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Published in:
Journal of Electromyography and Kinesiology

DOI (link to publication from Publisher):
10.1016/j.jelekin.2020.102441

Publication date:
2020

Document Version
Accepted author manuscript, peer reviewed version

New assistive walker improved local dynamic stability in young healthy adults

Bo Eitel Seiferheld, Jeppe Frost, Christian Andersen, Afshin Samani

Department of Health Science and Technology, Sport Sciences, Aalborg University, Aalborg, Denmark

To be submitted as an original paper for Journal of Electromyography and Kinesiology

Corresponding author:

Afshin Samani, PhD

Associate Professor

Sport Sciences - Performance and Technology, Department of Health Science and Technology, Aalborg University

Niels Jernes Vej 12 A5-213, DK-9220 Aalborg, Denmark

Phone: (+45) 99 40 24 11

E-mail: afsamani@hst.aau.dk

Keywords: Rehabilitation; Gait; Lyapunov exponent

Conflict of interest statement: None

Financial Disclosure statement: None

Abstract

In this study, we investigated the effect of walker type on gait pattern characteristics comparing normal gait (NG), gait with a regular walker (RW), and gait with a newly developed walker with vertical moveable handlebars, the Crosswalker (CW).

Partial weight bearing (PWB) of the feet, peak joint angles and largest Lyapunov exponent (λ_{max}) of the lower extremities (hip, knee, ankle) in the sagittal plane, and gait parameters (gait velocity, stride length, cadence, stride duration) were determined for 18 healthy young adults performing 10 walking trials for each walking condition. Assistive gait with the CW improved local dynamic stability in the lower extremities (hip, knee, ankle) compared with RW and was not significantly different from NG. However, peak joint angles and stride characteristics in CW were different from NG. The PWB on the feet was lower with the RW (70.3%) compared to NG (82.8%) and CW (80.9%). This improved stability may be beneficial for the elderly and patients with impaired gait. However, increased PWB is not beneficial for patients during the early stages of rehabilitation.

1. Introduction

Walkers, crutches and walking sticks have been developed to increase mobility and stability for the elderly and people with various clinical conditions (Bateni and Maki, 2005; Martins et al., 2015). Especially, the application of these devices may improve gait stability for people with deteriorating balance or post-surgery balance impairments (Martins et al., 2015; Youdas et al., 2005). Furthermore, these devices alleviate pain intensity in the lower extremities by letting the upper body carry more of the bodyweight, which is crucial post injury and surgery to facilitate the recovery process (Bohannon, 1997; Liu et al., 2009; Martins et al., 2015; Youdas et al., 2005).

Nowadays, various assisting devices are specifically designed for rehabilitating gait mal-functionalities but our knowledge about their interaction with the gait is limited. For example, in a review, Bohannon (1997) pointed out that wheeled walkers might have better utilization than standard framed walkers since elderly patients and healthy elderly people ambulated faster through an obstacle course with wheeled walkers and they preferred the wheeled walker over the standard framed walker.

Restriction of arm motion during normal gait increases metabolic cost of stability since the angular momentum of the swinging leg is not balanced out by the contralateral movement of the swinging arm, thus more muscle activity is recruited to stabilise the trunk (Collins et al., 2009). Furthermore, arm restriction may compromise the overall gait stability (Bruijn et al., 2010; Collins et al., 2009; Punt et al., 2015; Wu et al., 2016). It is conceivable that the restriction of arm movement in regular wheeled walkers, where the handlebars are locked horizontally, also involves similar disadvantages. Recently, the Crosswalker[1] as a new concept of a walker, has been developed and equipped with vertical moveable handlebars facilitating activity in the upper body. The handlebars are positioned about the elbow height and move in an alternating fashion as the user operates the walker, (Figure 1). At first glance, the gait with Crosswalker looks more similar to the normal gait

