

## Validity of DynaPort GaitMonitor for assessment of spatiotemporal parameters in amputee gait

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**Abstract**—Accelerometry can be used to objectively assess the walking ability of people with a lower-limb prosthesis inside and outside the laboratory setting. In this study, the validity of the DynaPort GaitMonitor software (McRoberts, The Hague, the Netherlands) for assessing spatiotemporal parameters of amputee gait was evaluated. Fourteen subjects with a lower-limb prosthesis walked on a straight level walkway at a self-selected walking speed over three different distances. During walking, we measured pelvis acceleration using a triaxial accelerometer (DynaPort MiniMod). Mean spatiotemporal parameters were derived from these signals using the DynaPort GaitMonitor software. Similar parameters were simultaneously determined from video. Overall, the number of steps, mean step time, step length, and walking speed were detected accurately by the GaitMonitor software. No systematic deviation was found, and the accuracy of the different parameters was within 6.5%. However, step times measured separately for both the intact and prosthetic legs differed considerably between the GaitMonitor and the video. Step time was systematically underestimated by the GaitMonitor for the intact leg and overestimated for the prosthetic leg. We concluded that the DynaPort GaitMonitor is a valid instrument for assessing mean spatiotemporal parameters in amputee gait, although systematic errors in prosthetic and intact heel strike detection prevent a reliable analysis of walking symmetry.

**Key words:** accelerometry, activity monitor, ambulatory monitoring, amputee, gait, prosthesis, step length, step time, walking ability, walking speed.

## INTRODUCTION

Regaining walking ability is one of the most prominent goals in the rehabilitation program after lower-limb amputation. Therefore, the capability to adequately monitor walking ability is required to evaluate and guide the rehabilitation process. Walking ability can be assessed in many ways. Functional walking ability can be assessed by observation or questionnaires [1]. These measures provide information on the ability of the patient to walk independently in different, more or less complex, situations of daily life. However, these subjective measures do not contain information on the quality of the gait pattern. The quality of gait can be studied in terms of joint kinematics and kinetics using laboratory-based motion-analysis systems [2–3]. These measurements provide detailed information on movement execution and joint loading but are confined to the controlled environment of the laboratory. Both types of walking ability assessment have their

**Abbreviations:** LoA = limits of agreement, BCM = body's center of mass, PAM = patient activity monitor, SD = standard deviation.

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distinct but different uses. However, an increasing need exists for an intermediate type of assessment that provides information on the quantity of daily physical activity as well as the quality of gait in a realistic everyday life environment [4–5].

In recent years, accelerometry-based systems have been developed and improved such that they meet the requirements of an ambulatory gait analysis system [6]. They allow the evaluation of several gait features of different patient populations in daily life. Spatiotemporal parameters of human gait, such as step time, step length, and walking speed, have been successfully derived from the acceleration signals [6–7]. However, the validity of the different methodologies and algorithms that have been developed strongly depends on the sensor configuration and probably also on the specific gait pattern of the (patient) population of interest. Applying these methods to new sensor configurations and patient groups should thus be conducted with care.

For prosthetic gait, several accelerometry-based instrumental setups could be considered. The patient activity monitor (PAM) (Össur; Reykjavik, Iceland), which was especially designed for prosthetic gait, consists of a separate biaxial and uniaxial accelerometer combined in one small device worn at the ankle. This setup and accompanying algorithm have been shown to be able to detect walking episodes and spatiotemporal parameters of gait with reasonable accuracy [8–9]. An alternative setup for accelerometer placement is placement at the pelvis at the approximate level of the body's center of mass (BCM) [10–11]. Using this setup, we can measure the general motion of the BCM and detect heel strike patterns of both feet separately using only one device. Hence, in principal, step parameters can be derived instead of only stride parameters, as is the case with the PAM. Zijlstra and Hof described a method for assessing spatiotemporal parameters of gait using an accelerometer at the trunk that is based on the inverted pendulum behavior of the human body during walking [11–12]. This method has been found to accurately assess spatiotemporal characteristics in nondisabled adults [11–12] and children [13], as well as in patients after total hip arthroplasty [14]. The method described by Zijlstra and Hof is incorporated in the commercially available DynaPort GaitMonitor software (McRoberts; The Hague, the Netherlands) [11].

