Medical Engineering & Physics 32 (2010) 545–552

Contents lists available at ScienceDirect

Medical Engineering & Physics

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

Review

Methods for gait event detection and analysis in ambulatory systems

Jan Rueterbories[a,*], Erika G. Spaich[a], Birgit Larsen[b], Ole K. Andersen[a]

[a] Center for Sensory–Motor Interaction (SMI), Department of Health Science and Technology, 9220 Aalborg, Denmark
[b] Neurodan A/S, 9200 Aalborg, Denmark

ARTICLE INFO

Article history:
Received 4 January 2010
Received in revised form 29 March 2010
Accepted 30 March 2010

Gait detection
Gait analysis
Gait events
Ambulatory measurements
Accelerometer
FSR
Gyroscope

ABSTRACT

After stroke, hemiparesis is a common problem resulting in very individual needs for walking assistance. Often patients suffer from foot drop, i.e. inability to lift the foot from the ground during the swing phase of walking. Functional electrical stimulation is commonly used to correct foot drop. For all supporting stimulation devices, it is vital to adequately detect the gait events, which is traditionally obtained by a foot switch placed under the heel. To investigate present methods of gait analysis and detection for use in ambulatory rehabilitation systems, we carried out a meta-analysis on research studies. We found various sensors and sensor combinations capable of analyzing gait in ambulatory settings, ranging form simple force based binary switches to complex setups involving multiple inertial sensors and advanced algorithms. However additional effort is needed to minimize donning/doffing efforts, to overcome cosmetical aspects, and to implement those systems into closed loop ambulatory devices.

Contents

1. Introduction

Walking is one of the most common human physical activities and plays an important role in our daily activities. It can be performed in a variety of ways and directions and is furthermore a highly energy-efficient method of locomotion. For most people walking is fully subconscious and requires no thought. This may change following a stroke which may result in patient-specific needs for rehabilitation of locomotion. Often, patients have problems to achieve sufficient hip flexion and/or suffer from foot drop, i.e. they are unable to lift the foot from the ground during the swing phase of gait. Restoration of walking can be supported by functional electrical stimulation (FES) which has become an accepted rehabilitation method [1].

One of the first systems to correct foot drop was successfully developed in 1961 [2] and many others followed in the years after (see review by Lyons et al. [1]). For a successful application, it is essential to receive information from external sensors to control the stimulation. Nowadays, this input data is obtained by electronic sensors that measure various parameters during the gait cycle.

The term *gait* is used to describe the way of walking and consists of consecutive gait cycles. During each gait cycle a sequence of events take place that mark the transitions from one gait phase

* Corresponding author. Tel.: +45 9940 7572.
E-mail address: jrut@hst.aau.dk (J. Rueterbories).

1350-4533/$ – see front matter © 2010 IPEM. Published by Elsevier Ltd. All rights reserved.
doi:10.1016/j.medengphy.2010.03.007

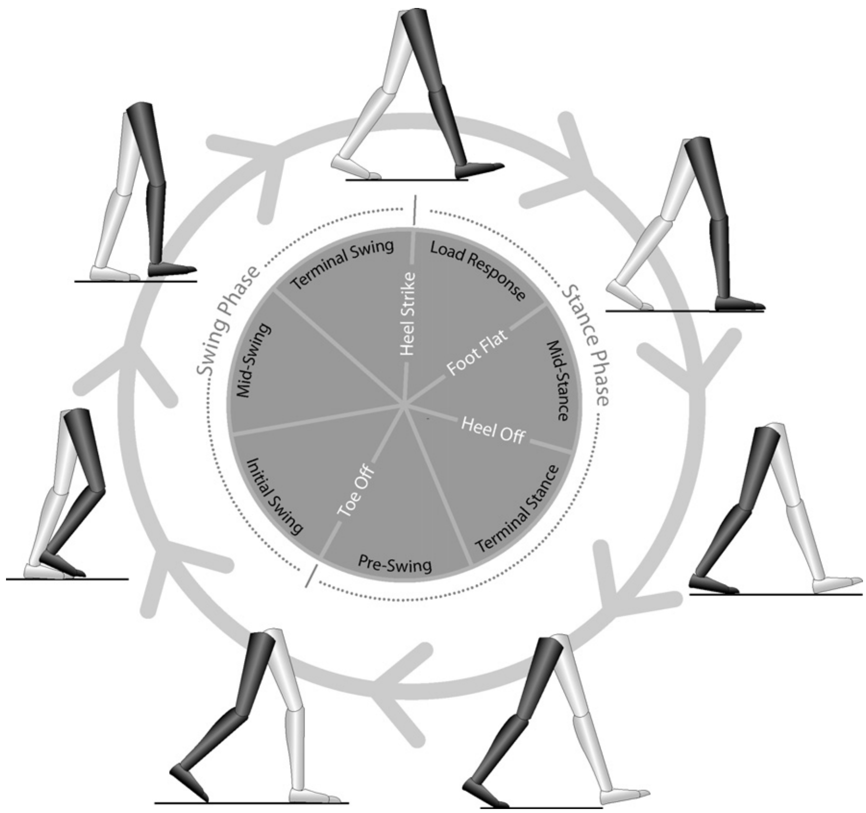


Fig. 1. Different phases during gait on a normalized time scale after Whittle [47].

