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Effects of orthopedic footwear on postural stability and walking in individuals with Hereditary Motor Sensory Neuropathy

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ABSTRACT

Background: Orthopedic footwear is often prescribed to improve postural stability during standing and walking in individuals with Hereditary Motor Sensory Neuropathy. However, supporting evidence in literature is scarce. The aim of this study was to investigate the effect of orthopedic footwear on quiet standing balance, gait speed, spatiotemporal parameters, kinematics, kinetics and dynamic balance in individuals with Hereditary Motor Sensory Neuropathy.

Methods: Fifteen individuals with Hereditary Motor Sensory Neuropathy performed a quiet standing task and 2-min walk test on customized orthopedic footwear and standardized footwear. Primary outcome measures were the mean velocity of the center of pressure during quiet standing and gait speed during walking. Secondary outcome measures included center of pressure amplitude and frequency during quiet standing, and spatiotemporal parameters, kinematics, kinetics, and dynamic balance during walking. Two-way repeated measures ANOVA and paired *t*-tests were performed to identify differences between footwear conditions.

Findings: Neither quiet standing balance nor dynamic balance differed between orthopedic and standardized footwear, but orthopedic footwear improved spatiotemporal parameters (higher gait speed, longer step length, shorter step time and smaller step width) during walking. Moreover, less sagittal shank-footwear range of motion, more frontal shank-footwear range of motion, more dorsiflexion of the footwear-to-horizontal angle at initial contact and more hip adduction during the stance phase were found.

Interpretation: Orthopedic footwear improved walking in individuals with Hereditary Motor Sensory Neuropathy, whereas it did not affect postural stability during quiet standing or dynamic balance. Especially gait speed and spatiotemporal parameters improved. An improved heel landing at initial contact for all footwear and reduced foot drop during swing for mid and high orthopedic footwear contributed to the gait improvements wearing orthopedic footwear.

1. Introduction

Hereditary Motor Sensory Neuropathy (HMSN) is the most common inherited neuromuscular disorder (prevalence 1:2500 people) (Murphy et al., 2012). HMSN is characterized by bilateral distal muscle weakness and sensory impairments, starting in the feet and lower legs (Sabir and Lyttle, 1984). Due to weakness of the foot muscles, foot deformities develop such as pes cavus and claw toes (Tazir et al., 2014). These foot deformities cause secondary abnormalities at the level of the hindfeet (varus), knees (hyperextension), and hips/pelvis (anterior tilt). The walking pattern associated with HMSN is characterized by foot drop, impaired push-off, and overloading of the lateral foot edge during roll-off (Vinci and Perelli, 2002). In addition, compensatory kinematic adjustments at the knee and hip joints are observed (Don et al., 2007a; Guzian et al., 2006; Newman et al., 2007; Nonnekes et al., 2021; Sabir and Lyttle, 1984). Overall, people with HMSN show a relatively low gait

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speed (Newman et al., 2007), shortened step length (Don et al., 2007b) and enlarged step width (Don et al., 2007b; Nonnekes et al., 2021) compared to persons without impairments. In addition, postural instability during standing (de França Costa et al., 2018; Lencioni et al., 2014; van der Linden et al., 2010) and walking (de França Costa et al., 2018; Don et al., 2007b; Newman et al., 2007; Sabir and Lyttle, 1984), and increased risk of falling (Eichinger et al., 2016; Ramdharry et al., 2011) have been reported.

To improve postural stability during standing and walking, orthopedic footwear is commonly prescribed to people with HMSN (Postema et al., 2018, 1991). Low orthopedic footwear aims to enable plantigrade foot loading during standing and walking by compensating for structural foot deformities, while high orthopedic footwear (with integrated orthotic support) may additionally compensate for weakness of the lower leg muscles during walking. However, formal evidence for the efficacy of orthopedic footwear in people with HMSN is hardly available. Only two studies investigated the effect of orthopedic footwear on standing and walking in individuals with HMSN (Geurts et al., 1992; Guzian et al., 2006). One study (n = 10) investigated orthopedic footwear during quiet standing. With orthopedic footwear, people with HMSN showed a tendency towards lower center of pressure (CoP) velocities characterized by a marked reduction of sway amplitude in the frontal plane, which coincided with a higher sway frequency in this plane (Geurts et al., 1992). The second study, a case study, found that orthopedic footwear enhanced gait speed, cadence and step length (Guzian et al., 2006). Furthermore, they reported better postural stability and no falls when the subject wore orthopedic footwear.

Generally, gait capacity can be categorized in three components to understand functional walking: (1) stepping, (2) postural stability, and (3) gait adaptability (Balasubramanian et al., 2014). Stepping is defined as moving forward with a cyclical pattern of limb and trunk movements, usually quantified by spatiotemporal parameters and by joint kinematics and kinetics. While moving forward, the body must maintain postural stability to keep the center of mass (CoM) within the changing base of support, taking into account inertial forces, usually quantified by dynamic balance measures describing the CoM relative to the base of support or CoP. In daily life, the stepping pattern and basic postural stability during walking must also be adjustable to changing environmental demands, which is commonly referred to as gait adaptability.