[1] www.crosswalker.dk

compared with a regular walker, thus, it is reasonable to assume that arm motion with the Crosswalker may enhance gait stability (Bruijn et al., 2010; Collins et al., 2009; Punt et al., 2015; Wu et al., 2016) and increase walking speed (Bohannon, 1997; Liu et al., 2009; Youdas et al., 2005). On the other hand, due to dynamic movement and placement of the arms, the weight bearing on the Crosswalker may be reduced and that, in turn, may cause more pain on the lower extremities (Bohannon, 1997; Liu et al., 2009; Youdas et al., 2005). Since maintaining stability during gait is fundamental in everyday life (Li et al., 2003), it is important to justify that the Crosswalker, actually provide increased gait stability. Local gait dynamic instability is commonly measured by the largest Lyapunov exponent (λ_{max}) (Mehdizadeh, 2018; Stergiou and Decker, 2011). Here, the variation observed across several repetitions of the system trajectories in the state-space allows the quantification of how the system responds to small divergences of system trajectories (Dingwell and Cusumano, 2000; Mehdizadeh, 2018; Stergiou and Decker, 2011). Such a method has been extensively used to investigate the alteration in gait stability due to aging and disorders (Bruijn et al., 2009; Dingwell and Cusumano, 2000; Granata and Lockhart, 2008; Terrier and Reynard, 2015). Individual factors, such as, muscular fitness, physical activity and peripheral sensation, are known to influence gait stability (Hamacher et al., 2019) and markedly deteriorate with age (Decorps et al., 2014; Tieland et al., 2018), thus it is important to investigate the potential effect of the Crosswalker in a healthy population to rule out any possible effect modifiers that could mask the sole effect of the Crosswalker.

Therefore, the purpose of this study was to compare the gait local dynamic stability during normal gait (NG), gait with a regular walker (RW), and gait with the Crosswalker (CW). To the best of our knowledge, this is the first study examining the CW for its potential advantages in assisting the gait stability. We hypothesized that gait local dynamic stability with CW was more stable than gait with RW, albeit not more stable than NG. Secondly, the load on the lower extremities, expressed as the partial weight bearing (PWB) of the feet, with the CW would be greater than that of RW but lower than that of NG. Thirdly, that the gait parameters, gait velocity, stride length, cadence and stride duration with the CW would be closer to NG compared with RW.

2. Methods

2.1. Participants.

Eighteen healthy adults (13 men, 5 women, 183.5±8.1 m, 80.6±13.0 kg, 24.5±3.1 years) volunteered to participate in the present study. None of the participants had any history of injuries, balance impairments, neural disorders or diseases. All participants received both written and oral information about the overall procedure and aim of the study and provided written informed consent before participating. The experimental data recordings conformed to the standard ethical guidelines of The Scientific Ethical Committee for the Region of Northern Jutland.

2.2. Experimental protocol.

Gait of the participants was investigated in three different conditions; NG, RW and CW, (Figure 1), respectively. The order of performing the task in the three conditions was randomly counterbalanced following a 3x3 Latin square to minimize the risk of carry-over effects. Before each condition participants carried out a 5-minute acclimation to minimize learning effects and to find their preferred gait velocity with the different walkers. Each participant was instructed to perform 10 trials per condition and each trial consisted of 30 meters walking straight ahead. The participants were instructed "to walk with their preferred gait velocity" in all conditions. Participants had a short break of approximately 10 s between trials, and a 5-minute break between conditions. Fatigue was never an issue during the recordings. When using RW, the handlebars were adjusted to 48% of body height which is expected to be the optimal height for muscular loadings (Takanokura, 2010). Participants were instructed to find the most comfortable placement of their hands on the vertical handlebars on the CW as recommended by the manufacturer.

2.3. Data recording.

At every trial, full-body kinematic data were captured by a system based on inertial measurement units (IMU) from Xsens (Xsens Technologies BV, Enschede, The Netherlands), sampling at 240 Hz. Each IMU includes tri-axial gyroscopes, accelerometers, and magnetometers and were placed on seventeen body segments: head, sternum, shoulders, upper arms, lower arm, hands, pelvis, thighs, shanks, and each foot (Figure 2). The Xsens system fused the sensor information based on a proprietary algorithm to provide joint angles (Roetenberg et al., 2009). During walking the Pedar-x system (Novel GmbH, Munich, Germany) sampled vertical ground reaction forces (GRF) applied to the feet at 100 Hz. The system utilizes 99 capacitive sensors in a sole and was inserted into the participants' shoes and fitted to their shoe size. During walking data were stored on a SD memory card. The body weight of the participants was determined when quietly standing upright. Both systems could detect steps independently and onset and offset of the steps were used for synchronizing the data streams.