to another. Fig. 1 illustrates gait events and phases on a normalized timescale. A normal gait cycle ends and begins by definition with the *heel strike* [3,4]. This event is the initial ground contact of the leading limb during normal walking. This is also the beginning of the *load response phase* during which the leading limb takes over the body weight by placing the whole foot on the ground. The events *heel strike* and *foot flat* are characterized by a rapid loading of the limb. During the *double support phase* both feet have ground contact and the walker is most stable. In the following *mid-stance* phase, the body is moved forward and the opposite limb is in the swing phase. This is a position where the walker is least stable due to the small base of support and the relatively high center of gravity. The event *heel off*, where ground contact of the heel is lost, indicates the transition from the *mid-stance* phase to the *terminal stance* phase. During *terminal stance*, the body is propelled forward until the *pre-swing* phase starts. This propulsive movement causes the final *toe-off* event where the contact between toes and floor is lost and the *swing* phase begins. During the swing phase the swinging limb moves in front of the stance limb leading to a forward progression. The *swing* phase itself is divided into the sub phases of *initial swing* where the limb is accelerated forward, the *mid-swing* phase in which the limb passes the opposite stance limb, and the *terminal swing* phase where the limb is decelerated in preparation for the *heel strike* which will terminate the *swing* phase.

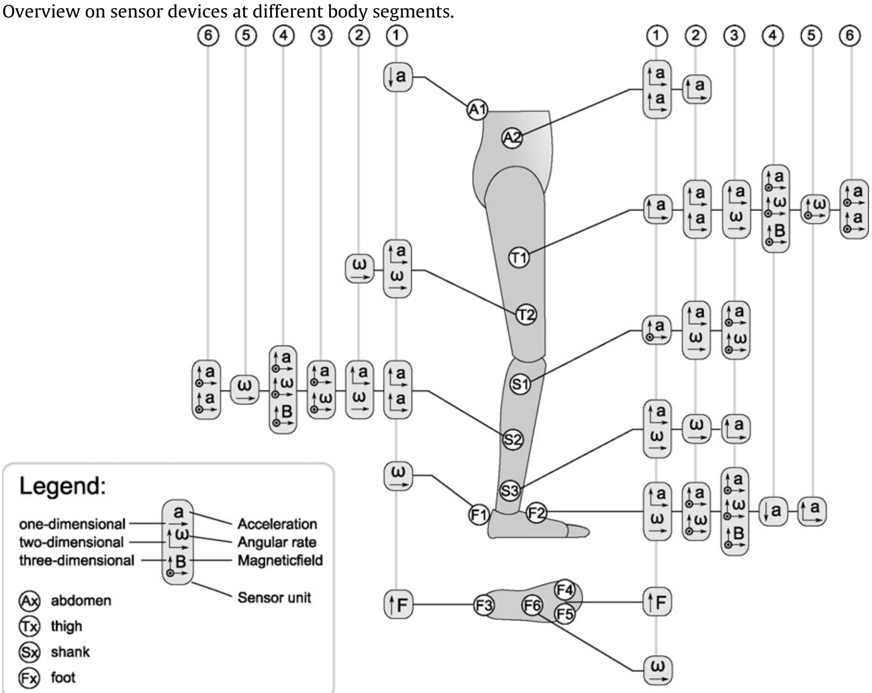
The definition of individual gait events and phases is an essential starting point for nearly all aspects of gait analysis and restoration. Hence gait is described in terms of temporal and spatial components. The temporal components are those periods of time during which events take place. The spatial components refer to the position and orientation of limbs and joints. When performing gait analysis, it is important to consider both aspects, since a disease or trauma can affect the gait spatial and temporal components independently [4].

The physical ways to analyze gait are based on *kinematics* and *kinetics*. The term *kinematics* is used to describe movement not taking into account the forces that cause the movement. One approach to measure movement is the direct measurement of linear and angular displacements provided by joint angles, limb velocities or accelerations. A second established method is the indirect measurement of movement with cameras or tracking systems. Such systems capture the trajectory of markers attached to the body during a movement, and determine hereby the desired quantities by inverse kinematics. A markerless system is currently under development by BioMotionLab [5]. The term *kinetic* describes the study of forces and moments that cause a movement. Those are for example gravitational, ground reaction, other external forces, or forces produced by muscle contractions.

Feedback for a stimulation controller can be provided by sensors like goniometers, accelerometers, angular rate meters, inclinometers, or force sensitive resistors, that are suitable for determining joint angles, body-segment acceleration, tilt angle, and times of foot contact, respectively. Many different sensor configurations have been used for gait cycle analysis or detection in humans. However, research in this field is an ongoing activity, since present gait detection systems or methods still have limitations regarding usability and applicability when used with portable devices for gait restoration.

In this review, we analyze existing methods for monitoring gait and detecting gait events that could be used in an ambulatory rehabilitation system. We discuss the aspects of sensor setup, signal processing, and evaluation of performance with focus on ambulatory systems. For this purpose we performed a meta analysis on research studies on measurement methods, matching one or more of the following search keys: gait kinematics, gait detection, gait analysis, gait events, locomotion, ambulatory measurements, wearable sensors, force sensitive resistors, accelerometer, gyroscope.

