Gait & Posture 37 (2013) 229–234

Contents lists available at SciVerse ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Quantitative estimation of foot-flat and stance phase of gait using foot-worn inertial sensors

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ARTICLE INFO

Article history:
Received 14 November 2011
Received in revised form 8 May 2012
Accepted 13 July 2012

Keywords:
Gait
Stance phase
Inertial sensors
Pressure insoles
Temporal parameters
Ankle osteoarthritis

ABSTRACT

Time periods composing stance phase of gait can be clinically meaningful parameters to reveal differences between normal and pathological gait. This study aimed, first, to describe a novel method for detecting stance and inner-stance temporal events based on foot-worn inertial sensors; second, to extract and validate relevant metrics from those events; and third, to investigate their suitability as clinical outcome for gait evaluations. 42 subjects including healthy subjects and patients before and after surgical treatments for ankle osteoarthritis performed 50-m walking trials while wearing foot-worn inertial sensors and pressure insoles as a reference system. Several hypotheses were evaluated to detect heel-strike, toe-strike, heel-off, and toe-off based on kinematic features. Detected events were compared with the reference system on 3193 gait cycles and showed good accuracy and precision. Absolute and relative stance periods, namely loading response, foot-flat, and push-off were then estimated, validated, and compared statistically between populations. Besides significant differences observed in stance duration, the analysis revealed differing tendencies with notably a shorter foot-flat in healthy subjects. The result indicated which features in inertial sensors' signals should be preferred for detecting precisely and accurately temporal events against a reference standard. The system is suitable for clinical evaluations and provides temporal analysis of gait beyond the common swing/stance decomposition, through a quantitative estimation of inner-stance phases such as foot-flat.

1. Introduction

In clinical gait evaluation, stance phase is defined as the period of time where the foot is in contact with the ground [1]. Stance has been also described as a succession of different sub-phases such as loading response, mid-stance, terminal stance and pre-swing [2]. Gait changes in elderly persons have been characterized by a longer foot-flat [3]. Those previous studies show that quantitative assessment of sub-phases of stance (referred as "inner-stance phases"), such as foot-flat, can bring additional insight into clinical gait assessment.

Stance phase has been detected using stationary devices such as optical motion capture, force-plate [4] and electronic walkways embedding pressure sensors [5]. Ambulatory devices such as footswitches [6], pressure insoles [7], accelerometers [8,9], gyroscopes [10,11], and combinations of inertial sensors and pressures sensors [12,13] were also used for this purpose. Applications range from the real-time triggering of electrical stimulators to the estimation of temporal parameters that have shown to be relevant for various clinical evaluations such as frailty in the elderly [10,14] or motor symptoms in Parkinson's disease [15].

Using ambulatory measurements for temporal analysis, information can be reliably derived from large datasets collected in natural long-distance gait. Nevertheless, in most previous studies, stance phase was considered as a single block without any subdivision from heel-strike to toe-off [6,9–11,16]. On the other hand, studies that considered inner-stance phase events [8,12,13], did not assess thoroughly the technical validity of their method in terms of temporal precision and accuracy against a gold standard. A detailed study of the reliability of gait events detection from various inertial sensors was recently proposed [17], but the authors mainly focused on the sensitivity and specificity of detection when using Foot Sensitive Resistors and on a limited population, rather than on temporal precision and accuracy.

The goal of this paper was two-fold. First, it aimed to show a novel method based on foot-worn inertial sensors to detect temporal events based on robust features of foot kinematic patterns, and extract inner-stance phases defined between pairs of successive events. As a technical validation, the performance of our

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http://dx.doi.org/10.1016/j.gaitpost.2012.07.012

method was compared to force reference measurements on a two-segment foot model. Second, we tested the efficacy of inner-stance phase estimates as a potential outcome measure for clinical gait evaluations, by using the system to compare healthy control subjects to age-matched patients suffering from ankle disease during a 50-m gait test.

