Automated Estimation of Initial and Terminal Contact Timing Using Accelerometers; Development and Validation in Transtibial Amputees and Controls

Ruud W. Selles, Margriet A. G. Formanoy, Johannes B. J. Bussmann, Peter J. Janssens, and Henk J. Stam

***Abstract*—The aim of this study was to develop and validate an automated accelerometry-based system for estimating initial contact (IC) and terminal contact (TC) timing information from walking patterns of healthy control subjects and transtibial amputees that can be used in daily life with minimal interference of researchers. Subjects were instrumented with two uniaxial accelerometers just below the knee while synchronized ground reaction force (GRF) recordings were used as reference measurements. An automated multiphase algorithm was developed to estimate the time of IC and TC in the acceleration signals of five healthy subjects and two transtibial amputees walking at different walking speeds. The accuracy of the detection algorithm in ten control subjects and eight transtibial amputees indicated mean errors ranging between 0.013 and 0.034 s for the TC and IC timing, with 95% confidence interval of the individual step errors ranging between −0.062 and 0.115 s. Correlation coefficients between the estimated stance phase duration and GRF data were 0.98 and 0.97 for controls and amputees, respectively. We concluded that IC and TC can be accurately and easily measured using this system in both healthy subjects and transtibial amputees walking at different walking speeds. The system can be used in clinical situations or gait labs as well as during daily life.**

***Index Terms*—Amputation, gait, rehabilitation, signal processing.**

I. Introduction

In PAST decades, gait analysis has become a widely used and powerful tool. It is used, for example, in the preoperative assessment, operative planning, and postoperative evaluation of various patient groups (e.g., [1] and [2]). In addition, gait analysis has been used to optimize functional electrical stimulation (FES) [3] by providing the timing information needed to coordinate the muscle stimulation patterns. Most of the present gait analysis is either observational or instrumented in a lab environment. While observational gait analysis has the advantage of being cheap and easy to perform, the interrater reliability is relatively poor [4], [5] and the data are only qualitative. During analysis in an instrumented gait lab, more repeatable and quantitative information can be obtained (e.g., [2], [6]). However, these measurements are relatively expensive, labor-intensive, measure only in a lab situation and may interfere with the natural walking pattern.

A relatively new alternative is ambulatory gait monitoring, defined as the use of body fixed sensors to observe or record quantitative gait characteristics while the patient is not restricted to a specific location. Recent technological developments have made ambulatory gait monitoring an increasingly interesting technique. Improved sensors and data loggers have become available and sensors as well as data loggers have become smaller, lighter, and cheaper. In addition, increased computational power of computers has made analysis easier and quicker.

Although the variables that can be measured during gait analysis depend on the technique selected, initial contact (IC) and terminal contact (TC) may be the most widely used. IC is a clinically relevant measure because it is used to define the beginning of a complete cycle and therefore provides information about cycle duration and frequency. Combined with TC, marking the start of the swing phase, it provides information about swing time, stance time, and left-right asymmetry. IC and TC have been used as outcome measures in clinical studies, or to provide feedback data for FES. In addition, since it provides information about the different phases of the gait cycle, it is often used as reference data for the measurement of other data such as the knee angle [7].

IC and TC can be measured in different ways. Force sensing resistors (FSRs) placed under the shoe or sole are used in many studies, but due to the sensor location these systems are susceptible to mechanical failure and have poor durability [3]. Based on the aforementioned arguments, Willemsen *et al.* [3] concluded that further improvement of these sensors might be limited. Gyroscopes [8], which measure angular velocity, and accelerometers [3], [9], which measure linear acceleration, have been used as an alternative because these sensors do not need to be placed under the foot and are, therefore, less susceptible to mechanical failure. However, in contrast to FSRs, extracting the timing information from these sensors is less straightforward. Most systems need manual selection of the events from the time series by a trained examiner (e.g., [10] and [11]). An automated system for detecting IC has been developed by Coleman

Manuscript received April 5, 2004; revised October 12, 2004; accepted December 6, 2004. This paper is based on results obtained in the project "Monitoring Amputee Progress with Sensor Socket" (MAPS), which is a research project supported by the Information Society Technologies (IST) program under the Fifth Framework Program of the European Commission, contract IST-2000-27519.