This present study is focused on the effects of orthopedic footwear on postural stability during quiet standing (static balance) as well as on stepping and postural stability during walking (dynamic balance) in individuals with HMSN. To this end, we compared measures of quiet standing balance as well as spatiotemporal parameters, kinematics, kinetics, and dynamic balance during walking when subjects wore their own customized orthopedic footwear with wearing minimal supportive, flexible footwear. We hypothesized a lower CoP velocity during quiet standing and higher gait speed during walking with the orthopedic footwear compared to minimal supportive, flexible footwear.

2. Method

2.1. Participants

Individuals with HMSN who visited the Sint Maartenskliniek between January 2017 and March 2018 were screened for eligibility by a rehabilitation physician. Inclusion criteria were: 1) diagnosed with HMSN, 2) between 18 and 70 years old, and 3) provided with customized orthopedic footwear for a minimum of two months to improve postural stability and/or to prevent falling. Exclusion criteria were: 1) inability to walk independently for 2 min without assistance, 2) pain and/or pressure sores related to the orthopedic footwear, 3) surgery of the lower extremities less than one year ago, and 4) other disorders influencing the gait pattern. The following demographic characteristics were registered upon inclusion: age, sex, height, and weight. In addition, clinical characteristics, like HMSN disease type and Medical Research Council (MRC) Scale scores (Avers and Brown, 2018) for muscle strength of the ankle dorsal- and plantar flexors were extracted from the medical records.

All participants gave written informed consent in accordance with the Declaration of Helsinki. The study was approved by the internal review board of the Sint Maartenskliniek and the regional medical ethics committee of Arnhem-Nijmegen (2018-4306).

2.2. Footwear

Orthopedic footwear (Fig. 1A) was custom made for each individual and molded to the individual's foot shape. The insole, an internal footwear feature, especially accommodates the foot deformity to relieve pain and pressure and to assist a neutral position of the hind foot. The aim is to accept plantar flexion of the first metatarsal (deepening of MT1) (Louwerens, 2018) by lowering the insole under MT1, without changing the position of the ankle joint (no increase in ankle plantar flexion). External footwear features like shaft height, heel adjustment en forefoot apex position, are based on the individual's characteristics, e.g. muscle strength and walking pattern, and treatment purpose. Common footwear features include shaft height, heel adjustment and forefoot apex position. Shaft height was defined as the height of the supplement in the shaft in relation to the ankle joint. Low orthopedic footwear consists of a shaft height below the level of the ankle, whereas the shaft height of mid and high orthopedic footwear is above the level of the ankle in order to control the movement of the ankle joint in the frontal plane. Adjustments to the heel can be made by rounding off the posterior edge (beveled heel) to decrease ankle dorsiflexion work in loading response or by adding a lateral flare to the heel (flared heel) to increase stability in the frontal plane (Daryabor et al., 2016). The forefoot rocker can be influenced by the position of the apex (forefoot apex position) (Preece et al., 2017), which is the position where the outsole begins to curve upwards under the forefoot. A neutral apex position is at the MTP joints, whereas the apex position could also be placed more proximal or distal to achieve an early or delayed forefoot rocker, respectively. Standardized footwear consisted of a minimal supportive sneaker with a flexible shaft made of canvas and a flat rubber sole without heel-to-toe drop (Fig. 1B).

2.3. Assessments

Participants visited the research department of the Sint Maartenskliniek once. Prior to the balance and gait measurements, the American Orthopedic Foot and Ankle Society (AOFAS) Ankle-Hindfoot Scale (Kakwani and Siddique, 2014) score and the classification of foot deformity proposed by Louwerens (Louwerens, 2018) were determined by the primary researcher (LdJ). This researcher also assessed footwear features of the customized orthopedic footwear and its intensity of use. The three footwear features included shaft height, heel adjustment and forefoot apex positioned, and were categorized in three levels. Shaft height was scored as follows: 'low height' below the ankle joint, 'mid height' max 10 cm above the ankle joint, and 'high height' $>$ 10 cm above ankle joint (Fig. 1C). Heel adjustments were categorized in: 'no adjustment', in which the posterior edge was perpendicular to the ground, 'beveled heel', in which the posterior edge of the heel was rounded off or 'flared heel', in which the heel was extended with a lateral flare (Fig. 1D). Forefoot apex position was classified as: 'neutral' in which is the apex position coincides with the MTP joints, 'proximal', in which the apex position is closer to the heel or 'distal', in which the apex position is closer to the toes (Fig. 1E).

Subsequently, participants were instrumented with 20 markers according to the Plug-in Gait lower body model (Plug-in-Gait, Vicon Motion Systems Ltd., Oxford, UK). The foot markers were placed on the footwear. Balance measurements were performed on a platform with integrated force plate (AMTI, Watertown, MA, USA) collecting force data at a sampling rate of 500 Hz. Gait measurements were performed

