older adults (Wheeler et al. 1985) and the arms may also substantially contribute to elevate the Center-of-Mass during the rising phase of STW. Hence, the lack of arms may limit the validity of our results in case arm support is available. In this scenario, not modelling the arms likely has exaggerated the trunk flexion, movement limitations, and/or lower leg activations to overcome the limited ankle range-of-motion in the ankle joint due to the AFO. However, not in all siting situations in daily life assistance of arm push-off is possible and in many neuromuscular diseases resulting in plantarflexor weakness, such as Charcot-Marie-Tooth, arm strength may also be impaired. Neglecting the arms may therefore still replicate a relevant scenario. In addition, our baseline model was based on Delp et al. (1990) (G2392) (Delp et al. 1990), mimicking a relatively old and weak male skeleton. This may limit the generalizability of the predicted AFO stiffness for people with other muscle strength or body properties. Despite our model simplifications and their inherent limitations, given the scarce research into STW with AFOs, our results still provide novel insights into STW that should be experimentally validated and tested.

Conclusion

In conclusion, our simulations show that in people with calf muscle weakness AFO stiffness substantial affects STW and that depending on individual characteristics, such as severity of weakness, and environmental factors, such as seat height, an optimal AFO stiffness for minimizing STW effort exists. As STW is the most important activity besides walking, this should be taken into account when prescribing AFOs and warrant the development of more complex AFO stiffness optimization algorithms or adaptive AFOs.

Disclosure statement

No potential conflict of interest was reported by the author(s).

Author contributions

CRediT: N. F. J. Waterval: Conceptualization, Formal analysis, Investigation, Methodology, Project administration, Software, Visualization, Writing – original draft; **E. van der Kruk**: Conceptualization, Methodology, Resources, Software, Supervision, Writing – review & editing.

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