2.4. Data analysis.

2.4.1. Largest Lyapunov exponent. Hip, knee and ankle joint angles were only extracted in the sagittal plane, due to known measurements errors from other planes (Zhang et al., 2013). Joint angles were analysed without filtering the data as recommended for nonlinear analysis of signals (Bruijn et al., 2009; Dingwell and Cusumano, 2000). For time series of the joint angles recorded in each trial q(t), a state space reconstruction was performed using:

$$S(t) = [q(t), q(t+\tau), \dots, q(t+(d_E-1)\tau] \quad (1)$$

with S(t) representing the d_E-dimensional state vector, τ being a fixed time delay, and d_E the embedding dimension (Figure 3) (Dingwell and Cusumano, 2000). For each time series, two initial steps were removed to reduce transient effects. Afterwards each time series consisted of 30 consecutive steps, and was resampled to 4500 data points, due to known sensitivity of the applied method to the data length (Mehdizadeh, 2018). The time delay for reconstruction was determined as the autocorrelation function dropped to 1/e (Fraser and Swinney, 1986). The procedure was performed for the entire set of trials and time delay was then determined as the average delay for the entire data set, because group average state space reconstruction method provides the highest reliability (Raffalt et al., 2018). The time delay was set to τ = 60, 41 and 31 samples for hip, knee and ankle joint angles, respectively. Embedding dimensions were calculated using a global false nearest neighbours analysis (Bruijn et al., 2009; Dingwell and Cusumano, 2000; Kennel et al., 1992). Embedding dimensions were found to be 5, in agreement with previous studies (Mehdizadeh, 2018). From the constructed state space, Euclidean distances between neighbouring trajectories were calculated as a function of time and averaged over all original pairs of nearest neighbours to get the logarithmic rate of divergence, using Wolf's algorithm (Wolf et al., 1985), thereby obtaining an estimate of λ_{max}.

2.4.2. Gait parameters. For each trial, the mean peak hip, knee and ankle joint angle in the sagittal plane, gait velocity defined as the mean velocity of the centre of mass, stride length defined as the distance between heel contact of the same limb, cadence defined as the number of steps per minute, and stride duration defined as the interval between consecutive heel strikes of the same foot were obtained as the mean across the same 30 steps as used in the estimation of λ_{max}.

2.4.3. Partial weight bearing. The onset of heel contact was determined when the vertical GRF exceeded 30 N and toe off were identified when the vertical GRF returned to a value of 30 N. Data were resampled to 100 data points representing 0 - 100% of foot contact and the average force during foot contact was calculated (Figure 4). Lastly, the average PWB was calculated as the mean weight bearing of the feet across the 30 steps, as a percentage of the participants' body weight.

The data analysis was performed using MATLAB version 2018b (MathWorks Inc., Natick, MA, USA).

2.5. Statistical analysis

One-way repeated-measures ANOVA, with Bonferroni correction, was performed to analyse differences between the three conditions (i.e. NG, RW, CW). The dependent variables were λ_{max} calculated for the hip, knee, and ankle joint angle, using Wolf's algorithm, mean peak values of the joints, gait velocity, stride length, cadence, stride duration, and PWB. Data are presented as mean ± standard deviation (SD) unless stated otherwise. All tests were conducted using SPSS version 25 (SPSS Inc., Chicago, IL). The significance level was set at $p < 0.05$.

3. Results

3.1. Local dynamic stability

Overall, local dynamic stability revealed a significant main effect of the conditions on hip, knee and ankle joint (Table 1). The local dynamic stability of the hip, knee and ankle joint was significantly more unstable with RW compared to the CW ($p < 0.001$) and NG ($p < 0.001$). No significant difference was revealed between NG and CW in all measured joints ($p \approx 1$).

3.2. Kinematic peak joint angles

Sagittal plane peak joint angles for the hip, knee and ankle joint revealed significant main effect of the conditions on the hip and ankle joint (Table 2). Peak hip flexion was higher ($p < 0.001$) and hip extension ($p < 0.001$) was lower with RW compared with other conditions, and no significant difference was observed between NG and CW ($p \approx 1$, $p = 0.5$ for hip flexion and extension, respectively). Dorsiflexion was significantly lower with RW compared to NG and CW ($p < 0.01$) and plantarflexion was significantly higher during NG compared to RW and CW ($p < 0.01$).