Table 1
Overview on sensor devices at different body segments.



Measurement Position	References, position by horizontal alignment					
	1	2	3	4	5	6
A1	[11]					
A2	[25]	[27]				
T1	[12,27]	[23,25,26]	[17]	[16,34,38]	[9]	[28]
T2	[29,32,33]	[15,14] [a]				
S1	[16]	[32,29]	[36]			
S2	[24–26]	[17]	[34]	[16,37,38] [a]	[9,13,39]	[28]
S3	[31,33]	[14] [a]	[27]			
F1	[10]					
F2	[30,31,33]	[34,35]	[37]	[39,40]	[27]	
F3	[6–8,10,18,27]					
F4	[6,7,10,18,27]					
F5	[6–8,10,18,27]					
F6	[18]					

[a] Attached to left and right segments Ax, Tx, Sx, and Fx describe sensor position at abdomen, thigh shank and foot, respectively.

2. Methods to measure gait

Different sensor configurations have been used for gait detection in ambulatory settings comprising single and multiple sensor configurations as a trade off between donning/doffing, cosmetically considerations, and detection reliability. Table 1 provides an overview on the sensors and positions used in the reviewed systems.

2.1. Force based measurements

A number of methods for gait measurements are based on the force exerted by the body to the ground. The only possible position for sensors is therefore between the sole of the foot and the ground. Available transducers are designed as mechanical, load dependent switches or force sensitive resistors (FSRs), capacitive or piezoelectric elements.

FES systems for drop foot correction have traditionally used foot switches [1]. Early systems used pure mechanical micro-switches, while recent system are based on threshold values obtained from one or more FSR transducers placed under the heel, at the ball of the foot or under the hallux (Table 1) [6,1]. An alternative to switches is the use of force sensitive insoles composed of a matrix of sensors covering the entire sole of the foot [7]. A second alternative is based on a pressure sensor connected to a small tube, glued to the outer perimeter of the sole of the shoe [8]. Changes of ground reaction forces cause pneumatic changes in the tube that were monitored by a pressure sensor.

While a single heel switch provides information about the events *heel off* and *heel strike*, additional sensors (placed under the ball of the foot, the hallux, or as an insole) provide information about the events *foot flat* and *toe-off* [1,7]. This is needed in case the pathological gait cycle does not start with heel contact.

The accuracy and reliability of such systems depend mainly on mechanical wear, as the contact sensors are exposed to repetitive force changes up to 2.2 kN [1]. A FES system using a FSR-insole was tested on three paraplegic subjects and showed a detection error for *heel strike* and *toe-off* events of 3.2–12% of the gait cycle duration, in 90% of the detected cycles [7]. The validation of the pressure-based method was achieved by comparing the output of the device with the vertical force signal from a force plate. Both signals were similar, whereas the progression of the pressure signal lagged the force plate signals with a mean delay of 0.3–3% of gait cycle [7].

Force based event detection either through force plates or sensors attached to the foot, is still the reference method for determining the accuracy of gait event detection in other systems. The sensors attached to the foot seem to be the natural choice for ambulant systems. They provide satisfactory results for normal walking and require only little signal conditioning. However, force sensitive sensors have some disadvantages. For instance positioning the sensors on subjects with abnormal gait is difficult [9] and distinguishing load changes generated during walking from those caused by weight shifting is not possible [10]. Further, shuffling gait reduces the reliability [11] of event detection, and finally the cosmetic acceptance and durability limit the applicability [12].

2.2. Angular rate measurement

Angular rate sensors (gyroscopes) provide measurements of angular displacement. In principle, these sensors measure the Coriolis force, i.e. the response to rotation of a moving mass. The Coriolis force that arises in a rotating reference frame is proportional to the angular rate of rotation. All studies using this sensor principle used the same type of one-dimensional angular rate sensor in a single sensor solution [13–15] or triple sensor [9] configuration. In all cases, the data were processed offline.

Various sensor positions have been tested: thigh [15], shank [13], shank and thigh [14], or both shanks and one thigh [9].

From the sensor signals, the knee-joint angle [14] or the hip joint angle [15] has been calculated by integration of the angular velocity. Different means were undertaken to compensate for the drift effect which is a common problem caused by signal offset. This offset leads to an angular drift following the signal integration. Another approach is the estimation of *heel off* and *heel strike* by wavelet processing of angular rate signals [9,13].

Validation of the systems was obtained with focus on the specific applications. Spatial parameters like stride length were determined with an accuracy of $\pm$15% in amputees with prosthetics under laboratory conditions, while in hemiplegic patients, an error of 25% was reported [15]. Measurements of the knee angle in healthy volunteers had a 93% correlation with measurements obtained from a motion capture system (Vicon, UK) under laboratory conditions [14]. Temporal parameters (heel strike and toe-off) correlated 95% with the FSR reference obtained from young and elderly volunteers under laboratory conditions [9]. In another study, the events heel strike, foot flat, and toe-off, and stair ascending and descending were detected with an overall reliability $> 94\%$ in elderly people who had limited mobility [13].