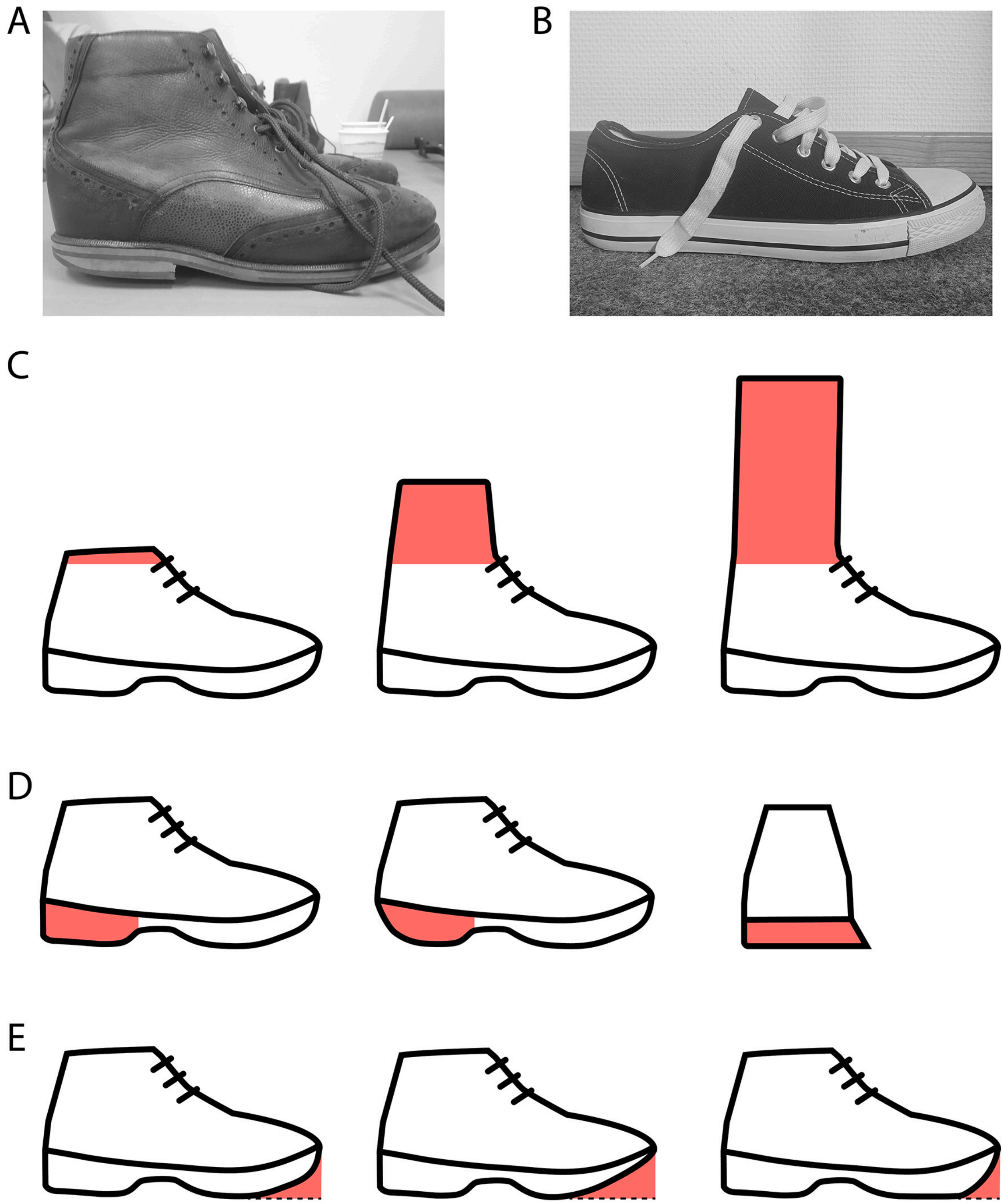

Fig. 1. Overview of the footwear. A. Example of orthopedic footwear. B. Standardized footwear. C. Shaft height: low height, mid height and high height. D. Heel adjustments: no adjustment, beveled heel and flared heel (posterior view) E. Forefoot apex position: neutral, proximal, distal.

on an instrumented treadmill, the Gait Real-time Analysis Interactive Lab (GRAIL, Motek Medical BV, the Netherlands). Marker position was captured by a ten-camera motion capture system (Vicon, Oxford, UK) at a sample frequency of 100 Hz. Force data were collected with two force plates embedded underneath the treadmill belt and sampled at 1000 Hz.

Participants first performed three practice trials to familiarize themselves with walking on the GRAIL. Thereafter, participants completed two tasks in the following, fixed order: 1) quiet standing task and 2) 2-min walk test (2MWT). These two tasks were performed with customized orthopedic footwear that participants brought to the assessment, and with the standardized footwear, that was provided on site. Participants were fitted into the standardized footwear without any additional modifications. After completion of both tasks with one type of footwear, participants changed to the other type of footwear. The order of measurements was randomized across participants.

2.3.1. Quiet standing

During the quiet standing task, participants stood upright on the force platform with their feet against a foot frame (medial sides of the heels 8.4 cm apart; each foot out-toeing at a 9° angle from the sagittal midline) (De Haart et al., 2004). Participants were instructed to stand as still as possible for 30 s either with open or closed eyes. Both conditions were performed twice, starting with eyes open followed by eyes closed.

2.3.2. 2-Minute Walk Test

The 2MWT was performed once per footwear condition on the GRAIL in a self-paced mode, i.e. the treadmill speed was automatically controlled by continuously comparing the position of the pelvis to the midline of the treadmill (Sloot et al., 2014). Walking forward or backward relative to the midline resulted in treadmill acceleration or deceleration, respectively. Participants were instructed to walk as far as possible in two minutes (Laboratories, 2002).