3.3. Gait parameters and partial weight bearing

Inspection of the gait parameters and PWB revealed a significant main effect of conditions (Table 3). NG had significantly faster gait velocity ($p < 0.001$) and longer stride length ($p < 0.05$) compared to RW and CW, and no significant difference was observed between RW and CW ($p \approx 1$, $p = 0.7$ for gait velocity and stride length, respectively). A lower cadence ($p < 0.001$) and longer stride duration ($p < 0.001$) was observed with the CW compared to NG and RW, but no significant difference was observed between NG and RW ($p = 0.07$, $p = 0.10$ for the cadence and stride duration, respectively). Lastly, PWB with RW was significantly lower than NG and CW ($p < 0.001$), where no significant difference was found between NG and CW ($p = 0.25$).

4. Discussion

The current study investigated the effect of walker type on local dynamic stability in lower extremities with young healthy adults. The main findings were that RW differs negatively from NG and CW as reflected in less stable local dynamics. Moreover, RW supported the participants' weight to a higher extent, therefore, less weight was supported by the feet (i.e., lower PWB) when comparing the conditions.

Our results showed that participants had a faster gait velocity during NG compared to walking with walkers. This is in accordance with literature showing a faster NG compared to assisted walking (Youdas et al., 2005). In agreement with previous studies, we found that an increased gait velocity was associated with an increased cadence, stride length, and a lower stride duration during NG (Liu et al., 2009; Youdas et al., 2005). Additionally, the obtained range of variation for the gait parameters in NG were similar to those reported in previous studies (Ostrosky et al., 1994). However, all gait parameters (i.e., gait velocity, cadence, stride length and stride duration) in CW were significantly different from NG, whereas only a lower gait velocity and a shorter stride length characterized the gait with RW compared to NG. This finding was different from what we hypothesized as we expected the gait parameters with CW would be more similar to NG. This is likely due to increased complexity of coordination of upper and lower limb with the given mechanical construct of the CW. Previous studies highlighted that the interlimb coordination of upper and lower extremities is performed by shared neural centres involved in the control of movements (Ehrsson et al., 2000). When comparing the walkers, given the higher cadence in RW and not a significantly different stride length, the gait velocity in RW would intuitively be expected to be higher than CW, which contrasts with our observation of not finding a significantly different velocity comparing the walkers. This is likely due to a longer mean stride length in CW, albeit not significant. Thus, it is then fair to assume that the cadence has a higher power discriminating RW from CW in comparison to gait velocity.

Previously, standard walkers and wheeled walkers have been tested, and the results showed that especially the use of wheels resulted in a more normal-appearing gait (Bohannon, 1997). Assuming that normal-appearing gait was the desired outcome, when comparing walkers, the RW would indicate a more similar gait parameters to NG, according to our observations. However, general gait kinematic patterns does not carry enriched information about the gait local dynamic stability and how robust the gait may be against small divergence of state trajectories (Bruijn et al., 2009; Dingwell and Cusumano, 2000).

We applied Wolf's algorithm to estimate λ_{max}, however, current literature is equivocal on the appropriate choice of algorithm in this context (Mehdizadeh, 2018). An alternative to Wolf's algorithm is Rosenstein's algorithm, which has been used more widely in gait analysis (Mehdizadeh, 2018). However, when analysing small experimental gait data sets, Wolf's algorithm seems to be more appropriate. For example, Cignetti and colleagues (Cignetti et al., 2012), using both algorithms, showed that the local dynamic stability in older individuals may improve compared to young individuals when they used Rosenstein's algorithm, which might seem counterintuitive, considering individual factors that are known to influence gait stability negatively with age (Decorps et al., 2014; Hamacher et al., 2019; Tieland et al., 2018).

Most evaluations of instability are performed on a treadmill, but treadmill walking is associated with increased stability (Dingwell et al., 2001). Therefore, investigating the local dynamics of gait stability on the floor is of importance for valid measurements applicable to everyday gait. If this recording is not performed on a treadmill, the possibility of acquiring many steps without causing any interruption to the gait is largely restricted. Therefore, some studies have suggested capturing the dynamics of systems through many trials (Mehdizadeh, 2018). In the present study, the number of steps was lower than the contradicting recommended number of steps (i.e. 54 and 150 steps or 2-3 minutes of walking) for a consistent estimation of λ_{max} (Mehdizadeh, 2018). As a remedy, we captured 10 trials per condition to reduce the variance of estimation (Sloot et al., 2011). However, the estimations of λ_{max} have been performed even with low number of strides (e.g. 30) for lower body joint angles (England and Granata, 2007).