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Digital Object Identifier 10.1109/TNSRE.2004.843176

[12] for lower-limb amputees using a single accelerometer worn around the ankle. The system is based on threshold detection of peaks in the acceleration signal [see Fig. 2(f)–(i)] using a specific threshold for each subject. While Coleman *et al.* [12] report 99% accuracy in the number of strides detected in a full day, the system only counts the number of strides and does not provide any timing information. An important improvement was made by Aminian *et al.* [9] who described an automated algorithm to detect both IC and TC. This system uses multiphase pattern recognition of signals from accelerometers located just above the knee and was found to be highly accurate in healthy subjects and hip arthroplasty patients.

In the present study, we developed and validated a completely automated accelerometry-based system for detecting IC and TC timing information. The study was part of the "Monitoring Amputee Progress with Sensor Socket" (MAPS) project. This project focuses on the development of a prosthetic sensor socket that is ambulatory and can measure physiological parameters such as pressure and oxygen saturation with minimal interference from researchers. The goal of adding accelerometers to the socket was twofold: 1) to provide an activity monitor capable of quantifying gait periods and gait characteristics during daily life while wearing the socket, and 2) to provide timing data of IC and TC for use as reference data to interpret pressure and oxygen saturation data from walking measured during daily life. In the future, longer measurements (months) may relate prosthetic wear and component durability to the quantity and characteristics of the daily activities of the amputees. The automated IC and TC detection system was developed using the data from five healthy control subjects and two transtibial amputees and was validated using an independent data set of ten control subjects and eight transtibial amputees.

II. Methods

A. Subjects

Fifteen healthy control subjects and ten unilateral transtibial amputees participated in the study. The transtibial amputees were included if they could walk without assistance for at least a few minutes, had no major cardiopulmonary, neurological or orthopedic disorders other than their amputation, had no skin problems of the residual limb, and were discharged from the rehabilitation program at least one year earlier. The nonamputee control subjects were selected if they had no known disorders that would influence their walking patterns. All subjects wore their usual daily shoes. The hospital's Medical Ethical Commission approved the study and all subjects gave informed consent to participate in the study.

B. Measurements

The assessment was carried out in a gait analysis lab on a 15-m straight track with a Kistler force platform at about 10 m from the starting position. Before measurement in each condition, subjects walked around the lab to get used to the instrumentation. In addition, test trials were performed to determine a starting position that would lead to hitting the force platform with the instrumented leg. Subjects were asked to look forward to the wall at about eye height and not to pay attention to hitting the force plate.

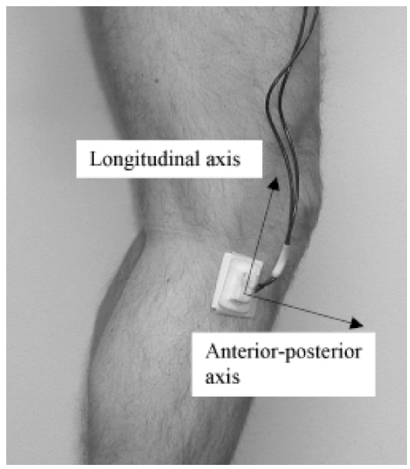

Fig. 1. Picture of the fixation of the accelerometers to the leg and the definition of each accelerometer axis, that is, the longitudinal axis (a_{long}) and the anterior-posterior axis (a_{ap}). In addition to the double-sided tape used to attach the sensors to the skin, an elastic band with Velcro was strapped around leg and sensors to fix the sensors relative to the leg.

Subjects were instructed to walk at different speeds. First, all subjects walked at their self-selected walking speed (SSWS). Then, in the control subjects, we imposed different walking speeds by asking the subjects to synchronize their steps to a metronome at 1, 1.33, and 1.67 Hz (60, 80, and 100 steps/min). Since we anticipated that some transtibial amputees may have problems synchronizing to the metronome, we varied the walking speed in these subjects by asking them to walk slower or faster than their SSWS, without exactly specifying a speed. To determine the speed difference between the conditions in the amputees, walking speed was measured by manually timing the walking duration of the 15-m track. In both groups, after the SSWS condition, we randomly assigned the other conditions to avoid order effects. Between conditions, subjects rested at least two minutes. All subjects walked the 15-m track in each condition until six successful trials were completed, with a successful trial defined as one in which the whole instrumented leg hit the force plate, while the contralateral leg did not hit the force plate.