2.4. Data analysis and outcome measures

For quiet standing, signals were processed using a custom-made program after a 16-bit AD-conversion (Anker et al., 2008; Geurts et al., 1993). CoP during standing (CoP_{st}) was calculated as the point of application of the resultant of the ground reaction forces in the anterior-posterior (AP) and medial-lateral (ML) directions, separately. The CoP data was low-passed filtered (cut-off frequency 6 Hz). Firstly, the root mean square (RMS) amplitude of the CoP displacement ($aCoP_{st}$) [mm] in both AP and ML directions was calculated. Then, after a first-order differentiation, the RMS velocity of the CoP ($vCoP_{st}$) [mm/s] in either direction was calculated as the primary outcome measure. The mean CoP frequency ($fCoP_{st}$) in each direction was determined as the ratio between $aCoP_{st}$ and $vCoP_{st}$, using the following equation: $fCoP_{st} = vCoP_{st} / (aCoP_{st} \times \sqrt{2} \times 4)$ (Geurts et al., 1992).

Marker data of the 2MWT were filtered using the Woltring cross-validity quintic spline routine (MSE = 20) before running the Vicon Plug-In-Gait model and software (Woltring, 1986). Thereafter, marker and model data was filtered using a zero lag, fourth-order low-pass Butterworth filter (cut-off frequency 20 Hz). Instants of heel strike and toe-off were identified using the markers on both feet as described by Zeni et al. (Zeni et al., 2008).

The primary outcome measure for walking was gait speed. Secondary outcome measures included spatiotemporal parameters, joint kinematics and kinetics, and dynamic balance measures. The first 20 s of the 2MWT were excluded from all analyses to remove the starting phase of walking. All outcome measures were averaged over steps between the 20th and 120th second of the 2WMT and calculated for each leg separately. The most affected leg, based on MRC of the dorsal- and plantar flexors, was used for analysis. If no differences were present, the leg was randomly selected.

Gait speed was defined as the mean treadmill speed [m/s]. Step length [cm] and step width [cm] were determined for each step and defined as the AP and ML distance between the heel markers at heel strike, respectively. Step time [s] was defined as the mean time between a heel strike on one side to the subsequent heel strike of the contralateral foot. The standard deviation over all steps was used to calculate the variability of the step length, step time and step width.

Due to placement of the markers on the footwear, not the ankle angle inside the footwear but the angle of the shank relative to the footwear was measured. Kinematics (angles) and kinetics (internal moment and power) between the shank and footwear, and of the knee and hip joints in the sagittal and frontal planes were calculated per gait cycle using the Vicon Plug-In-Gait model and software. Furthermore, the shank-to-vertical angle and the footwear-to-horizontal angle were calculated per gait cycle. The shank-to-vertical angle was defined as the angle between the knee and shank-footwear joint center, and the vertical in the sagittal plane (Owen, 2010), whereas the footwear-to-horizontal was defined as the angle between the toe and heel marker, and the horizontal in the sagittal plane (Owen et al., 2018). The shank-to-vertical angle at midstance and the footwear-to-horizontal at heel strike were determined for each gait cycle. Kinetic data were excluded from analysis when the foot had hit both force plates during the stance phase. Range of motion (RoM) [deg] was calculated as the maximal minus the minimum joint angle during one gait cycle. Peak moment [Nm/kg] and peak power [W/kg] were defined as the maximum joint moment and power during the stance phase, respectively. Propulsive force was estimated by the propulsive impulse [N/s/kg], which was calculated as the time integral of the positive anterior ground reaction force during the stance phase (Bowden et al., 2006).

Dynamic balance assessment was based on the relation between the CoM or extrapolated center of mass (XCoM) and base of support or the CoP during walking (CoP_w). The CoM was estimated using the average of the four pelvis markers (Whittle, 1997). The XCoM was calculated using the equation proposed by Hof (Hof et al., 2005). The CoP_w was calculated using force plate data that was filtered using a zero lag, fourth-order low-pass Butterworth filter (cut-off frequency 20 Hz). A continuous CoP_w signal was obtained by the weighted average of the CoP_w values derived from both force plates (Sloot et al., 2015).

As a measure of dynamic balance, we calculated the $XCoM\text{-}CoP_{ML}$ [cm], which is the shortest distance between the XCoM and CoP at the instant of heel strike in the ML direction (Lugade et al., 2011). Lower values indicate better postural stability during walking. The $XCoM\text{-}CoP_{ML}$ was reported to be reliable in previous studies (de Jong et al., 2020).

2.5. Statistical analysis

The COP outcomes of the two quiet standing task performances in the same condition (footwear, vision) were averaged into a mean value per condition per subject. Then, a two-way repeated-measures ANOVA ($\alpha = 0.05$) was performed to determine the effects of footwear (orthopedic vs. standardized) and vision (eyes open vs. eyes closed) on the CoP measures. To evaluate walking, paired *t*-tests ($\alpha = 0.05$) were used to determine the effects of footwear on the group means for gait speed, spatiotemporal parameters, joint kinematics and kinetics, and dynamic balance. Statistical Parametric Mapping (SPM) was performed to assess where in the gait cycle differences between footwear were present for the joint kinematics and kinetics (Pataky et al., 2013).