A great advantage of gyroscopic motion analysis is the fact that it is not affected by gravitation [16], i.e. the measured angular velocity is not superimposed on a gravitation component like accelerometer measurements. Further, vibration of sensors during heel strike does not affect the gyroscope output [17]. Gyroscopes are less sensitive to positioning on the body due to their measurement principle. Sensors can be placed anywhere along the same plane on a body segment giving almost an identical signal output [14]. Importantly, movements in other planes are not captured, e.g. change in direction [14].

2.2.1. Angular rate and force measurements in combination

A system composed of three FSR sensors detecting vertical load combined with a gyroscope that measured the rotational velocity of the foot in the sagittal plane has been investigated regarding detection of *heel off* and *heel strike* events in a FES foot drop system [10].

The FSRs were placed under the heel and the first and fourth metatarsal heads. The uni-axial gyroscope was attached to the heel.

The system was capable of detecting the *stance* and *swing* phases in real-time [10].

The validation was performed under laboratory conditions with able-bodied individuals and subjects with walking impairments. The validation included several aspects: validation against a motion captures system, walking and non-walking tasks as well as a reliability test. The detection rate for both groups was above 99% for the abled-bodied subjects and above 96% for the subjects with impaired gait [10]. The system was later embedded into an insole used with a FES foot drop correction system, and tested regarding the quality of stimulation [18].

The most important advantage of this system is its reliability and robustness. The system did not generate false triggers, e.g. during standing, caused by weight shifting from one leg to the other, standing up, or sitting down [18]. The overall performance and implementation into an insole support a practical application. However, the use of FSRs which are known to fail with time due to mechanical wear and the necessary wire connection from the insole to the stimulator might be a disadvantage of this system. Announced future developments may address these issues [18].

2.3. Accelerometry

The technology of MEMS (Micro-Electro-Mechanical Systems) allow the development of miniature, low powered, accelerometer devices that are suitable for monitoring over ground walking [19]. Therefore, accelerometers are an ideal choice to analyze locomotion [20]. The physical mechanisms underlying MEMS accelerometers are based on a miniature mass-spring system. During motion, the sensing element moves with its inertia relative to a fixed base inside the part which can be detected and transformed into an electric signal.

During the last decade, body mounted inertial sensors have been used to obtain kinematic values like the inclination angles of the shank and thigh or the knee angle [21–23]. These data can be derived by integration of angular acceleration or calculated by double sensor difference measurements [23]. Common for these measurement systems was the use of devices containing several inertial sensors attached typically to body segments involved in gait, e.g. foot, shank, thigh, or trunk.

The sensor units measured accelerations in two [11,12,24–26] and three dimensions [16,23,27]. Either composed by several single axis sensors [16,24,25] or dual axis sensors [11,12,26,28], or triaxial sensors [23,27].

In order to measure rotational and translational acceleration, typical sensor positions were shank [16,23,24], thigh [12], shank and thigh [25,28], shank and thigh and pelvis [26], foot and shank and thigh and pelvis [27], or trunk [11].

Processing of accelerometer data has been performed offline as well as in real-time [12,16,23,25,27,28]. Offline processing comprises time analysis of vertical acceleration in the sagittal plane from negative to positive changes and from positive to negative changes which have been correlated to the *heel strike* and *heel off* events, respectively [11]. Another approach showed the gait events *toe-off*, *initial swing* phase, *terminal swing* phase, and *heel strike* based on the angular and translational velocity data in the sagittal and coronal planes [24]. It has been shown (see Section 3) that real-time systems are capable of detecting *load response*, *mid-stance*, *terminal stance*, and *pre-swing* [16,28], or *heel strike* and *heel off* [12]. Lower extremity angles have also been calculated online, based on a rigid-body model [25] or an artificial neural network [27] or by a double sensor difference based algorithm [23]. The accuracy of all methods using accelerometers was reported as sufficient in regard to their specific applications: to substitute goniometers, an accuracy of $\pm$0.01 to $\pm$0.05 rad, depending on the measured joint, has been obtained under laboratory conditions [25,26]. Inter-joint angles have been obtained with a cross correlation of 0.96–0.99, depending on the joint [27]. Two-dimensional body-segment angles were measured with an error of 4.7–8.9 depending on the walking speed in healthy subjects [23]. Accelerometers may also replace foot switches for detection of gait events. Compared to a heel FSR sensor, accelerometry may detect *heel strike* with less than 5% deviation in healthy subjects as well as in stroke patients [12]. The reliability for event detection ranges from 86 to 91% (healthy subjects, laboratory conditions, real-time system) [16] to 92.4–98.7% for normal and simulated hemiplegic walking trials (off-line analysis) [11].

In general, the use of accelerometers requires additional signal processing and means to compensate the influence of gravity. Further drift problems may occur with integration of the acceleration data [25]. Attachment of sensors is another source of imprecision due to the movement of muscles during walking, which appears as a high frequency error in the signals [25]. On the other hand, *heel off* is detectable even with poorly defined heel contact [16], or during shuffling gait [11]. The current generation of MEMS sensors is designed to satisfy industrial and military standards, in other words, they have low energy consumption, small sizes, low cost and high quality [19].

2.3.1. Accelerometry and angular rate measurements

Measuring angular rate can be achieved using one or more inertial sensors and a gyroscope fitted into one device that is typically attached to different body segments like foot, shank, thigh, or torso.