Subjects were instrumented with two uni-axial piezoresistive accelerometers (±5 g, IC-Sensors 3021) located on the lower leg just below the lateral tibial condyle (see Fig. 1). The sensitive axis of the first accelerometer $(\bar{a}_{\mathrm{long}})$ was in line with the longitudinal axis of the shank, while the second accelerometers was aligned perpendicular, along the anterior-posterior axis of the shank $(\bar{a}_{ap})$. Due to the design of the accelerometers, the signals from the sensors are a combination of an acceleration component as well as gravitational component [13], [14]. The sensors were attached with double-sided tape to the skin so that the axes of the accelerometers remain fixed relative to the shank. Instead of using separate sensors, a single two- or tri-axial accelerometer can also be used. In addition, an elastic band with Velcro was strapped around leg and sensors to assure that the sensors would be fixed during the walking trials. Synchronized ground reaction force (GRF) data (Kistler Instrumente AG, Winterthur, Switzerland) were used as reference measurements. The system

was developed to be used for ambulatory measurements using portable data loggers. However, in the present study, we connected the sensors directly to a pc-based 12-b data acquisition system to obtain synchronized accelerometer and ground reaction force data. Both acceleration and GRF data were sampled at 500 Hz.

C. Data Analysis

Two data sets were created from the measurements. Data from the first five control subjects and the first two transtibial amputees included in this study were used to develop the automated algorithm, while data from the other ten control subjects and eight amputees were used to determine the validity of the developed algorithms.

Data analysis and the development of the automated algorithm were performed in Matlab (The MathWorks, Inc., Natick, MA). In the GRF data, used as reference data, IC and TC were defined as the first moment the vertical ground reaction force was above (IC) or below (TC) 20 N [15] . Stance phase duration (SPD) was defined as the difference between IC and TC.

D. Developed Algorithm

Based on the data from the first five control subjects and the first two transtibial amputees, the automated detection algorithm for IC and TC was developed. The development was an iterative process in which different algorithms were developed and tested by comparing their outcome (estimated IC and TC) with the GRF data. As a final step in the development, the values for the different filter frequencies, time windows, and threshold values for the fast and slow walking speed distinction used in the algorithm were optimized by testing all feasible combinations and comparing its outcomes with the IC and TC derived from the GRF data.

The main steps of the automated algorithm to estimate IC and TC from the acceleration time series are shown in Fig. 2. For all signals, first, the longitudinal acceleration signal $(\bar{a}_{\mathrm{long}})$ was filtered with a second-order low-pass zero-phase-lag Butterworth filter with a cutoff frequency of 0.75 Hz [Fig. 2(A)]. This signal generally had one "local minimum" (both neighboring samples have higher values) per gait cycle. The time between two consecutive minimum values was used to roughly estimate stride duration ("approximate stride duration").

To obtain optimal detection of IC and TC at different walking speeds, separate calculations were performed on the faster and slower strides. Therefore, based on the approximate stride duration, strides were divided into faster strides (defined as strides with an approximate stride duration of less then 1.5 s) and slower strides (approximate stride duration of 1.5 s or more).

For the faster strides, the raw $\bar{a}_{\mathrm{long}}$ signal was filtered again with a second order low pass zero-phase-lag Butterworth filter [Fig. 2(B) and (C)] with a cutoff frequency (f) as a function of the approximate cycle duration $(T_{S,\mathrm{approx}})$ using the following equation:

$$f = 2.5 - 0.4 * T_{S,\mathrm{approx}}.$$

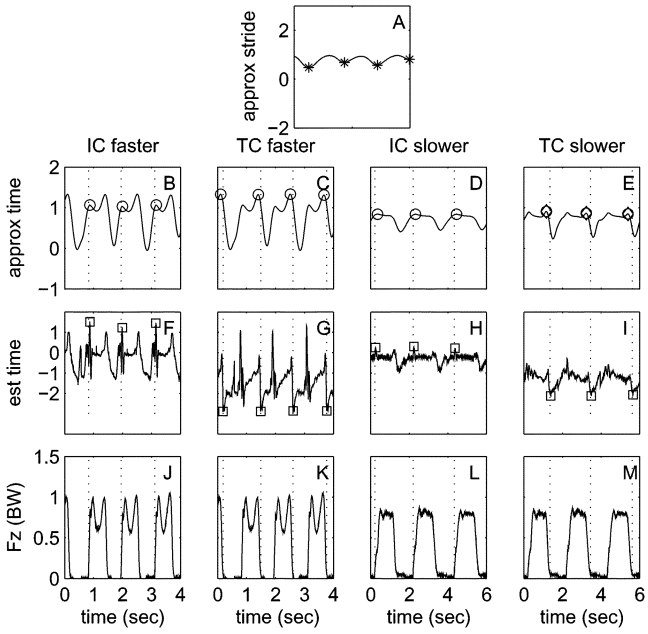

Fig. 2. Graphical representation of the algorithm used to detect IC and TC. For visualization purpose, about three cycles in which the force plate is hit are plotted as consecutive strides. See Section II for detailed explanation. (A) Filtered $\bar{a}_{\mathrm{long}}$ signal, with $*$ indicate the minimum values used to roughly estimate stride duration. (B) Filtered $\bar{a}_{\mathrm{long}}$ signal in which the first peak roughly indicates IC timing for the faster strides. (C) Filtered $\bar{a}_{\mathrm{long}}$ signal in which the second peak roughly indicates TC timing for the faster strides. (D) Filtered $\bar{a}_{\mathrm{long}}$ signal in which the first peak roughly indicates IC timing for the slower strides. (E) Filtered $\bar{a}_{\mathrm{long}}$ signal in which the last peak roughly indicates TC timing for the slower strides.(F) Estimated IC based on peak in the $\bar{a}_{\mathrm{long}}$ signal. (G) estimated IC based on minimum in the $\bar{a}_{ap}$ signal. (H) estimated IC based on peak in the $\bar{a}_{\mathrm{long}}$ signal. (I) estimated IC based on minimum in the $\bar{a}_{ap}$ signal. (J)-(M) Fz, indicating the reference IC and TC [dotted lines in (B)-(I)]. Abbreviations: IC: initial contact; TC: terminal contact; $\bar{a}_{\mathrm{long}}$: longitudinal axis acceleration signal; $\bar{a}_{ap}$: anterior-posterior axis acceleration signal; Fz: vertical ground reaction force; BW: body weight; est: estimated; approx: approximated.

It can be seen from the equation that the cutoff $_{\mathrm{frequency}}$ decreases with increasing approximate stride duration. The resultant signal generally has two peaks within each approximate stride. The first peak roughly corresponds with IC ["approximate IC;" Fig. 2(B)], the last peak with TC ["approximate TC;" Fig. 2(C)]. Finally, IC was estimated as the time of the maximum in the raw $\bar{a}_{\mathrm{long}}$ signal occurring maximally 0.04 seconds before the approximate IC [Fig. 2(F)]. TC was estimated as the time of the minimum in the raw $\bar{a}_{ap}$ signal occurring maximally 0.16 s after the approximate TC [Fig. 2(G)].

For the slower strides, the raw $\bar{a}_{\mathrm{long}}$ signal was filtered twice to obtain the "approximate IC" and "approximate TC." To obtain the approximate IC, the signal was filtered with a second order low pass zero-phase-lag Butterworth filter with a cutoff frequency (f) as function of the approximate cycle duration $(T_{S,\mathrm{approx}})$ using the following equation:

$$f = 1.1 + 0.4 * T_{S,\mathrm{approx}}.$$

In the resultant signal, within each approximate cycle, the first peak roughly corresponds with IC ["approximate IC;" Fig. 2(D)]. A final estimate of IC was obtained as the time of

the maximum in the raw $\bar{a}_{\text{long}}$ signal occurring maximally 0.64 s before the approximate IC [Fig. 2(H)]).

To obtain the approximate TC, the signal was filtered with a second order low pass zero-phase-lag Butterworth filter with a cutoff frequency (f) as a function of the approximate cycle duration $(T_{S,\text{approx}})$ using the following equation:

$$f = 0.3 + 0.4 * T_{S,\text{approx}}.$$

In this resultant signal, within each approximate cycle, the last peak roughly corresponds with TC ["approximate TC;" Fig. 2(e)]. A final estimate of TC was obtained as the time of the absolute minimum in the raw $\bar{a}_{ap}$ occurring maximally 0.36 s after the approximate TC [Fig. 2(I)].