It was hypothesised that the gait local dynamic stability with the CW would be more stable than RW, but not more than NG. When comparing the conditions, the results showed that the local dynamic stability with RW was significantly more unstable for the hip, knee and ankle joint compared to NG and CW. These findings showed that this newly developed walker provided more gait stability than RW. Furthermore, no significant difference was observed in stability between NG and CW. The difference in gait velocity may explain why the local dynamics stability was not significantly different comparing NG and CW, as gait velocity would influence the estimated λ_{max} (Bruijn et al., 2009; England and Granata, 2007). The increased local dynamic stability with the CW is, likely, due to the effect of arm swing (Punt et al., 2015; Wu et al., 2016) which may contribute to reduced metabolic costs as seen during normal gait compared to restricted arm swing (Collins et al., 2009). From the standpoint of the dynamical system theory, the degrees of freedom of the whole-body joint kinematics of the system in CW, compared with RW, had a closer dimensionality to that of the system in NG, especially in the hip joint, possibly due to the upper extremities being restrained in RW. However, one should make a distinction between the intra-limb coordination stability reflected in the variability of the joints relative phase (Seay et al., 2006), compared with local dynamic stability which reflects the time-dependent structure of joint angles. Prior literature presented the theoretical aspects of these two concepts (Jordan et al., 2009).

From the perspective of rehabilitation practitioners, our results showed that the PWB was significantly lower using the RW, meaning that more of the body weight was carried by the RW. According to the literature it is important to alleviate pain and off-load the lower extremities to reduce the recovery time from surgeries and injuries (Bohannon, 1997; Liu et al., 2009; Martins et al., 2015; Youdas et al., 2005). Therefore, the CW might not be applicable in early stages of recovery, where weight bearing is crucial. However, as the recovery progresses, loading to the lower extremities should increase and PWB would be of less significance (Martins et al., 2015). Given lower PWB and more stable gait local dynamics, CW may be a better choice of an assisting device at later stages of the recovery to regain the function of lower extremities. One thing to note is that the tested product was an initial prototype to be modified at later stages. Therefore, further studies are warranted to establish the effect of CW on the gait local dynamic stability in the elderly population and patients with impaired stability to fully understand the potential implications of this device.

5. Conclusion

In summary, we found that the gait with CW had more stable local dynamics than the gait with RW in the lower extremities. Further, gait with CW was, surprisingly, as stable as normal gait. Our results showed that gait with CW facilitated similar peak joint kinematics to normal gait in the lower extremities, especially in the hip joint. Despite the potentially beneficial effects of CW, it supported the participants weight to a lower extent compared with RW and, this prototype, may not be sufficient during the early stages of rehabilitation and recovery from surgery. Nevertheless, the evidence suggests that CW could be a steppingstone towards normal gait with respect to the elderly and patients with impaired stability.

Acknowledgements

The authors would like to thank B. Vangsgaard, CEO and founder of Human Walking ApS, for providing the Crosswalker used in this study. We would also like to thank the volunteers for their participation.

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Table 1. The mean values of the local dynamic stability measured with Wolf's algorithm with the corresponding standard deviations are presented.

λ_{max} (bits/sec)	NG	RW	CW	F(2,34)	P	η^2
Hip joint	0.56 ± 0.25	0.81 ± 0.26*‡	0.57 ± 0.19	19.43	< 0.001	0.53
Knee joint	0.99 ± 0.29	1.18 ± 0.28*‡	1.00 ± 0.27	14.50	< 0.001	0.46
Ankle joint	2.57 ± 0.91	3.24 ± 1.05*‡	2.47 ± 0.80	19.79	< 0.001	0.54

*Indicates a significant difference between NG. ‡Indicates a significant difference between CW.

Table 2. Comparison of joint kinematics on the lower extremity in the sagittal plane. Mean and standard deviation are presented in for both flexion and extension for hip, knee and ankle joint. The statistical results from ANOVAs are included.