2. Method

2.1. Measurement devices and sensor configuration

Ambulatory pressure insoles (Pedar-X, Novel, DE) were used as a reference system to measure the contact time of different regions of the foot with the ground. This pressure sensor technology has shown high linearity, low creep, low hysteresis, and low variability for all performances over the whole sensor matrix [18]. Additionally, it has been reported as accurate and reliable in gait measurements compared to force-plate [7] and repeatable in different foot regions and on different days [19]. Finally, Pedar insoles have been successfully used instead of force-plate for force measurement during gait [20] and clinical evaluation based on temporal and pressure parameters [21]. Therefore, Pedar pressure insoles were considered as a validated reference for this study. Subjects wore the pressure insoles embedded in custom-made shoes (Fig. 1). One inertial measurement unit (IMU) consisting of 3D gyroscopes and 3D accelerometers was installed on the forefoot over the bases of first and second metatarsals, such that one gyroscope, referred to as pitch, was aligned to foot's sagittal plane (Fig. 1). The IMU was connected to a portable data-logger (Physilog, BioAGM, CH) with an internal low-pass analog filter (17 Hz). Both pressure insoles and IMU devices recorded signals synchronously at 200 Hz.

Fig. 1. Sensor configuration worn by a subject with inertial measurement unit (IMU) fixed on forefoot and pressure-insoles (reference system) beneath the foot.

2.2. Temporal events detection

Stance phase is the period between initial contact, referred to as Heel-Strike (HS), and terminal contact, referred as Toe-Off (TO). Additionally, stance encapsulates the instant where toes touch the ground and make the foot land flat, referred as Toe-Strike (TS), and the instant where the heel rises from the ground, referred as Heel-Off (HO). {HS, TS, HO, TO} are defined as the temporal events of stance (Fig. 2a).

2.2.1. Kinematic features from inertial sensors signals

During one stride, the two negative peaks of pitch angular velocity of shank are known to be robust approximate estimates of HS and TO on both healthy and patient populations [10,22]. Foot pitch angular velocity (Ω_p) shows similar negative peaks for HS and TO. Consequently, those peaks were detected and used to split gait trials into cycles and define limited time windows for further robust detection of the kinematic features. Candidate features for detecting HS and TO were identified by the minimum (MIN), maximum (MAX) and zero-crossing (ZERO) time sample of the three following signals: Ω_p, the norm of 3D accelerometer signal ($||A||$) and the derivative of 3D gyroscope signal norm ($||\Omega||'$), where $||X||$ is the Euclidian norm of vector X.

The phase between TS and HO, so-called foot-flat, is characterized by a lower amount of movement since the ground constrains the foot. So, candidate features for detecting TS and HO were identified by the first and last sample for which signals of $||\Omega||'$, Ω_p, and the absolute value of the derivative of accelerometer signal's norm ($|||A||'|$), were below a specific threshold. Signals norms were preferentially selected in order to be independent of IMU positioning. All these detection rules, and the six subsequent kinematic features extracted for each event are detailed in Table 1 and illustrated in Fig. 2b and c.

2.2.2. Reference force features from pressure insole signals

A foot frame was defined with its X-axis as the horizontal projection of vector from the great tuberosity of calcaneus to the head of second metatarsal, Y-axis to the left and Z-axis upwards. The foot was divided into two segments: hindfoot and forefoot, and the coordinates of the 99 sensor cells of the insole were determined. Sensors cells with X-coordinate lower than the midpoint between bony landmarks of the navicular and cuboid bones were assigned to hindfoot, while other sensor cells were assigned to forefoot. The vertical force exerted on each segment (F) was

Table 1
List of features and their differences among 3193 recorded gait cycles. Temporal events are detected based on signal from inertial sensors (k_1 to k_{24}) and pressure insoles (f_1 to f_4). Ω_p and $||\Omega||'$ correspond to the pitch angular velocity of the foot and the derivative of the norm of foot angular velocity. $||A||$ and $|||A||'|$ correspond to the norm of foot acceleration and its absolute derivative. F_h and F_f are the vertical force signals estimated on the hindfoot and forefoot segments. Minimum value of differences for each event is indicated in bold italic.