The validity of the DynaPort GaitMonitor has not yet been established for amputee gait. Given the specific changes in the gait pattern of amputees [2–3], group-

specific validation is required before this tool can be used in research or clinical practice for individuals with amputation. Hence, the purpose of the current study was to evaluate the validity of the DynaPort GaitMonitor for the assessment of spatiotemporal parameters of amputee gait.

## METHODS

### Participants

Fourteen participants with a lower-limb prosthesis enrolled in this study (**Table 1**). Participants were recruited from the prosthetic in- and outpatient units at the Helimare Rehabilitation Center (Wijk aan Zee, the Netherlands). People were eligible to participate when they were able to walk with their prosthesis indoors for 40 m at least six times without the use of a walking cane, frame, or crutch (Special Interest Group in Amputee Medicine [SIGAM] classification D–F [15]). The study was approved by the

**Table 1.**  
Subject characteristics ( $n = 14$ ).

Variable	Mean $\pm$ SD or $n$
Sex	
Male	10
Female	4
Age (yr)	49 $\pm$ 18
Height (m)	1.74 $\pm$ 0.08
Weight (kg)	75 $\pm$ 13
Time Since Injury (yr)	26 $\pm$ 24
Amputation Level	
Transtibial	6
Knee Disarticulation	2
Transfemoral	4
Rotationplasty	1
Bilateral Transtibial	1
Cause of Amputation	
Trauma	7
Vascular	4
Tumor	3
SIGAM Classification*	
D	3
E	1
F	10

\*Ryall NH, Eyres SB, Neumann VC, Bhakta BB, Tennant A. The SIGAM mobility grades: A new population-specific measure for lower limb amputees. *Disabil Rehabil.* 2003;25(15):833–44. [\[PMID: 12851094\]](https://pubmed.ncbi.nlm.nih.gov/12851094/)  
SD = standard deviation, SIGAM = Special Interest Group in Amputee Medicine.

medical ethics committee of the VU University Medical Center. All participants provided written informed consent.

### Procedure

Participants were asked to walk over several distances in a straight corridor ( $50 \times 4$  m) in the rehabilitation center. They walked at a self-selected comfortable walking speed without additional walking aids and wearing their own prosthesis and shoes. Each subject walked six times. First, they walked two times in a straight line over 40 m. Subsequently, they walked two times over 25 m and two times over a randomly selected 20, 30, or 35 m stretch. This random selection was applied to blind the assessor (who had no knowledge of any predefined distances) from the actual distance walked. Also, the order in which subjects walked over the latter two distances was randomized. The different distances were marked on the floor by 0.3 m square targets for starting and stopping. Participants started and ended each trial with a 5-second period of standing on one of these targets. The two trials over 40 m were used for calibration (see “Data Analysis” section). The remaining four trials were used to validate the spatiotemporal outcome measures: number of steps, step time, step length, and walking velocity, as calculated by the DynaPort GaitMonitor.

### Instrumentation

Spatiotemporal parameters of the gait pattern were simultaneously monitored with a triaxial accelerometer (DynaPort MiniMod, McRoberts; The Hague, the Netherlands) and a digital video camera setup. The DynaPort MiniMod is a lightweight, ambulant measurement device ( $62 \times 41 \times 18$  mm, 53 g) consisting of a triaxial accelerometer, a secure digital memory card, and a lithium polymer power supply. The MiniMod was worn in a neoprene belt around the waist such that the MiniMod was located at the posterior side of the sacrum at the approximate height of the BCM. Data were stored on the memory card at a sample rate of 100 Hz. Recording of the MiniMod was manually started and stopped approximately 10 seconds before and after each walking trial. For validation of the GaitMonitor data, video recordings (25 Hz) of the walking participants were made using a digital camcorder (DCR-TRV110E, Sony Corp; Tokyo, Japan).

### Data Analysis

After the measurements, data were transferred from the memory card in the MiniMod onto a personal computer and uploaded to the manufacturer’s online Web server for data

analysis. On this server, data are analyzed by the GaitMonitor software. This data analysis is automated, thus no personal interference from the manufacturer was required nor occurred. Nevertheless, randomization and blinding of walking distances were applied to prevent any possible bias. Along with the raw data, only information on the distance walked in the two calibration trials and the subject’s leg length was uploaded to the manufacturer’s Web server. Hence, no information on walking distance, duration, or other aspects of the walking trials to be analyzed was available in the GaitMonitor analysis.