3. Results

3.1. Participants

Demographics, clinical characteristics, MRC scores and orthopedic

Table 1
Participants' demographic and clinical characteristics ($n = 15$).

Characteristics	Mean ± SD	Frequency
Age [years]	49.6 ± 14.8	
Sex, male/female		10/5
Height [cm]	179.4 ± 9.8	
Weight [kg]	82.0 ± 17.9	
HMSN disease type, 1/2/4		9/5/1
AOFAS Ankle-Hindfoot Scale score	78 ± 14	
MRC score ankle plantar flexors, 0/1/2/3/4/5*		1/2/7/0/3/1
MRC score ankle dorsal flexors, 0/1/2/3/4/5*		3/1/4/4/1/2
Orthopedic footwear features		
Shaft height, low/mid/high		4/9/2
Heel adjustment, no/beveled/flared		2/11/2
Forefoot apex position, normal/proximal/distal		6/9/0
Use of orthopedic footwear		
Days per week, 0/1/2–3/4–5/6–7		0/0/0/1/14
Hours per day, <1/1–4/4–8/8–12/>12		0/0/0/8/7

HMSN: Hereditary Motor Sensory Neuropathy, AOFAS: American Orthopedic Foot and Ankle Society, MRC: Medical Research Council.
* $n = 14$.

footwear features of the 15 participants enrolled in this study are displayed in Table 1. For one patient, no MRC scores were available. Due to missing markers during the 2MWT, kinematic and kinetic parameters could not be calculated for three other participants, whom were therefore excluded from the kinematic and kinetic analyses.

3.2. Quiet standing

No main or interaction effect of footwear was found for any outcome measure during quiet standing (Table 2). Higher $vCoP_{st}$, $aCoP_{st}$, and $fCoP_{st}$ values in both the AP and ML directions were found during eyes closed compared to eyes open, irrespective of footwear.

3.3. 2-Minute Walk Test

Orthopedic footwear significantly improved gait speed ($t(14) = 4.1$, $P = 0.001$), step length ($t(14) = 4.0$, $P = 0.001$), step time ($t(14) = -2.9$, $P = 0.01$) and step width ($t(14) = -4.2$, $P = 0.001$) compared to standardized footwear (Table 3).

Fig. 2 shows the kinematics and kinetics of the shank-footwear, knee and hip joints in the sagittal (Fig. 2A) and frontal (Fig. 2B) planes during the gait cycle. SPM revealed that the hip showed lower extension

Table 2
Outcomes of the quiet standing task.

	Eyes open			Eyes closed			Shoes		Task		Shoes ×Task	
	Orthopedic footwear	Standardized footwear	Mean difference	Orthopedic footwear	Standardized footwear	Mean difference	F	*P*	F	*P*	F	*P*
$aCoP_{st}$ AP [mm]	4.8 ± 1.7	5.4 ± 2.6	−0.5 ± 2.0	9.3 ± 4.6	9.5 ± 3.8	−0.3 ± 3.3	0.57	0.46	41	**<0.001**	0.09	0.16
$aCoP_{st}$ ML [mm]	3.1 ± 1.7	4.1 ± 2.3	−1.0 ± 1.6	5.2 ± 4.0	6.9 ± 3.8	−1.7 ± 4.0	4.3	0.06	27	**<0.001**	0.7	0.42
$vCoP_{st}$ AP [mm/s]	14.7 ± 4.5	17.1 ± 8.5	−2.4 ± 9.0	36.5 ± 19.2	36.9 ± 18.3	−0.4 ± 20.9	0.01	0.92	33	**<0.001**	2.3	0.16
$vCoP_{st}$ ML [mm/s]	8.1 ± 4.5	10.8 ± 6.8	−2.6 ± 7.2	16.3 ± 13.9	24.6 ± 18.7	−8.3 ± 19.3	2.8	0.12	16	**0.001**	2.2	0.15
$fCoP_{st}$ AP [Hz]	0.56 ± 0.12	0.58 ± 0.16	−0.03 ± 0.16	0.72 ± 0.21	0.69 ± 0.19	0.02 ± 0.19	0.00	0.99	15	**0.002**	1.1	0.32
$fCoP_{st}$ ML [Hz]	0.49 ± 0.15	0.48 ± 0.14	0.02 ± 0.14	0.58 ± 0.17	0.57 ± 0.17	0.02 ± 0.24	0.04	0.85	11	**0.005**	0.00	0.97

CoP: center of pressure, aCoP: RMS CoP amplitude, vCoP: RMS CoP velocity, fCoP: mean CoP frequency, AP: anterior-posterior, ML: medial-lateral.
Bold: significant difference between footwear conditions ($P < 0.05$).

Table 3
Spatiotemporal outcomes of the 2MWT.

Outcome	Orthopedic footwear	Standardized footwear	Mean difference	*P*
Gait speed [m/s]	1.33 ± 0.28	1.10 ± 0.35	0.23 ± 0.22	**0.001**
Step length [cm]	66 ± 13	58 ± 17	8.5 ± 8.2	**0.001**
Step time [s]	0.49 ± 0.04	0.52 ± 0.05	−0.03 ± 0.04	**0.01**
Step width [cm]	12 ± 3.9	15 ± 4.4	−2.8 ± 2.6	**0.001**
Step-length variability [cm]	31 ± 9.4	38 ± 16	−7 ± 14	0.08
Step-time variability [s]	0.01 ± 0.01	0.03 ± 0.02	−0.01 ± 0.02	0.10
Step-width variability [cm]	33 ± 7.7	36 ± 11	−2.4 ± 7.4	0.23

Bold: significant difference between footwear conditions ($P < 0.05$).