The measurement method is either based on a two-dimensional model in the sagittal plane [17,29–33] or on a three-dimensional model [34–36]. For two-dimensional models, the sensor units are comprised of one one-dimensional gyroscope and two one-dimensional inertial sensors [17], or alternatively one two-dimensional inertial sensor [29–33], that measures tangential and radial acceleration components. The three-dimensional systems typically used one three-dimensional inertial sensor and one three-dimensional gyroscope [34,35] or alternatively several uni- or dual axis sensors [36].

The sensor units were mounted either on one foot [30,35], shank and thigh [17,29,32,36], foot, shank and thigh [34,33], or both feet and shanks [31].

The recorded data was primarily processed offline [17,29,30,32,34], but also in real-time [31,35,36]. The processing provided information on segment acceleration and velocity [17,34], joint angle [29,35], and gait events like *heel strike* and *toe-off* [30,31], and *foot flat* [32].

Validation of systems containing different sensor types has been performed in healthy [17,29,30,34,33], hemiparetic [32], and spinal cord injured subjects [31,32,35]. The obtained data was verified either against a motion caption system (WATSMART^{TM} [34], Vicon^{TM} [17], Zebris Medical GmbH [29], or foot switches [30–32]. The mean error for the angular rate and acceleration measurements was reported to be below 7% [17] and the inter-joint angle could be derived with an error of less than 1.1° [29] or 1 rad [34] when compared to a motion capture system. Further, the walking speed was obtained with an error of less than 1.58% [30]. The detection of foot contact onset/offset was reported to be as accurate as that obtained from foot switches [31]. A timing correlation of 0.97 for *heel strike* and 0.95 for *toe-off* has been reported [32]. Further, the accuracy for the calculated stride length was determined with 3% compared to foot switches [35].

The additional use of gyroscopes reduced the error caused by accelerometer vibration during *heel strike* resulting in a detection accuracy of 93% compared to a video motion capture system [17]. Foot contact detection based on linear foot acceleration or angular foot velocity can correctly identify foot contact onset and offset in both normal and pathological gait patterns as accurate as foot switches [31]. However, gyroscopes suffer from the same drift problem as accelerometers.

2.3.2. Accelerometry, angular rate, and magnetic field measurements in combination

Measurement of the earth magnetic field vector by a magnetometer provides besides the earth gravity field a reference measure for body orientation. Opposite to accelerometry, the earth magnetic field signal is not affected by dynamic motions.

Custom made sensor devices composed of several uni [36] or dual-axis sensors [37] in combination with a magnetometer have been used, as well as a commercially available sensor unit (NEC Tokin) which provides rotation angles around its body axes as output [38]. This can be used for assessing body segment orientation whereby joint angles can be quantified.

Such devices have been fixed to the foot and shank of one leg [37], or to both shanks and thighs [38] where inter-joint angles were quantified. Sensor units have also been attached to both shanks via an orthosis in combination with a third unit worn on an abdominal belt for detecting gait phases [36].

The measurement systems were capable of determining 3D inter-joint angles via off-line data analysis [37] or in real-time [36,38], they were furthermore capable of detecting five gait phases (*loading response*, *mid-stance*, *terminal stance*, *pre-swing*, and *swing*) using acceleration measures and its first derivative [36]. *Heel strike* and *toe-off* have also been determined by identification of peaks in the shank rotation in the sagittal plane [38]. A model of a double pendulum can be used for calculation of stride length, and eventually, ranges of shank and thigh rotation [38].

All systems were only verified with able-bodied subjects under laboratory conditions. The stride length was estimated under five different stride conditions with an average error of 2.3% [38]. Ankle joint angles were measured in 12 defined, static angle positions; the reported error ranged from 0.55° to 4° depending on the direction of movement [37]. The reason for assessing the non-static behaviour is that the accuracy decreases if the rotation axis is not orthogonal to the reference vectors of gravity and magnetic field [37]. The detection accuracy for the swing phase, pre-swing phase and heel loading response was reported to be greater then 80% with a resolution of 40 ms and determined joint angles ranged from 3.2° to 3.8° [36].

The additional measurement of the earth magnetic field vector provides a second non-gravity affected reference which may increase the measurement accuracy.

2.3.3. Inclinometry

A tilt sensor that uses inertia can be used for detection of body tilt. It consists of an inertial element that senses gravity, like a liquid (electrolyte, magnetic fluid, or mercury) or a mass-spring system with solid elements. When the sensor tilts, gravity causes a relative movement of the inertial element which is then sensed as changes in resistance or capacitance. Tilt sensors have rarely been used for gait detection in FES systems. However, there have been attempts to use them as replacement for standard foot switches in drop foot correction systems [39,40].

The shank or the thigh were found to be suitable positions for attachment of tilt sensors; with the shank signal being superior because of better correlation to the *toe-off* and *heel strike* events [40].

Four different sensor types (an electrolytic, two magneto-resistant and a mercury based) were compared with regard to their step response, frequency content, and temperature characteristics. The magneto-resistive type was subsequently selected for a FES system for foot drop correction [39]. Recordings were obtained from six subjects with spinal cord injury or stroke with positive observations, however, detailed findings are not yet published [39]. In another study with the same setup, insufficient reliability was observed due to false triggering during non-walking tasks [10]. Tilt sensors also respond to accelerations, which is another inherent problem of this sensor type [39]. However, inclinometry is used in commercially available devices as it is not considered to be a complex method for detection of gait events [40,41].