Joint (°)	Movement	NG	RW	CW	F(2,34)	P	η^2
Hip Peak	Flexion	21.82 ± 2.28	25.55 ± 3.18*‡	21.80 ± 2.37	19.16	< 0.001	0.53
	Extension	22.75 ± 3.35	16.70 ± 4.41*‡	22.27 ± 3.29	54.17	< 0.001	0.76
Knee Peak	Flexion	68.34 ± 3.48	68.10 ± 3.11	67.94 ± 3.45	0.97	0.39	0.05
	Extension	0.69 ± 2.67	0.95 ± 2.87	0.46 ± 3.05	2.26	0.12	0.12
Ankle Peak	Dorsiflexion	15.48 ± 2.73	14.26 ± 2.52*‡	16.04 ± 2.52	17.89	< 0.001	0.51
	Plantarflexion	19.02 ± 5.91†‡	15.78 ± 5.88	16.53 ± 5.77	14.20	< 0.001	0.46

*Indicates a significant difference between NG. †Indicates a significant difference between RW. ‡Indicates a significant difference between CW.

Table 3. Comparison of gait parameters as well as partial weight bearing between conditions are shown with the mean and standard deviation.

Variables	NG	RW	CW	F(2,34)	P	η^2
Gait velocity (m/s)	$1.51 \pm 0.16^{\dagger\ddagger}$	1.39 ± 0.18	1.38 ± 0.18	16.23	< 0.001	0.49
Stride length (m)	$1.59 \pm 0.13^{\dagger\ddagger}$	1.51 ± 0.14	1.54 ± 0.14	7.75	< 0.01	0.31
Cadence (steps/min)	116.89 ± 5.75	115.32 ± 5.85	$111.06 \pm 6.24^{*\dagger}$	30.80	< 0.001	0.64
Stride duration (s)	1.08 ± 0.05	1.09 ± 0.05	$1.14 \pm 0.06^{*\dagger}$	27.46	< 0.001	0.62
PWB (%BW)	82.77 ± 6.50	$70.26 \pm 7.33^{*\ddagger}$	80.94 ± 5.96	55.35	< 0.001	0.77

*Indicates a significant difference between NG. †Indicates a significant difference between RW. ‡Indicates a significant difference between CW.

Figure captions

Figure 1. The three conditions that the participants performed during the experiment.

Figure 2. Instrumentation of the 17 IMU sensors (orange boxes) and the Pedar-X system. IMUs were placed on the interior of the suit.

Figure 3. Schematic representation of local dynamic stability analyses of the knee joint flexion/extension angle. Original knee angle time series data (A). The 3D reconstructed state space according to equation (1) giving a mapping of the knee motion in relation to the time delay (B). For visualization purposes only 3D reconstruction was presented. Note that the stride-to-stride variation adds a noise like complexity to the behaviour.

Figure 4. Schematic representation of data processing of the data acquired with the Pedar-x system. Heel contact and toe off were detected using the 30 N cut-off and subsequently divided into 0-100% of foot contact for each foot.

Figure 1.

Figure 2.