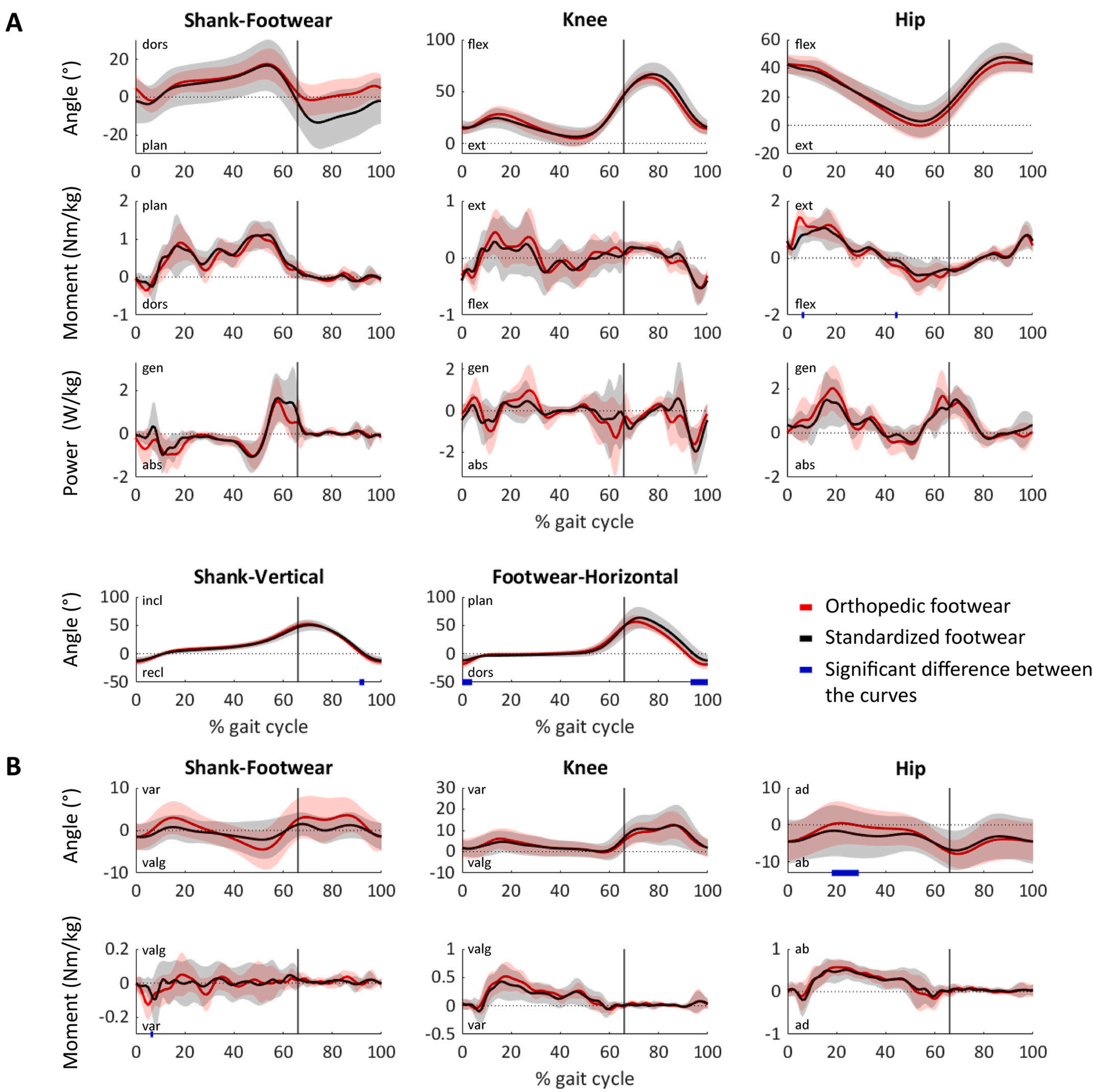

Fig. 2. Kinematics (angles) and kinetics (internal joint moment and power) of the 2MWT in the sagittal (A) and frontal (B) planes for orthopedic footwear (red line) and standardized footwear (black line). Lines represent the mean values and shaded areas the standard deviations. Blue horizontal bars on the X-axis indicate differences between the curves. dors: dorsiflexion, plan: plantar flexion, flex: flexion, ext.: extension, gen: generation, abs: absorption, incl: inclination, recl: reclination, var.: varus, valg: valgus, ad: adduction, ab: abduction. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

moments during loading response (6–7%) and terminal stance (44%) with orthopedic compared to standardized footwear ($P = 0.002$ and P = 0.01, respectively). The shank-to-vertical angle was more reclined during terminal swing (91–93%) with orthopedic footwear ($P = 0.04$). During initial contact (1–4%) and terminal swing (93–100%), the footwear-to-horizontal was more in dorsiflexion with orthopedic footwear ($P = 0.02$ and P = 0.02, respectively). In the frontal plane, the hip angle showed more adduction during midstance (18–29%) with orthopedic footwear (P = 0.01). The shank-footwear showed higher varus moments during loading response (6%, P = 0.02) with orthopedic footwear.