3. Gait detections methods

The challenge of gait detection is to develop algorithms that determine gait events while the person is walking (real-time detection). Traditionally, this has primarily involved foot switches (mainly FSR and simple threshold processing) to detect foot contact events. Alternatives to foot switches have been broadly discussed in the previous sections, while ignoring the fact that detection algorithms are needed to obtain the desired gait events.

All published algorithms consist of a set of rules that identify certain characteristics of the gait measurements. The rules have been applied with the following approaches: functional analysis of raw [9,42–44] or derived [10] data, inductive learning techniques [7,12,16,45] or finite state machines [10,46].

Functional analysis comprises mathematical methods for curve sketching to extract features that correspond to or indicate certain gait phases or events. It has been shown that the first and second derivative of the vertical and horizontal accelerometer data from the foot allow detection of heel strike and toe-off by simple threshold rules which were processed in real-time [42,43]. A similar method was used to analyze the vertical velocity of the forefoot, searching for characteristic features corresponding to *toe-off* (maximum values of the time continuous accelerometer signal) and *heel strike* (local minimum of the time continuous accelerometer signal) [44] (Fig. 2, triangle markers). This algorithm was used for off-line analysis, however, a real-time implementation might be feasible also on small embedded systems since no advanced mathematical functions are needed. Wavelet transformation has been claimed to be suitable for gait event identification [9]. The *toe-off* and *heel strike* events consist of combined features, that can be located in the frequency and time domains which then are detected by wavelet analysis (Fig. 2, features are marked by circles).

Pre-processed data, for example a body-segment angle obtained through integration of an angular rate signal, has been used to detect gait using a finite state machine in real-time. The gait cycle was divided into four states and seven allowed transitions [10]. Every transition was then described by rules consisting of *if...then...* clauses and boolean algebra [10].

Inductive machine learning is a branch of artificial intelligence and comprises the design of algorithms that allow a system to learn by extracting rules and patterns out of a set of data. The resulting rule base is a compilation of the associations between input values (e.g. accelerometer data) and the output (gait phases). The associations are described in form of *if...then...* statements that are weighted according to their probability of truth. Different approaches for generating these rules have been tried in relation to online gait detection. For example, acceleration data was processed as input data to match foot contact data as desired output. The weights were calculated using the *Matlab*™ *Neural Network Toolbox* and afterward used in the implementation of the neural network on a micro-controller to detect *heel strike* and *heel off* in real-time [12]. A mutual information classifier has been used to produce a decision tree by maximizing the average mutual information gain at each step. The algorithm used the entropy as measure of information and calculated thresholds of hip- and knee-joint angles as well as foot contact forces used within a decision tree to provide real-time detection of foot contact events [45]. Commercial machine learning programs based on *Rough Sets*™ and Adaptive Logical Networks have been used with accelerometer data in a real-time setup [16]. A fuzzy logic rule base [7] has also been used to implement supervised

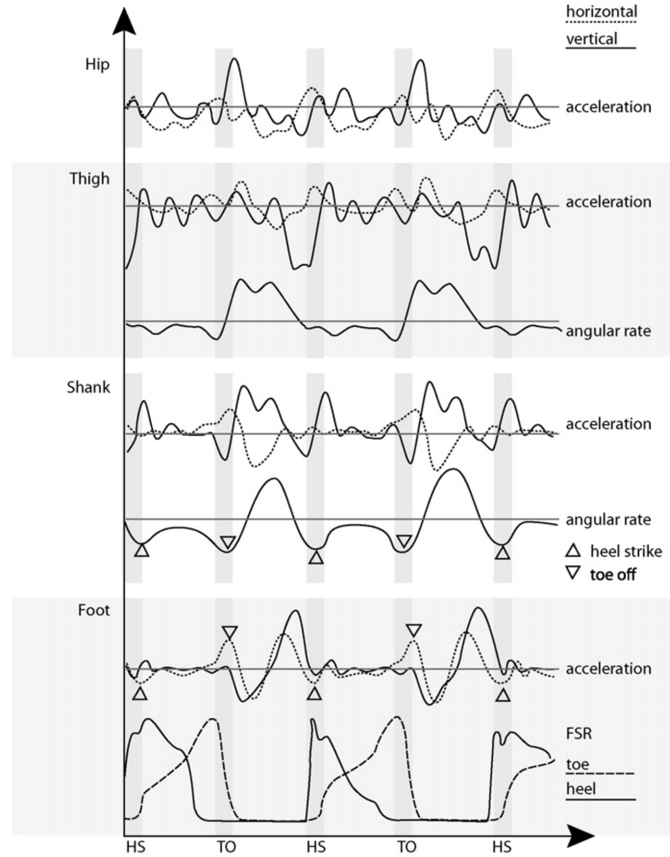


Fig. 2. This figure illustrates kinematic data of gyroscope and accelerometer at different body positions in relation to foot contact signals. The signals are a compilation from different papers and were biased to emphasize the features used by different algorithms. Acceleration data is shown in vertical and horizontal components of the sagittal plane.

machine learning in a real-time gait detection application based on data from insole FSRs.