	Kinematic			Force			Difference (ms)			
	Signal	Rule	Feature	Signal	Rule	Feature	Mean	MAE	STD	MAD
Heel-Strike	Ω_p	MIN	k_1	F_h	>5% of BW	f_1	29	26	*8*	*6*
		0	k_2				−39	43	17	13
	$\|\|A\|\|$	MIN	k_3				*1*	*8*	13	9
		MAX	k_4				37	36	14	8
	$\|\|\Omega\|\|'$	MIN	k_5				36	43	32	18
		MAX	k_6				−6	12	13	10
Toe-Strike	$\|\|\Omega\|\|'$	< -0.02 rad/s^2	k_7	F_f	>5% of BW	f_2	74	73	52	42
		< -0.06 rad/s^2	k_8				24	44	52	39
	Ω_p	> -1 rad/s	k_9				−23	41	44	38
		> -2 rad/s	k_{10}				*−4*	*31*	*37*	*31*
	$\|\|\|A\|\|'\|$	< 0.05 m/s^3	k_{11}				75	74	49	36
		< 0.2 m/s^3	k_{12}				12	47	53	45
Heel-Off	$\|\|\Omega\|\|'$	> -0.02 rad/s^2	k_{13}	F_h	<5% of BW	f_3	*4*	*41*	54	40
		> -0.06 rad/s^2	k_{14}				60	73	66	50
	Ω_p	< -1 rad/s	k_{15}				76	81	*51*	*36*
		< -2 rad/s	k_{16}				121	130	63	45
	$\|\|\|A\|\|'\|$	> 0.05 m/s^3	k_{17}				113	125	87	61
		> 0.2 m/s^3	k_{18}				169	176	71	50
Toe-Off	Ω_p	MIN	k_{19}	F_f	<5% of BW	f_4	−33	35	14	11
		0	k_{20}				63	65	21	17
	$\|\|A\|\|$	MIN	k_{21}				−81	85	15	11
		MAX	k_{22}				*−3*	*11*	*13*	*9*
	$\|\|\Omega\|\|'$	MIN	k_{23}				5	22	22	21
		MAX	k_{24}				−70	71	18	12