Across the gait cycle, orthopedic footwear showed a decrease in shank-footwear RoM in the sagittal plane ($t(11) = -2.8$, $P = 0.02$) and an increase in shank-footwear RoM in the frontal plane ($t(11) = 3.4$, $P = 0.006$). Lower shank-footwear peak power was found for walking with orthopedic footwear ($t(11) = -2.8$, $P = 0.02$). The footwear-to-

Table 4
Kinematic and kinetic outcomes of the 2MWT.

Outcome		Orthopedic footwear	Standardized footwear	Mean difference	*P*
Sagittal plane					
Propulsive impulse [N/s/kg]		0.25 ± 0.06	0.22 ± 0.1	0.03 ± 0.07	0.16
Ankle	RoM [deg]	24 ± 8.2	33 ± 13	−9.3 ± 11.6	**0.02**
	Peak moment [Nm/kg]	1.6 ± 0.5	1.7 ± 0.8	−0.09 ± 0.5	0.58
	Peak power [W/kg]	2.6 ± 1.2	3.8 ± 2.0	−1.2 ± 1.5	**0.02**
Knee	RoM [deg]	62 ± 6.6	64 ± 7.3	−1.7 ± 5.9	0.34
	Peak moment [Nm/kg]	0.9 ± 0.4	0.8 ± 0.4	0.07 ± 0.2	0.34
	Peak power [W/kg]	2.6 ± 1.0	2.6 ± 1.5	0.04 ± 1.3	0.92
Hip	RoM [deg]	48 ± 4.7	48 ± 8.7	0.16 ± 6.9	0.94
	Peak moment [Nm/kg]	1.8 ± 0.4	1.7 ± 0.6	0.19 ± 0.6	0.29
	Peak power [W/kg]	3.2 ± 1.1	2.8 ± 1.2	0.39 ± 1.0	0.20
Shank-to-vertical angle [deg]		10 ± 3.5	9.4 ± 4.0	0.92 ± 2.4	0.21
Foot-to-horizontal angle [deg]		−18 ± 7.1	−13 ± 9.1	−5.4 ± 4.2	**0.001**
Frontal plane					
Ankle	RoM [deg]	11 ± 3.5	7.4 ± 3.3	2.8 ± 2.4	**0.006**
	Peak moment [Nm/kg]	0.21 ± 0.1	0.22 ± 0.1	0.02 ± 0.1	0.74
Knee	RoM [deg]	18 ± 3.6	19 ± 4.6	−1.2 ± 2.3	0.37
	Peak moment [Nm/kg]	0.66 ± 0.2	0.63 ± 0.3	0.08 ± 0.1	0.55
Hip	RoM [deg]	12 ± 3.7	11 ± 2.8	0.72 ± 2.5	0.37
	Peak moment [Nm/kg]	0.79 ± 0.1	0.76 ± 0.2	0.04 ± 0.1	0.50

RoM: range of motion.
Bold: significant difference between footwear ($P < 0.05$).

horizontal angle at heel strike was more in dorsiflexion with orthopedic footwear compared to standardized footwear ($t(11) = -4.5$, $P = 0.001$). No significant differences between footwear conditions were found for any other kinematic or kinetic outcome measure (Table 4).

The XCoM-CoP_{ML} showed no significant difference between orthopedic footwear (11.6 ± 2.3) and standardized footwear (12.2 ± 2.1; $t(14) = -1.7$, $P = 0.11$).

4. Discussion

Orthopedic footwear improved stepping, whereas it did not affect postural stability during quiet standing (static balance) or walking (dynamic balance) in individuals with HMSN. More specifically, with orthopedic footwear, walking improved in terms of gait speed and spatiotemporal parameters (increased step length and decreased step time and step width). Interestingly, reduced step width may still indicate some improvement in frontal plane postural stability. Generally, there was a limited impact on gait kinematics or kinetics, but we observed a decrease in sagittal and an increase in frontal shank-footwear RoM, a decreased sagittal shank-footwear peak power, more dorsiflexion of the footwear-to-horizontal angle at initial contact and more hip adduction during the stance phase. Improved heel landing at initial contact for all orthopedic footwear and reduced foot drop during swing for mid and high orthopedic footwear seemed to be the main contributors to gait improvement when wearing orthopedic footwear.

Compared to healthy controls, individuals with HMSN showed larger values of RMS CoP amplitude and velocity (Geurts et al., 1993). In line with a previous study, no significant effect of footwear on CoP velocity during quiet standing was found (Geurts et al., 1992), neither with eyes open nor with eyes closed. In contrast with this previous study, we did not find a trade-off between CoP amplitude and frequency in the ML direction, which is probably attributable to the fact that the control condition in the previous study was barefoot instead of using standardized footwear. If a positive effect of footwear was present, it could have been eliminated by the proximal forefoot apex position of most orthopedic footwear, which decreased the base of support and might negatively affect postural stability. As CoP velocity is closely related to the velocity and acceleration of the body's CoM during quiet standing, it seems safe to conclude that orthopedic footwear in people with HMSN has no beneficial effect on static balance compared to standardized footwear.