4. Discussion

The questions addressed by the present review were: what are the current possibilities to detect gait events with an ambulatory system. The main finding is that possibilities for detecting gait are manifold and offer a wide range from simple to complex solutions (Table 2). Various sensors and sensor combinations are capable of providing relevant physical signals that allow analysis of gait in ambulatory settings. Accelerometers, [16,19–22,24,25] tend to be the most used sensors, often used in combination with gyroscopes [17,29–32,34–36]. Furthermore, sensor positioning seems less critical as placing the sensor at nearly any combination of foot, shank, thigh, and trunk of one or both legs are possible with appropriate signal processing. Several solutions are already part of FES systems for foot drop correction [10–12,16,27,35,40], while others were used solely for online or off-line gait analysis. However, additional processing effort is needed in order to implement the latter systems into closed loop ambulatory devices that improve reliability and minimize donning/doffing efforts and cosmetic considerations of the users.

Hence, the sensor devices should be small and light-weight and no wires should be needed in order to be worn comfortably and be cosmetically accepted. Light-weight coheres strongly with energy efficiency in battery powered devices. Modern microcontrollers and sensors are advertised as ‘low power’ devices, however this is relative. Gyroscopes tend to consume up to several hundred milliampere while accelerometers are in the range of a few microampere. Micro-controllers and other peripherals like radio chips easily add double digit milliampere to the current consumption. Therefore the hardware design has a great impact on usability and user acceptance and is to be chosen to use as few components as possible. The fixation of the sensor devices should be done in a way that facilitates easy donning and doffing, including a certain tolerance to the exact position, is provided. Further, it should ultimately provide the possibility to walk barefoot with the device, since this is part of life quality and perhaps culture [12]. However, the most important requirement for a successful use in an ambulatory rehabilitation system is a sufficient reliability in the detection of gait events during daily use. Detecting the trigger event with reliability between 70 and 90% is insufficient for practical applications [10].

Future work supporting gait detection should focus on the question of which body segment provides the richest information about the gait cycle, especially considering typical impaired gait patterns of different patient groups. This may also support the decision about the optimal sensor set and sensor position which remains still unclear. However inertial sensors in general are appropriate devices for monitoring gait. The challenge and most difficult task is therefore proper interpretation of the measured signals by an algorithm with low demands for CPU capacity and memory.

In conclusion, this review outline the latest research carried out on methods of gait analysis and event detection in relation to an ambulatory use. At the present time, there are many potential applications, but only a few which are suitable for daily ambulatory use. Most methods have been developed for use under labora-

Table 2
Overview on sensor configuration, obtained measures and involved subjects regarding their gait

Reference	Sensor device	Measurement of	Number of subjects	
			Healthy	Impaired
[11]		Foot contact events	4	–
[23,25]		Knee angle, segment angle	1,8	–
[27]		Inter-joint angles	6	–
[12]		Foot contact events	5	3
[26]		Inter-joint angles	1	–
[17]		Segment a and ω	10	–
[30]		Foot contact events	1	–
[34]		Segment displacement, a and ω	3	–
[16]		Knee angle	1	–
[38]		Gait events, spatial parameters	1	–
[28]		Gait events	3	10
[36]		Knee angle	3	–
[24]		Segment angular rate and position	3	–
[37]		3D inter-joint angle	2	–
[13]		Foot contact events	10	10
[9]		Foot contact events	20	–
[31]		Foot contact events	26	15
[14]		Knee-joint angle	1	–
[10]		Foot contact events	10	6
[35]		Foot orientation	–	1
[39]		Foot contact events	1	1
[40]		Foot contact events	–	1
[7]		Foot contact events	–	3
[8]		Foot contact events	1	–
[18]		Foot contact events	–	2

tory conditions causing design constraints which limit their clinical usability in ambulatory systems.

Conflict of interest

The authors state no conflict of interest.

References

[1] Lyons GM, R TS, Burridge JH, Wilcox DJ. A review of portable FES-based neural orthosis for the correction of drop foot. IEEE Transactions on Neural Systems and Rehabilitation Engineering 2002;10:260–79. doi:10.1109/TNSRE.2002.806832.

[2] Liberson W, Holmquest H, Scot D, Dow M. Functional electrotherapy: stimulation of the peroneal nerve synchronized with the swing phase of the gait of hemiplegic patients. Archives of Physical Medicine and Rehabilitation 1961;42:101–5.

[3] Perry J. Gait analysis: normal and pathological function. USA: Slack; 1992.

[4] Trew M, Everett T. Human movement—an introductory text. Elsevier Ltd.; 2005.

[5] Corazza S, Undermann LM, Chaudhari AM, Demattio T, Cobelli C, Andriacchi TP. A markerless motion capture system to study musculoskeletal biomechanics: visual hull and simulated annealing approach. Annals of Biomedical Engineering 2006;34:1019–29. doi:10.1007/s10439-006-9122-8.

[6] Strojnik P, Kralj A, Ursic I. Programmed six-channel electrical stimulator for complex stimulation of leg muscles during walking. IEEE Transactions on Biomedical Engineering BME-26 1979:112–6. doi:10.1109/TBME.1979.326520.