E. Data Comparison

For the ten control subjects and eight amputees used in the validation part of the study, estimated IC, TC, and stance phase duration (SPD) from the six gait cycles in each condition were compared with the corresponding GRF data. In addition, average values of the six strides per subject in each walking condition were calculated. To compare estimated SPD with the GRF data, scatter plots were created and correlation coefficients were calculated (SPSS Inc., Chicago, IL). For the SPD, Bland–Altman plots were used to display the agreement between the estimated SPD and the force plate data. For IC, TC, and SPD, mean error and its standard deviation, as well as the 95% confidence interval of the errors were calculated. P-values smaller than 0.05 were considered statistically significant.

III. Results

A. Subjects

Eight of the 15 control subjects were male. The average age of the control subjects was 29 years (standard deviation (SD)$= 8$, see Table I). The average body weight was 69 kg $(\text{SD} = 11)$ and the average height was 1.74 m $(\text{SD} = 0.1)$. Descriptive data of all transtibial amputees are presented in Table I. Visual inspection during the experiments indicated that all control subjects were able to synchronize their step frequency with the metronome at all three frequencies. Most subjects reported the 1.33 and 1.66 Hz condition to be relatively comfortable, and the 1 Hz condition to be uncomfortably slow. One of the transtibial amputees was unable to walk slower than his relatively slow SSWS and was only measured at SSWS and fast walking speed (see Table I). SSWS in the patients ranged from 0.41 to 1.44 m/s, reflecting a relatively large difference in physical ability and gait quality between the amputees. On average, in the patients, the slow walking speed was 21% slower than SSWS and the fast walking speed was 19% higher than the SSWS.

B. Accuracy of Detection

To indicate the accuracy of detection, the estimated SPD (the difference between estimated IC and TC) was plotted against the reference data for the ten controls and eight transtibial amputees used in the validation part of the study (Fig. 3). For the control subjects, a correlation coefficient of 0.98 $(p < 0.001)$ with the GRF data was found for all strides during all walking conditions, while for the transtibial amputees, a correlation coefficient of 0.97 $(p < 0.001)$ was found. When the relation was calculated using the mean values of the six strides of each subject in each condition, the correlation further increased (see Fig. 3).

TABLE I
DESCRIPTIVE DATA OF EACH INDIVIDUAL TRANSTIBIAL AMPUTEE (TTA) PLUS GROUP MEAN AND SD OF THE AMPUTEES AND CONTROL SUBJECTS. DIAGNOSIS REFERS TO THE REASON FOR AMPUTATION. DATA FROM THE FIRST TWO SUBJECTS WERE USED TO DEVELOP AND OPTIMIZE THE ALGORITHM, WHILE THE OTHER SUBJECT DATA WERE USED TO VALIDATE THE ALGORITHM

S	H (m)	BW (kg)	A (yr)	M/F	D	Walking speed (m/sec)		
						SSWS	Slow	Fast
TTA 1	1.93	80	23	M	T	1.41	1.01	1.51
TTA 2	1.58	72	73	M	V	1.25	0.85	1.54
TTA 3	1.76	88	60	M	V	1.44	0.97	1.41
TTA 4	1.65	80	62	F	T	0.71	0.62	1.23
TTA 5	1.86	73	53	M	V	1.04	0.96	1.27
TTA 6	1.60	64	74	F	V	0.89	0.71	0.94
TTA 7	1.96	101	50	M	V	1.12	0.68	1.33
TTA 8	1.83	73	36	M	V	1.16	0.65	1.16
TTA 9	1.82	88	69	M	V	0.41	0.48	0.59
TTA 10	1.73	89	56	M	T	0.42	-	0.61
Mean TTA's	1.77	81	56			0.94	0.74	1.12
SD	0.13	11	16			0.36	0.17	0.34
Mean controls	1.74	69	29	8M,				
SD	0.1	11	8	7 F				

S = SUBJECT, H = HEIGHT, BW = BODY WEIGHT, A = AGE, M = MALE, F = FEMALE, D = DIAGNOSIS, T = TRAUMATIC, V = VASCULAR,

To further indicate the similarity between the estimated SPD and the SPD from the GRF data, Bland–Altman plots were created (Fig. 4) to indicate the mean errors, the errors of the individual steps, and the 95% confidence intervals of the errors. From these plots, we concluded that the error was not strongly influenced by the duration of the SPD in both groups, suggesting that the SPD error is not influenced by walking speed. For the control subjects, the mean error of the SPD was 0.015 s (Table II). For the transtibial amputees, the mean error of the SPD was 0.020. The 95% confidence intervals indicated that the errors of the individual IC, TC, and SPD estimation were maximally 0.133 s (for the stance phase duration in the transtibial amputees), although generally less. It can be seen from Table II that the 95% confidence interval of the error was smaller when the average of six strides was used.