Individuals with HMSN walked slower on both footwear conditions compared to healthy controls with the same age (de Jong et al., 2020). Congruent with our hypothesis and in line with the case study by Guzian et al. (Guzian et al., 2006), gait speed and spatiotemporal parameters improved when our subjects with HMSN were walking with orthopedic footwear compared to standardized footwear. Moreover, thirteen out of the fifteen participants showed an increase in gait speed exceeding the minimal clinically important difference (MCID) of 0.10 m/s (Bohannon and Glenney, 2014). The increase in gait speed was due to 10% increase in step length and 6% increase in cadence. Remarkably, the increase in step length was not reflected by an increase in propulsive force nor in sagittal shank-footwear power. Moreover, the sagittal shank-footwear peak power was reduced in orthopedic footwear. Instead, orthopedic footwear decreased the sagittal shank-footwear RoM during the gait cycle compared to standardized footwear, which was caused by reduced plantar flexion during the end of the swing phase and at initial contact (Fig. 2). This reduced plantar flexion was also represented by an increased dorsiflexion of the footwear-to-horizontal angle during terminal swing and at initial contact. No clear differences were found in knee or hip kinematics or kinetics in the sagittal plane, nor in the shank-to-vertical angle. Hence, the main effect of orthopedic footwear may be that it enables individuals to load the foot properly (Nonnekes et al., 2019) due to decreased foot drop during initial contact for all orthopedic footwear and during swing for mid and high orthopedic footwear. As a consequence, subjects were able to walk with a heel strike instead of a mid- or forefoot landing, which may have resulted in a more efficient first rocker. This heel contact and improved efficiency of the first rocker has most likely contributed to an increase in both step length and cadence, leading to a higher walking speed.

In addition to improvements in the sagittal plane, orthopedic footwear also caused changes in the frontal plane. Participants wearing orthopedic footwear walked with a 20% (3 cm) smaller step width compared to standardized footwear, suggesting improved postural stability in the frontal plane. Yet, the smaller step width did not induce a significant decrease in the distance between the XCoM and CoP at heel strike. Movement of the shank-footwear and hip in the frontal plane was increased when wearing orthopedic footwear, which is probably an epiphenomenon of the smaller step width and the longer step length. When the swing leg is placed closer to the line of progression, the hip is more adducted during midstance. The increased shank-footwear RoM in the frontal plane should be interpreted with caution due to the limitations of the used marker model. The marker model treats the foot as a rigid model, only registering movement of the foot relative to the shank, which includes varus/valgus and foot deformities in the same curve.

A few limitations of this study need to be addressed. Since the markers were placed on the footwear, we did not measure the ankle angle inside the footwear. However, the heel-to-toe drop at the lateral side of the footwear was near zero in both footwear types. This was supported by an almost similar shank-to-vertical angle during the whole gait cycle in both footwear. Therefore, we expect that the ankle kinematics and kinetics will be almost similar to the shank-footwear kinematics and kinetics. Furthermore, the offset ankle angle inside the footwear (i.e. maximum 0.5 cm heel-to-toe-drop results in maximum 2 degrees) is within the measurement error of sagittal joint angles (McGinley et al., 2009). Another limitation is that participants were not used to walk on the standardized footwear. However, practice trials were performed to familiarize themselves with walking with standardized footwear. The standardized footwear were flat flexible sneakers without any support function minimizing the influence on the walking pattern.

Although orthopedic footwear is commonly prescribed to individuals with balance and gait problems due to HMSN, this is the first study to support its beneficial effects on the gait pattern in a larger group of affected individuals. Unfortunately, our sample size does not allow relating the effects of individual orthopedic footwear features to specific kinematic and kinetic gait characteristics. Therefore, for future research, it is important to relate individual footwear features to specific gait characteristics in people with HMSN using larger sample sizes. Furthermore, other balance and stability measures, like foot placement strategy (Vlutters et al., 2016) or reactive balance control (McAndrew Young et al., 2012), could be investigated to assess other dimensions of the gait capacity.

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CRediT authorship contribution statement

Lysanne A.F. de Jong: Conceptualization, Methodology, Software, Validation, Formal analysis, Investigation, Data curation, Writing – original draft, Writing – review & editing, Visualization, Project administration, Funding acquisition. **Yvette L. Kerkum:** Conceptualization, Methodology, Writing – review & editing, Supervision, Funding acquisition. **Viola C. Altmann:** Resources, Writing – review & editing. **Alexander C.H. Geurts:** Supervision, Writing – review & editing. **Noel L.W. Keijsers:** Conceptualization, Methodology, Software, Formal analysis, Writing – review & editing, Supervision, Project administration, Funding acquisition.

Declaration of Competing Interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests: This study is part of the GaReC project, which is co-funded by OIM Orthopedie and the PPP Allowance made available by Health ~ Holland, Top Sector Life Sciences & Health, to stimulate public-private partnerships. Lysanne de Jong is employed by OIM Orthopedie. Neither OIM Orthopedie, nor Lysanne de Jong have (financial) benefits related to this project. There are no other conflicts of interest associated with this study.

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