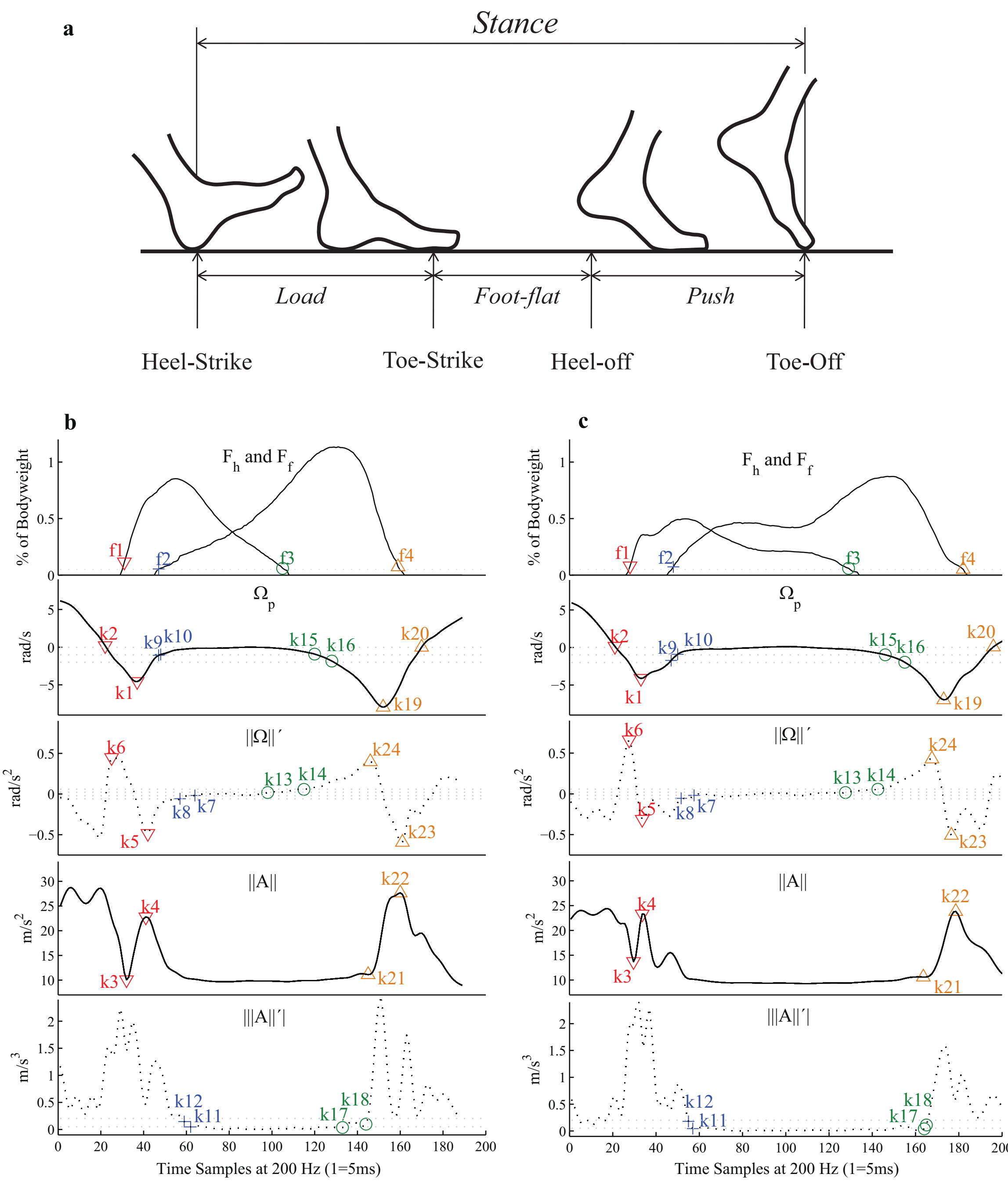

Fig. 2. (a) Temporal events during Stance and corresponding inner-stance phases (in italic). (b) Kinematic and force signals with the detected features (as listed in Table 1) at Heel-strike (▽), Toe-Strike (+), Heel-Off (o), and Toe-Off (△), showed for one typical gait cycle of a healthy subject and a c) subject with ankle osteoarthritis.

calculated based on pressure (P) and sensor cell area (A):

$$F = \sum_{j}^{J} P_j A_j \tag{1}$$

where j is the sensor cell index and J the set of segment cells. For the vertical force signal on hindfoot (F_h) and forefoot (F_f) segments, a threshold of 5% of bodyweight (BW) was used to detect the time of each segment's contact with the ground. HS (respectively TS) was detected on the rising of F_h, (respectively F_f), whereas HO (respectively TO) was detected on the lowering of F_h (respectively F_f). Those four force features (f_1 to f_4) constituted the reference values for temporal events (Table 1).

2.3. Inner-stance phases and foot-flat estimation

Based on detected temporal events, stance and inner-stance phases can be objectively quantified at each gait cycle. Thereby, the duration of stance was computed as:

$$\text{Stance} = t(\text{TO}) - t(\text{HS}) \tag{2}$$

where $t(\ )$ is the occurrence instant of the event. Subsequently, the duration of the three inner-stance phases composing Stance, namely loading response (Load), foot-flat and push-off (Push) were computed as:

$$\text{Load} = t(\text{TS}) - t(\text{HS}) \tag{3}$$

$$\text{Foot-flat} = t(\text{HO}) - t(\text{TS}) \tag{4}$$

$$\text{Push} = t(\text{TO}) - t(\text{HO}) \tag{5}$$

Absolute values were calculated in milliseconds, and relative values, Load_R, Foot-flat_R and Push_R, were expressed as a percentage of Stance.

2.4. Measurement protocol

Both healthy subjects and patients with different degrees of ankle disease were considered to test the proposed method's performance. In total, 42 subjects participated in this study: 10 healthy subjects (HY), 12 patients with ankle osteoarthritis (AO), 11 patients treated by total ankle replacement (TAR) and nine patients treated by ankle arthrodesis (AA). Both measurement systems were installed on subjects, on the affected foot for patients, and they were asked to walk at self-selected speed in a hospital corridor for two trials of 50 m. The Foot Function Index (FFI) and the American Orthopedic Foot and Ankle Society scale for ankle-hindfoot (AOFAS) were registered to evaluate the degree of ankle disease and illustrate the outcome of inner-stance phases (Table 3). The local ethics committee approved the experimental protocol and the subjects gave their informed consent prior to testing.