Separate analysis of the accuracy of IC and TC indicated that the errors were within the same range as the SPD (Table II).

IV. Discussion and Conclusion

Ambulatory measurement of daily life activities can be a useful addition to clinical or laboratory measurements for various reasons. While mobility measured in a clinical setting primarily assesses what a patient can do in a controlled environment, ambulatory measurement has the advantage of quantifying what a patient does during daily life [12], [14]. As such, ambulatory measurements can be an alternative to questionnaires or patient interviews in which inherent subjectivity can be problematic [16], [17]. More specifically, for the transtibial amputees investigated in this study, clinical gait

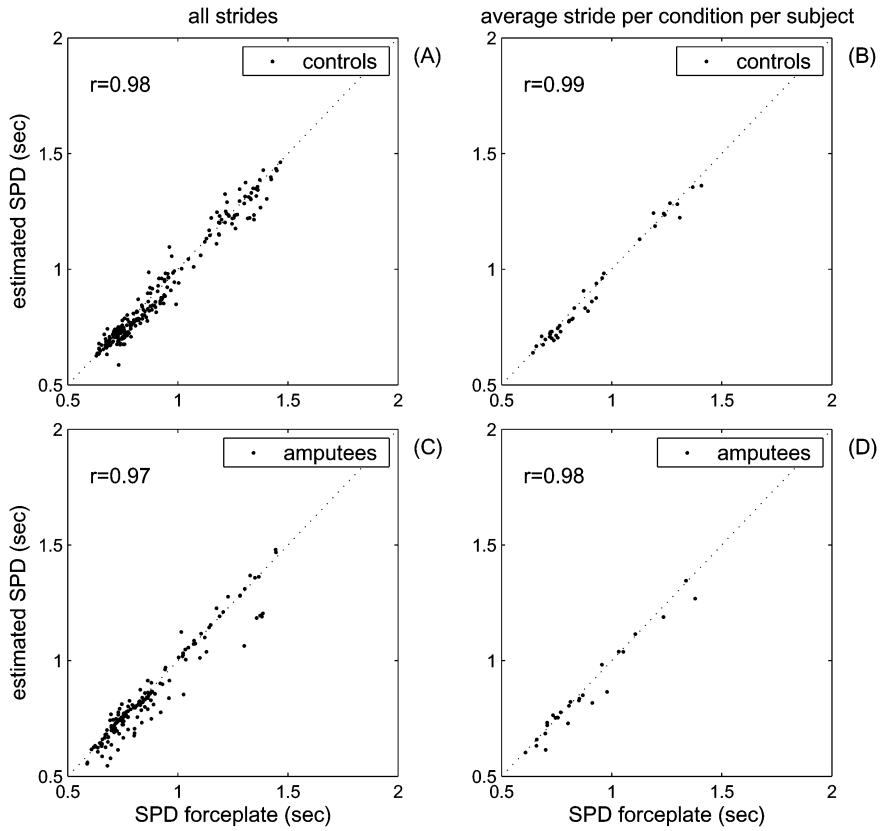

Fig. 3. Correlation between the SPD calculated using the GRF data and the estimated SPD using the developed algorithm for the ten control subjects (A) and eight transtibial amputees (C). For each subject, six separate stance phases are plotted from each walking condition. (B) and (D) show the average data of the six trials for each subject in each walking condition.

analysis evaluates how well the patient walks with a prosthesis in a standardized environment, while the ambulatory measurement technique introduced in this study can evaluate how much patients use their prosthesis as well as how patients walk qualitatively (e.g., step frequency and asymmetry) in daily life. Ambulatory measurements during a few days can therefore provide a clinician or researcher with information on how active a patient is during daily life and which activities the patient performs [18], while measurements of weeks or months may indicate, for example, how much a prosthesis, or a part of it, has been used before it needs to be replaced.

The present study introduces an automated ambulatory measurement system for detecting IC and TC timing in transtibial amputees and healthy control subjects. The system is based on two uniaxial accelerometers placed on the shank, and is capable of measuring in daily life without any interference of researchers. When developing this system, we preferred accelerometers to the often-used FSRs. FSRs have been reported to show mechanical failure [3], while accelerometers are known to be reliable, commercially available and relatively low in cost.