The GaitMonitor software first transforms the three-dimensional accelerometer data to a global reference frame with the  $y$ -axis along the field of gravity and the  $z$ -axis perpendicular to the field of gravity in the mediolateral direction [16]. The GaitMonitor then self-detects episodes of walking in the transformed data based on the threshold of the acceleration vector and the step frequency found during the calibration trials of each individual. Within these walking episodes, steps are recognized using a heel strike detection algorithm as outlined by Zijlstra and Hof [11]. Left and right heel strikes were discerned based on the mediolateral acceleration signal [11]. From the heel strike time series in each detected walking episode, step times were calculated. Step length was calculated using an inverted pendulum model [11]. For this purpose, vertical acceleration was integrated twice to obtain vertical displacement. Step length was then calculated by

$$\text{Step length} = 2\sqrt{2lh - h^2} \quad ,$$

where  $l$  represents leg length and  $h$  represents the maximal vertical displacement during a step. Since this model has been shown to underestimate step length, a calibration factor (obtained from two 40 m trials) was used to correct step length (or in other words to correct effective leg length [ $l$ ] of the pendulum).

The results of the GaitMonitor analyses are returned by the Web server to the user (in this case, the investigators) as a data sheet containing all steps detected, step times, and step lengths calculated by the GaitMonitor software. From these data, we calculated the number of steps, the mean step time, the mean step length, and the mean walking speed for each trial. In addition, for one of the 40 m trials, we also calculated the mean step times of the intact and prosthetic legs separately based on the reference to left and right steps in the data sheet.

The previously mentioned parameters were compared with the same parameters derived from the video. From the video recordings, we counted the number of steps, disregarding the final step taken that placed the second foot on the stop target. Mean step time was calculated by dividing walking time (i.e., the time from the first foot movement at the start of the trial to the first heel strike on the stop target) by the number of steps. Mean step length was calculated by dividing walking distance by the number of steps. To calculate separate intact and prosthetic step times, we derived all heel strike times from video for one of the 40 m trials and averaged the times between intact and prosthetic heel strike (prosthetic step) and between prosthetic and intact heel strike (intact step) separately.

### Statistics

The association of individual gait parameters between the DynaPort GaitMonitor and the video data was described using Pearson correlation coefficients ( $r$ ) ( $p < 0.05$ ) and Bland-Altman plots [17]. The latter associates the mean difference between the two methods for a certain spatiotemporal outcome measure and the 95 percent limits of agreement (LoA) to the mean value calculated from both methods.

## RESULTS

All 14 subjects performed two trials over 25 m. Of these 14 subjects, 3 subjects also performed two trials over 20 m, 8 subjects over 30 m, and 3 subjects over 35 m. Data from two different subjects for one 25 m trial had to be discarded because of technical malfunctioning of the Mini-Mod. Thus, 54 trials remained for analysis. The mean self-selected walking speed over all trials was  $1.19 \text{ m}\cdot\text{s}^{-1}$  ( $\pm 0.20$  standard deviation [SD], range = 0.79–1.47), mean step time was 0.68 s ( $\pm 0.20$  SD), and mean step length was 0.59 m ( $\pm 0.05$  SD) (all based on video analysis).

The number of steps detected by the GaitMonitor was correct in 30 of 54 cases. In 8 cases, 1 step was missed, while in 13 cases, 1 step too many was counted. In 3 cases, more than 1 extra step was detected (2 steps twice, 4 steps once). The Pearson correlation coefficient between the number of steps detected by the GaitMonitor versus the video was 0.99 ( $p < 0.001$ ). A Bland-Altman plot of step count is presented in **Figure (a)**.