2.5. Statistical analysis

2.5.1. Temporal parameters validation

To compare the temporal event detection ability of the proposed system against reference, accuracy (Mean) and precision (STD) were calculated on the data sets of time differences between kinematic and force features at each gait cycle. The median absolute deviation (MAD), as a measure of statistical dispersion, and the mean absolute error (MAE), were also computed. The set of best kinematic features obtained using IMU was finally evaluated for reliability by computing Intraclass correlation coefficients ICC (1,1). Furthermore, the mean and standard deviation of the sets of differences between inner-stance phases' estimates from both the proposed and reference system were computed in each subjects' group.

2.5.2. Comparisons of subject groups

The median and interquartile range (IQR) were estimated for inner-stance phases measured by inertial sensors, clinical scores, and physical characteristics of each population. The results of each patient group (AO, TAR, and AA) were compared to the results of the healthy group (HY) using the Wilcoxon rank-sum test. Rank-sum test, as a robust non-parametric test, was chosen for pair-wise comparisons since population sizes were small and all metrics did not have a normal distribution, and possibly included outliers.

3. Results

3.1. Temporal events detection

After discarding the three first and last gait cycles of each trial from all tested subjects, a total of 3193 gait cycles were recorded and analyzed. Fig. 2 shows typical samples of recorded signals during stance with detected features on kinematic and force signals.

Table 1 summarizes the differences between kinematic features and reference force features for each event. For HS, k_1, detected at Ω_p minimum peak showed the best precision (8 ms), while the best accuracy (−2 ms) was obtained with k_3, at the minimum peak of $||A||$. For TS, the best results were obtained with k_{10}, detected at low Ω_p, showing an accuracy ± precision of -8 ± 39 ms; k_{12}, detected at low $|||A||'|$, showed a better accuracy (2 ms) but a bigger MAE than k_{10} (respectively 47 ms and 31 ms). For HO, the best accuracy (4 ms) was obtained with k_{13}, at low $||\Omega||'$, while the best precision (46 ms) was obtained with k_{15} extracted at low Ω_p. Finally for TO, the best results were obtained for k_{22}, detected at the maximum peak of $||A||$, showing an accuracy ± precision of -6 ± 12 ms.

According to Table 1, the optimal set of kinematic features for detecting {HS, TS, HO, TO}, was obtained with the set of rules {k_3, k_{10}, k_{13}, k_{22}}. For this set of rules, Coefficients of Intraclass correlation were calculated and show fair-to-good reliability for {HS, TS, HO} with ICC (1,1) of respectively {0.72, 0.51, 0.74}, and excellent reliability for TO with ICC (1,1) of 0.97.

3.2. Inner-stance phases estimation

Using the optimal kinematic features set from temporal events' detection, inner-stance phases were computed. Median values over all gait cycles were then calculated for each subject and compared to the reference system (Table 2). Globally, the average (mean ± STD) error was -3 ± 4 ms for Stance, -1 ± 10 ms for Load, 19 ± 14 ms for Foot-flat and -16 ± 13 ms for Push phases. Relative limit of agreement intervals [23], computed as average difference ±1.96 standard deviation in percentage of Stance were −0.6% to 1.8% for Load, 1.3% to 4.7% for Foot-flat and −3.6% to 0% for Push, showing good agreement between IMU and reference system.

3.3. Group comparisons

Table 3 presents median ± IQR of physical characteristics, clinical scores, and inner-stance phases obtained for the four populations. No significant differences were observed for age and height. All three patient groups showed significant difference ($p < 0.01$) with the healthy group for both clinical scores. The healthy group showed significantly shorter Stance compared to all patient groups, shorter Load compared to AO and AA, and shorter Foot-flat compared to TAR and AA. This is also qualitatively illustrated in the typical example given in Fig. 2 where a patient with ankle osteoarthritis (Fig. 2c) showed a longer foot-flat than a healthy subject (Fig. 2b). Although a tendency for longer Push_R and shorter Foot-flat_R in healthy subjects was observed, it was not significant.