Accurate estimation of IC and TC provides the researcher with a number of clinically relevant parameters such as stride cycle duration, swing time, stance time, and step frequency. Measurement of the same variables of both legs can provide information about left-right asymmetry of stance time and swing time. Evaluating the accuracy of the IC and TC detection in eight controls subjects and eight transtibial amputees indicated mean errors ranging between 0.013 and 0.034 s. Based on an average gait cycle duration of 1.25 s in transtibial amputees [19], these mean errors equal about 0.5% and 2.7% of the gait cycle, respectively. It should be noted that the small mean errors are the result of individual errors that can be larger, but are distributed around zero (see Fig. 4 and Table II). Therefore, a single stride measurement from a single subject may lead to larger errors, as can be seen from the 95% confidence intervals of the individual step errors (Table II). The result of calculating the error based on six strides per subject (last two columns in Table II) indicates how the size of the 95% confidence interval for the error decreased when the average data for each subject in each condition were used compared to the individual stride data. This indicates that in studies more accurate estimates for IC, TC, and SPD can be obtained by calculating the average of a number of strides.

The present study used a single algorithm to estimate IC, TC, and SPD in the transtibial amputees and nonamputees. Although the differences were relatively small, we found that the errors in estimating the events were somewhat larger in the amputees compared to the transtibial amputees (See Table II and Figs. 3 and 4). From the present data, it can not be concluded whether this is related to, for example, a somewhat different or more irregular walking pattern of the amputees compared to the control subjects. Whether the errors found in the present study are acceptable in clinical measurement or research studies depends on the specific question that needs to be answered. While the error of a single individual stride can be relatively high using the present method (see Table II and Fig. 4), the smaller average

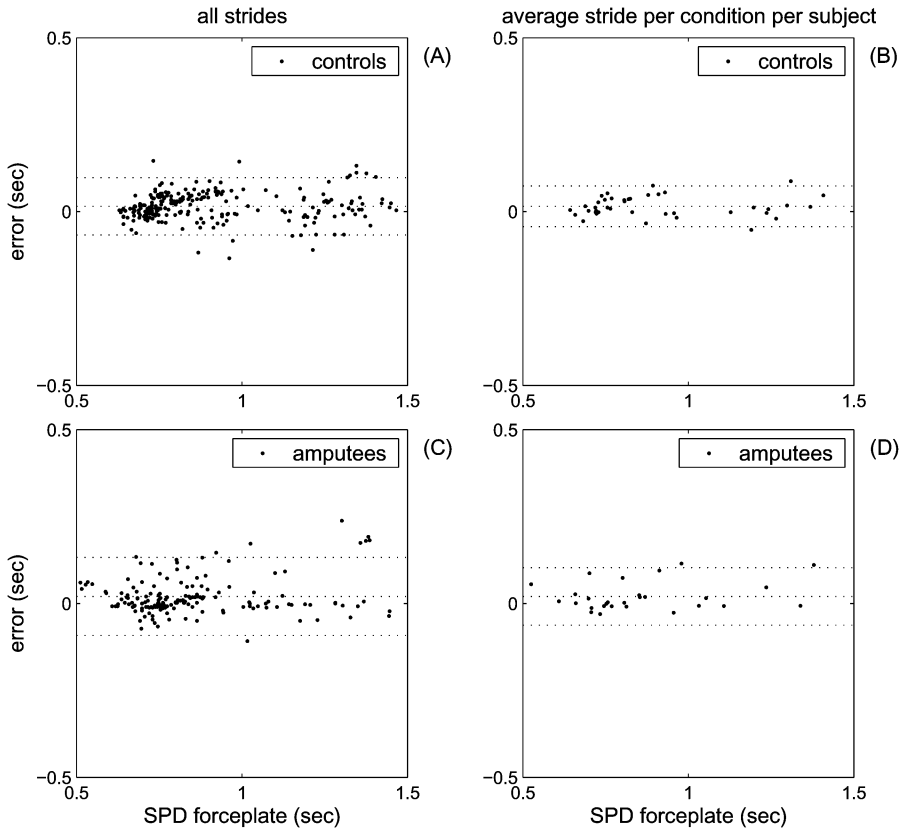

Fig. 4. Bland-Altman plot showing the SPD from the ground reaction force data plotted against the error in the estimation of the SPD using the automated algorithm for the ten control subjects (A) and eight transtibial amputees (C). For each subject, six separate stance phases are plotted from each walking condition. Dotted lines indicate the mean error and the 95% confidence interval of the error. (B) and (D) show the the average error of the six trials for each subject in each walking condition.