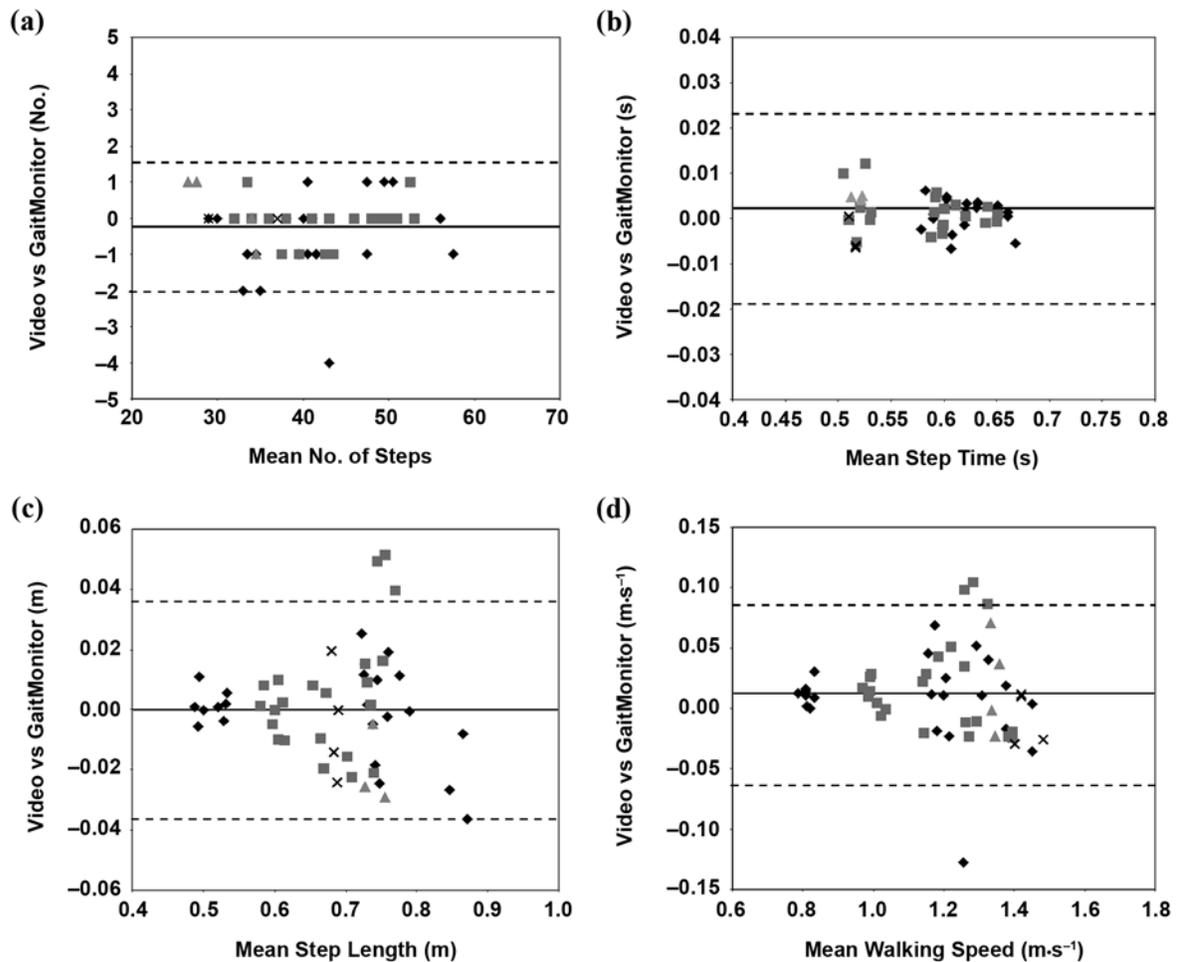
The correlation coefficients between GaitMonitor and video assessment for mean step time, step length, and walking speed are shown in **Table 2**. A high correlation was found for all evaluated parameters ( $r = 0.98$ ). The mean differences and the 95 percent LoA between both methods are presented in Bland-Altman plots in **Figure (b–d)**, and values on absolute and relative differences are presented in **Table 2**. In general, the mean difference was close to zero and did not appear to depend on the magnitude of the given parameter. The LoA between both methods indicate that the difference might be as large as 3.6 percent for mean step time, 5.3 percent for mean step length, and 6.3 percent for walking speed. We should note that these LoA are affected by one outlier, in which case the number of steps was overestimated by four steps by the GaitMonitor, causing a relatively large error in step time (0.072 s, which is beyond the scale of **Figure (b)**) and walking speed ( $-0.12 \text{ m}\cdot\text{s}^{-1}$ ).