4. Discussion

In this study, we showed that main temporal events during stance phase can be detected precisely and accurately using a single IMU attached to the foot. Based on these events, the corresponding inner-stance phases were computed to estimate loading response, foot-flat and push-off durations in normal and pathological gait. These metrics give promising perspective in ambulatory gait analysis, through the analysis of stance phase composition between more active (Load and Push) and passive (Foot-flat) periods. They allow quantitative analysis of the different temporal strategies among healthy subjects and patients with gait disorder.

Table 2
Mean (Accuracy) and STD (Precision) of difference between inner-stance phases obtained from inertial sensors system with the set of kinematic features {k_3,k_{10},k_{13},k_{22}} for {HS,TS,HO,TO}, and reference system in different groups of subjects (HY: Healthy, AO: with ankle osteoarthritis, TAR: after total ankle replacement, AA: after ankle arthrodesis).

		AO		AA		HY		TAR		ALL	
Phase	Difference	Mean	STD	Mean	STD	Mean	STD	Mean	STD	Mean	STD
Stance	ms	−5	5	−3	3	−2	2	−3	2	−3	4
Load	ms	12	11	−8	4	−3	4	−3	3	1	10
	% of Stance	1.9	1.3	−0.6	0.3	0.0	0.6	0.6	0.1	0.6	1.2
Foot-flat	ms	15	25	16	6	25	3	21	3	19	14
	% of Stance	3.2	3.1	2.5	0.6	3.3	0.4	3.0	0.2	3.0	1.7
Push	ms	−29	18	−7	3	−12	2	−14	2	−16	13
	% of Stance	−3.7	2.4	−0.5	0.3	−1.1	0.1	−1.2	0.2	−1.8	1.8

Table 3
Physical characteristics, clinical scores (FFI and AOFAS) and inner-stance phases durations in different groups of subjects (HY: Healthy, AO: with ankle osteoarthritis, TAR: after total ankle replacement, AA: after ankle arthrodesis) presented as median (IQR). Significant differences with healthy subjects are indicated with *(p-value <0.05) and **(p-value <0.01).

		HY	AO	TAR	AA
Physical characteristics	Age (years)	59.0(27.0)	60.5(17.0)	67.0(20.3)	65.0(13.0)
	Height (cm)	166.0(13.0)	169.0(9.0)	170.0(9.0)	177.0(11.3)
	Weight (kg)	66.6(12.6)	79.4(22.4)	82.6(12.5)**	87.7(9.2)**
	Sex	3 M,7 F	10 M,4 F	8 M,3 F	8 M,1 F
FFI	Total	0(0)	45.8(22.0)**	8.9(18.7)**	6.8(24.7)**
	Pain	0(0)	55.0(26.1)**	10.6(18.2)**	5.4(48.9)**
	Disability	0(0)	51.4(33.3)**	10.1(17.8)**	17.1(19.5)**
	Activity	0(0)	17.6(29.2)**	10.0(18.6)**	5.0(15.7)**
AOFAS	Total	100(0)	46.0(18.8)**	78.0(8.0)**	67.0(26.0)**
	Pain	40(0)	20.0(20.0)**	30.0(0.0)**	30.0(25.0)*
	Function	50(0)	28.0(12.5)**	38(7.8)**	30.0(6.3)**
	Alignment	10(0)	5.0(5.0)**	10.0(0.0)	10.0(0.0)
Stance (s)		0.60(0.06)	0.68(0.05)**	0.69(0.04)**	0.71(0.04)**
Load (s)		0.09(0.01)	0.11(0.03)*	0.09(0.04)	0.12(0.03)**
Load_R (%)		13.45(3.67)	15.55(5.63)	13.28(4.49)	15.20(4.29)
Foot-flat (s)		0.27(0.10)	0.32(0.09)	0.33(0.10)*	0.33(0.04)*
Foot-flat_R (%)		44.07(9.2)	46.02(10.27)	49.82(14.89)	49.21(5.31)
Push (s)		0.24(0.03)	0.27(0.05)	0.24(0.06)	0.26(0.04)
Push_R(%)		39.94(7.03)	38.09(7.45)	36.85(12.59)	36.73(6.76)

4.1. Technical validity

The comparison between features obtained from foot-worn inertial sensors and reference systems showed a high accuracy and precision for detecting HS and TO, and slightly lower but acceptable performance for detecting TS and HO. Our study revealed an important limitation of previous methods for HS and TO detection based on negative peaks of Ω_p, which showed a systematic bias leading to an underestimation of stance phase duration. Additionally, the proposed method showed smaller errors than results reported in previous studies using vertical foot velocity (16 ± 15 ms for HS and 9 ± 15 ms for TO [16]), or shank angular velocity (-8.7 ± -12.5 ms for HS and -2.9 ± 26.8 ms for TO [22]). Moreover, we observed that foot angular velocity signals were particularly useful for detecting TS and HO events (i.e., Foot-flat), whereas accelerometers provided better results for detecting HS and TO events (i.e., Stance).