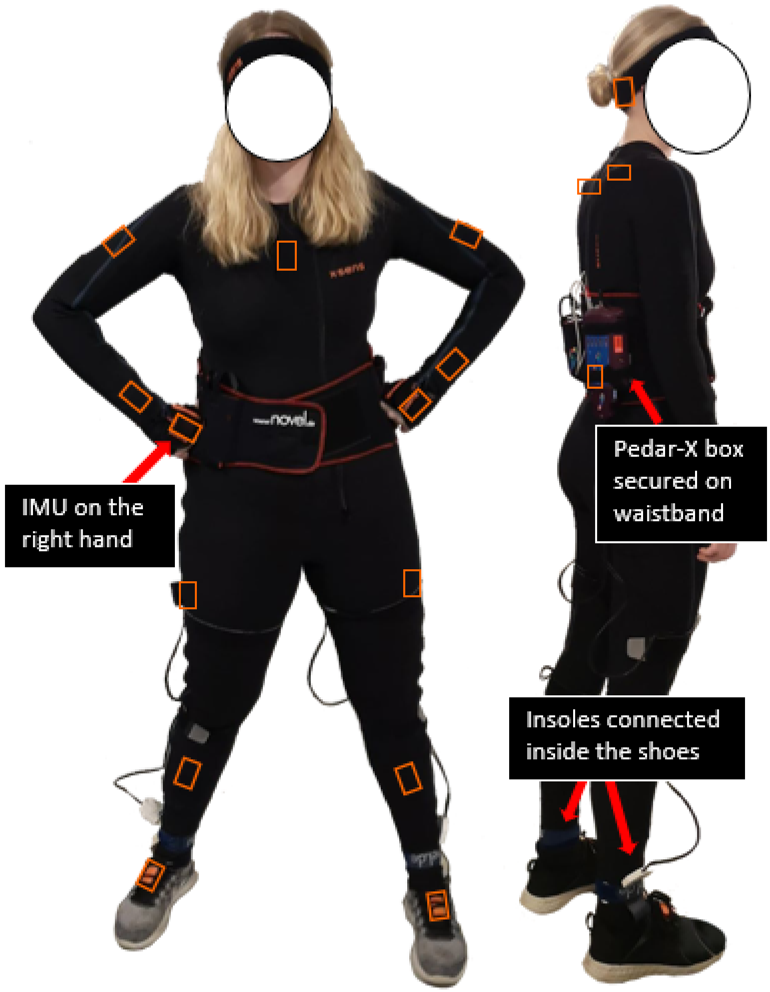
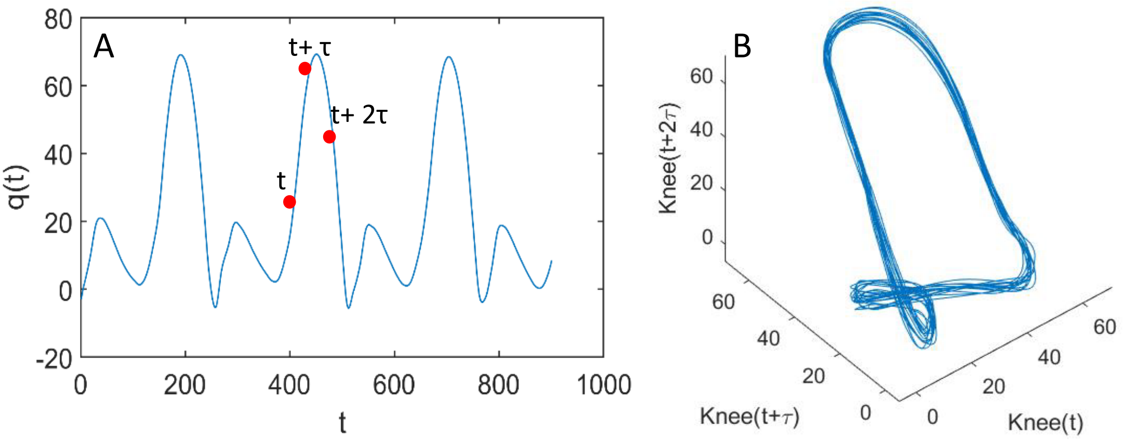

IMU on the right hand
Pedar-X box secured on waistband
Insoles connected inside the shoes

Figure 3.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B
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t+ τ
t+ 2τ
q(t)
t
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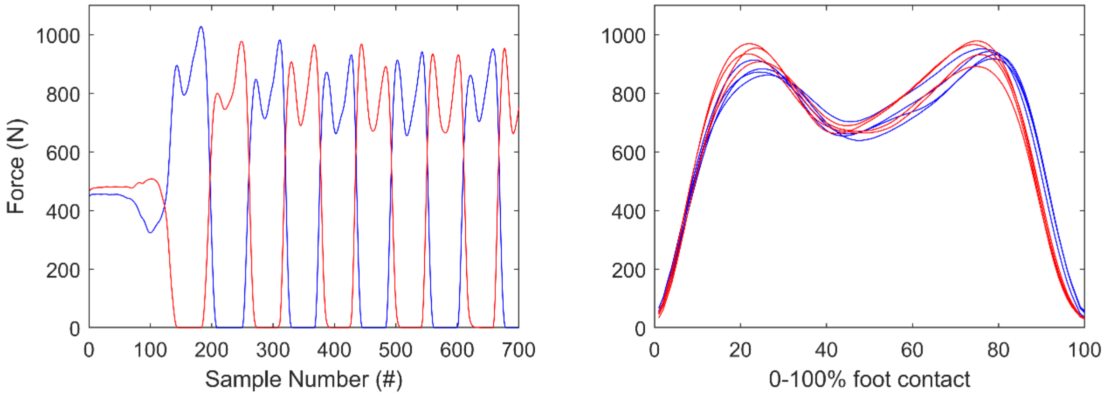
60
40
20
0
-20
0
200
400
600
800
1000
Knee(t+2τ)
Knee(t+τ)
Knee(t)
60
40
20
0

Figure 4.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700
Sample Number (#)
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0-100% foot contact