In the **Figure**, a distinction is made for the data for the transfemoral (including knee disarticulation) and trans-tibial amputees and for the single bilateral amputee and person with a rotationplasty. Although these subgroups were small, precluding a detailed statistical subgroup analysis, we can see that no clear difference in measurement error was found between these groups in this study.

Separate mean step times for the prosthetic leg (intact until prosthetic heel strike) and intact leg (prosthetic until intact heel strike) from one trial (40 m) of each participant were compared with the separate step times manually derived from video. Thirteen pairs (excluding the data from the double amputee) were analyzed. As can be seen in **Table 2**, the mean step time of the intact leg was substantially lower (7.85%) and the mean step time of the prosthetic leg was significantly higher (6.60%) when derived from the GaitMonitor than from the video analysis.

## DISCUSSION

This study evaluated the validity of the DynaPort GaitMonitor in assessing spatiotemporal parameters of amputee gait. Steps were adequately detected by the GaitMonitor. In most cases, no more than one step was missed by the step-detection algorithm. If steps were missed, these appeared always to be the first or final steps made by the subject (observed through inspection of the individual step reports of the GaitMonitor), which was probably caused by slow and cautious starting or stopping of walking. For the group mean, the difference



### Figure.

Bland-Altman plots for (a) mean number of steps, (b) mean step time, (c) mean step length, and (d) mean walking speed. Solid horizontal lines represent mean difference (vertical axis) between video and DynaPort GaitMonitor (McRoberts; The Hague, the Netherlands) data with respect to mean outcome (horizontal axis) for entire population. Dashed lines represent 95 percent limits of agreement (confidence interval of this difference). Different subgroups are indicated by different symbols: transfemoral and knee disarticulation (♦), transtibial (■), bilateral transtibial (▲), rotationplasty (×).

between the GaitMonitor and video analysis for mean step time, step length, and walking speed was very small (<1%). However, on an individual basis, the differences were larger, ranging from a potential difference (based on the 95% LoA) of  $\pm 3.6$  percent for mean step time to  $\pm 6.3$  percent for mean walking speed. These results did not seem to differ between transfemoral or transtibial amputees in this study.

Our results are comparable or better than results previously reported for amputee gait at equivalent walking speeds. Comparing the PAM against motion analysis, Ramstrand and Nilsson found a comparable difference in walking speed of  $0.01 \pm 0.10 \text{ m}\cdot\text{s}^{-1}$  for treadmill walking

at equivalent walking speeds as in this study [9]. In contrast, Bussmann et al. found a relative error in walking speed of as large as  $7.2 \pm 10.3$  percent [8]. Bussmann et al. indicated that vibrations of the treadmill might have affected acceleration signals of the PAM that was attached to the ankle, which subsequently could affect the analysis of walking distance and speed [8]. However, their setup was very similar to that of Ramstrand and Nilsson, who did not seem to experience this problem [9]. In both studies using the PAM, walking distance was greater than in the present study but how walking distance would affect (relative) measurement error is unclear.

**Table 2.**  
Mean difference and correlation between DynaPort GaitMonitor and video analysis.

Validity	Absolute Mean Difference $\pm$ 2 SD	Relative Mean Difference $\pm$ 2 SD*	Pearson Correlation Coefficient	p-Value <sup>†</sup>
Step Count (No.)	-0.24 $\pm$ 0.89	-0.60 $\pm$ 2.21	0.99	<0.001
Step Time (s)	0.002 $\pm$ 0.021	0.39 $\pm$ 3.58	0.98	<0.001
Step Length (m)	-0.000 $\pm$ 0.036	-0.05 $\pm$ 5.32	0.98	<0.001
Walking Speed (m·s <sup>-1</sup> )	0.012 $\pm$ 0.075	1.00 $\pm$ 6.33	0.98	<0.001
Step Time Intact Leg (s)	0.043 $\pm$ 0.095	7.85 $\pm$ 15.93	0.82	0.01
Step Time Prosthetic Leg (s)	-0.042 $\pm$ 0.094	-6.60 $\pm$ 14.73	0.82	0.01

\*Relative to mean of both methods (%).