Although the pressure insole was validated for force measurement, errors in location measurement of bony landmarks (the midpoint between navicular and cuboid bones) used to assign sensor cells to foot segments may influence the detected TS and HO moments. However, the dispersions of landmarks' location measurement errors have been shown to be much smaller than sensor cells' length [24]. Advantageously, the detection of TS and HO moment using IMU is not affected by this source of error.

Although threshold values were selected empirically to detect TS and HO, limited movement and low-pass filtering of signal prevented sensitivity to signal artifacts. Proposed thresholds were robust in all healthy subjects and patients with ankle dysfunction. Still, as inertial sensors can present a bias due to extrinsic factors such as temperature and humidity, the threshold values given in Table 1, except for the detection of minimal and maximal values, might require some tuning. Finally, the use of adaptive thresholds such as those proposed recently [25], could further enhance the detection's robustness.

By using the norm of signals, the detection algorithm is less sensitive to misalignment of the sensors relative to the foot, making it more repeatable without a specific sensor positioning, and generally applicable to any other foot-mounted IMU.

The use of single sensor configurations on the foot is also possible since we have proposed and validated the detection of temporal events using only kinematic features of one gyroscope around pitch ($k_{\{1,2,9,10,15,16,19,20\}}$), or only the use of a single 3D accelerometer ($k_{\{3,4,11,12,17,18,21,22\}}$).

4.2. Clinical applications

Since it was technically validated, the proposed system can then be miniaturized and integrated in the footwear as a fully wearable device and be used in clinical evaluations. This study also investigated the proposed system's clinical suitability.

Inner-stance phases results obtained for healthy subjects were in agreement with reference normative values reported [2], with 16.7% for Loading Response, 33.3% for Mid-stance, and 50% for the sum of Terminal Stance and Pre-Swing in percentage of Stance, where we found respectively $13 \pm 4\%$, $44 \pm 9\%$ and $40 \pm 7\%$ for Load_R, Foot-flat_R and Push_R. The slight differences are due to the subdivision of stance into three phases rather than four.

Similar performances were obtained for estimating inner-stance phases among the different subject groups, showing that the selected kinematic features were robust to the various healthy and pathological gait patterns recorded. Load, foot-flat, and push durations showed different tendencies in patient groups compared to healthy subjects. However, a dedicated clinical protocol with higher sample size is needed to confirm the clinical significance of these parameters.

Detection algorithms proposed in this paper were based on simple rules that could be further implemented for real-time rehabilitation applications, where precise and accurate triggers of stimulation sequences during walking are needed [13]. Finally, the presented method can be combined with other methods that estimate spatial gait parameters using IMU [26] to provide a simple, wearable, and reliable tool for objective and quantitative evaluation of both spatial and temporal gait parameters. The clinical utility of inner-stance phases should be further confirmed in other populations, and particularly in subjects whose gait is characterized by a longer foot-flat, such as elderly at risk of fall, or people with early phases of Parkinson's disease. Nevertheless, the application to other severe pathological gait patterns, such as those characterized by toe landing at initial contact due to foot-drop after stroke [27] or increased tone (without foot-flat during stance) would require further investigation.

Acknowledgments

The work was supported by ‘‘Fonds National Suisse de la Recherche Scientifique’’, grant no. 3200B0-120422/1. The authors would also express their gratitude to Dr. Brigitte M. Jolles for her advice, and to Mr. Pascal Morel and Mr. Jean Gramiger for their help and assistance in measurements.

Conflict of interest statement

Each of the authors discloses any financial and personal relationships with other people or organizations that could inappropriately influence (bias) this work.

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