TABLE II
MEAN, SD, AND 95% CONFIDENCE INTERVALS OF THE ERROR OF THE DETECTION OF IC, TC, AND SPD FOR THE CONTROL SUBJECTS AND THE TRANSTIBIAL AMPUTEES

	Mean error individual steps (SD)	95% Confidence Interval individual steps	Mean error average step (SD)	95% Confidence Interval average step
Control subjects				
Initial contact	0.034 (0.025)	[-0.015, 0.083]	0.034 (0.016)	[0.003, 0.066]
Terminal contact	0.019 (0.036)	[-0.053, 0.091]	0.019 (0.027)	[-0.036, 0.076]
Stance phase duration	0.015 (0.041)	[-0.067 0.098]	0.015 (0.029)	[-0.043, 0.074]
Transtibial Amputees				
Initial contact	0.033 (0.041)	[-0.049 0.115]	0.033 (0.030).	[-0.027 0.094]
Terminal contact	0.013 (0.038)	[-0.062 0.088]	0.013 (0.030)	[-0.046 0.072]
Stance phase duration	0.020 (0.056)	[-0.092 0.133]	0.020 (0.041)	[-0.062 0.103]

errors indicate that analysis of a larger number of strides will be more accurate (see Table II and Fig. 3) than the analysis of one or two strides. In addition, the high correlation coefficients between estimated SPD and the GRF data suggest that the present methods will be very sensitive in distinguishing between, for example, patients who are walking slower and faster, or will be sensitive to changes within a subject after intervention such as gait training or surgical treatment. As a result, we believe the present algorithm can be used as an outcome measure in clinical studies in which, for example, the effect of an intervention on temporal gait characteristics is evaluated.

The present algorithm is relative complex compared to more simple methods in the literature such as threshold detection methods (e.g., [12]). Testing such a threshold method on the present data, however, we found that the threshold needed to be adjusted in each subject, and needed to be adjusted to different walking speeds. In addition, we found that even after adjusting the threshold for each subject, the threshold would still regularly lead to events that are not detected (when the acceleration peak was to low during the event) or to detect events that did not take place (when there is a peak in the acceleration signal surpassing the threshold that is not related to a specific IC or TC). We believe that the present method has the advantage of not needing adjustment to specific subjects or walking speed and that it is therefore more robust in detecting the events than threshold detection methods.

While the algorithm is relatively complex, we found that calculating all events from several minutes of walking on a modern PC (2-GHz 512-MB internal memory) took only a few seconds of calculation. This suggests that evaluation of measurements of several days with an up-to-date personal computer will not be problematic from a computing point of view.

The present system differs from a system proposed by Aminian *et al.* [9] by using sensors on the shank instead of the thigh. This setup was chosen mainly for practical reasons, since the system will be included in an instrumented prosthetic liner for transtibial amputees. It should be noted that the acceleration signals depend on the orientation of the sensitive axis of the accelerometers. In addition, the signal will also depend on the location of placement. Therefore, the results of the present study can not directly be used when the sensor location or orientation is significantly changed. In our experience, however, accelerometer signals are not importantly influenced by small orientation or location changes.

In contrast to Arminian *et al.* [9], we systematically studied the effect of different walking speeds. We found that separate algorithms needed to be developed for slower and faster walking speeds, and that the cutoff values of some of the low-pass filters needed to be adjusted for cycle duration to get optimal results for all subjects at all walking speeds. Since many clinical studies will evaluate subjects that walk slower than healthy subjects, we believe that accurate detection of slower walking speeds is a clinically important feature of the present method. In the present study, we found correlations between estimated SPD and the reference data of 0.97–0.98. This is in line with Armenian [9], finding correlations of 0.97. Comparison of the SPD values reported by Arminian (0.5–0.8 s) and the present study (0.6–1.5 s) suggests that the subjects in the present study walked considerably slower. However, the similarities between both studies suggest that this approach of measuring gait characteristic is a reliable, valid, easy to use, and affordable technique that can be used in healthy subjects as well as in different patients with different levels of amputation or in other diagnosis groups.

ACKNOWLEDGMENT

The authors would like to thank J. Storm for his help during the experimental setup of this study, and F. Schasfoort and H. Horemans for their review of this manuscript.

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