<sup>†</sup>p-values apply to Pearson correlation coefficient only.

SD = standard deviation.

The analysis algorithm used in the GaitMonitor has been validated previously for nondisabled adults [11] and children [13]. Despite the abnormalities in prosthetic gait, the differences in mean spatiotemporal parameters between GaitMonitor and video assessment were comparable or even slightly smaller in our study than in the studies on nondisabled subjects. Brandes et al. reported a possible error of  $0.4 \pm 8.2$  percent for mean walking speed in children [13], and Zijlstra and Hof reported an mean error of less than  $0.05 \text{ m}\cdot\text{s}^{-1}$  [11]. The comparable errors found for nondisabled subjects and amputees indicate that for mean gait parameters, the GaitMonitor algorithms are robust for the deviations in the gait pattern of amputees.

The DynaPort MiniMod accelerometer and accompanying GaitMonitor software are substantially more expensive than other often-used pedometers (U.S. \$3,000 vs U.S. \$10–\$200). The latter, however, are primarily designed to evaluate general activity levels (based on the number of steps measured), and only some of these devices report distance traveled. These economic pedometers have been found acceptable for assessing general physical activity [18–20]. However, walking distance has been shown to be measured less adequately with these pedometers, with relatively large errors of more than 10 percent [20]. Hence, more complex and more expensive accelerometry-based systems have their distinct use.

Despite the positive results for mean spatiotemporal parameters of amputee gait, the separate assessment of the mean step times of the intact and prosthetic legs was less successful. Mean step time of the intact leg was systematically underestimated, while that of the prosthetic leg was systematically overestimated. Obviously, these differences disappear when adding them both in stride time or mean step time over both legs. We could not analyze in this study whether heel strike detection of the

prosthetic or intact leg was inadequate, but probably one or both had a systematic error. Because of this error, gait asymmetry could not be analyzed reliably using the DynaPort GaitMonitor. The heel strike detection algorithm would need to be adapted for prosthetic gait to allow this kind of analysis.

From our data, we can conclude that good agreement is reached between both methods for prosthetic gait analysis when group means are compared. Mean differences between both methods are close to zero. On the individual level, larger differences occur, although these differences do not exceed 7.5 percent in 95 percent of all cases for any parameter. Deciding whether this maximal error between methods and repetitions is clinically acceptable is difficult. It depends on the specific aim of the measurement. However, in general, clinically relevant changes in walking ability have been set at 10 to 15 percent [21]. On this basis, the validity of accelerometry-based gait analysis using the DynaPort GaitMonitor would be sufficient to detect such changes.

Some considerations have to be made with respect to the generalizability of the results of this study. In this validation study, we have used a simple, well-controlled walking task. People walked over a short straight distance, without turns or distraction. We found that for this condition, the DynaPort GaitMonitor is a valid tool for assessing spatiotemporal parameters of amputee gait that can be used for clinical evaluation. However, the aim of this ambulatory gait analysis system is also that it be used for evaluation of walking activity during daily living. These walking activities will be less regular (for instance in speed or gait pattern) and therefore the measurement error is expected to be larger. The validity of this system during walking in daily living circumstances will have to be investigated next. In addition, it should be taken into

account that in this study subjects were more or less proficient walkers, since they had to be able to walk without aids and (in retrospect) their self-selected walking speeds were relatively high. The validity of the PAM has been shown to decrease at walking speeds below  $0.75 \text{ m}\cdot\text{s}^{-1}$  [8–9]. Despite the fact that we did not find either a clear effect of walking velocity on the magnitude of the error in our data or a clear difference between transtibial and transfemoral amputees, generalization of our results to less-able amputees should be considered with care.

## CONCLUSIONS

The DynaPort GaitMonitor appears to be a valid tool for assessing mean spatiotemporal parameters in prosthetic gait during a controlled, unidirectional, level-ground walking task. No systematic error was found, and the accuracy of the different parameters was found to be within 6.5 percent. Systematic errors in prosthetic and intact heel strike detection prevented reliable analysis of walking symmetry. Algorithms should be adapted for this. Furthermore, this analysis was made under controlled walking conditions. The validity of the system under more complex walking conditions in daily life should be the topic of future study.

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