[7] Skelly MM, Chizeck HJ. Real-time gait event detection for paraplegic FES walking. IEEE Transactions on Neural Systems and Rehabilitation Engineering 2001;9(1):59–68. doi:10.1109/7333.918277.

[8] Nilsson J, Stokes VP, Thorstensson A. A new method to measure foot contact. Journal of Biomechanics 1985;18(8):625–7. doi:10.1016/0021-9290(85)90017-X.

[9] Aminian K, Najafi B, Bla C, Lezvraz PF, Robert P. Spatio-temporal parameters of gait measured by an ambulatory system using miniature gyroscopes. Journal of Biomechanics 2002;35:689–99. doi:10.1016/S0021-9290(02)00008-8.

[10] Pappas IP, Popovic MR, Keller T, Dietz V, Morari M. A reliable gait phase detection system. IEEE Transactions on Neural Systems and Rehabilitation Engineering 2001;9:113–25. doi:10.1109/7333.928571.

[11] Mansfield A, Lyons GM. The use of accelerometry to detect heel contact events for use as a sensor in FES based assisted walking. Medical Engineering and Physics 2003;25:879–85. doi:10.1016/S1350-4533(03)00116-4.

[12] Shimada Y, Ando S, Matsunaga T, Misawa A, Aizawa T, Shirahata T, et al. Clinical application of acceleration sensor to detect the swing phase of stroke gait in functional electrical stimulation. The Tohoku journal of Experimental Medicine 2005;207:197–202. doi:10.1111/j.1525-1403.2006.00074.x.

[13] Coley B, Najafi B, Paraschiv-Ionescu A, Aminian K. Stair climbing detection during daily physical activity using a miniature gyroscope. Gait & Posture 2005;22:287–95. doi:10.1016/j.gaitpost.2004.08.008.

[14] Tong K, Granat MH. A practical gait analysis system using gyroscopes. Medical Engineering and Physics 1999;21:87–94. doi:10.1016/S1350-4533(99)00030-2.

[15] Miyazaki S. Long-term unrestrained measurement of stride length and walking velocity utilizing a piezoelectric gyroscope. IEEE Transactions on Biomedical Engineering 1997;44:753–9. doi:10.1109/10.605434.

[16] Williamson R, Andrews BJ. Gait event detection for FES using accelerometers and supervised machine learning. IEEE Transactions on Rehabilitation Engineering 2000;8(3):312–9. doi:10.1109/86.867873.

[17] Mayagoitia RE, Nene AV, Veltink PH. Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems. Journal of Biomechanics 2002;35(4):537–42. doi:10.1016/S0021-9290(01)00231-7.

[18] Pappas I, Keller T, Mangold S, Popovic MR, Dietz V, Morari M. A reliable gyroscope-based gait-phase detection sensor embedded in a shoe insole. IEEE Journal of Sensors 2004;4(2):268–74. doi:10.1109/JSEN.2004.823671.

[19] Kavanagh JJ, Menz HB. Accelerometry: a technique for quantifying movement patterns during walking. Gait & Posture 2008;28:1–15. doi:10.1016/j.gaitpost.2007.10.010.

[20] Culhane KM, O'Connor M, Lyons D, Lyons GM. Accelerometers in rehabilitation medicine for older adults. Age and Ageing 2005;34:556–60. doi:10.1093/ageing/afi192.

[21] Crago PE, Chizeck HJ, Neuman MR, Hambrecht FT. Sensors for use with functional neuromuscular stimulation. IEEE Transactions on Biomedical Engineering 1986;33(2):256–68. doi:10.1109/TBME.1986.325808.

[22] Wong WY, Wong MS, Lo KH. Clinical applications of sensors for human posture and movement analysis: a review. Prosthetics and Orthotics International 2007;31(1):62–75. doi:10.1080/03093640600983949.

[23] Liu K, Liu T, Shibata K, Inoue Y, Zheng T, Novel approach to ambulatory assessment of human segmental orientation an a wearable sensor system, Journal of Biomechanics; doi:10.1016/j.jbiomech.2009.08.008.

[24] Morris JRW. Accelerometry—a technique for the measurement of human body movements. Journal of Biomechanics 1973;6:729–36. doi:10.1016/j.medengphy.2004.11.006.

[25] Willemsen ATM, van Alst J, Boom HBK. Real-time gait assessment utilizing a new way of accelerometry. Journal of Biomechanics 1990;23(8):859–63. doi:10.1016/0021-9290(90)90033-Y.

[26] Willemsen ATM, Friego C, Boom HBK. Lower extremity angle measurement with accelerometers—errors and sensitivity analysis. IEEE Transactions on Biomedical Engineering 1991;38(12):1186–93. doi:10.1109/10.137284.

[27] Duric MD, Dosen S, Popovic MB, Popovic D, p. 4 Sensors for assistive system for restoring of the walking. In: Proceedings 52nd ETRAN Conference, Society for Electronics, Telecommunications, Computers, Automatic Control and Nuclear Engineering. 2008.

[28] Duric M. Automatic recognition of gait phases from accelerations of leg segments. In: Neural Network Applications in Electrical Engineering, 2008. NEUREL 2008. 9th Symposium. 2008. doi:10.1109/NEUREL.2008.4685586.

[29] Dejnabadi H, Jolles BM, Aminian K. A new approach to accurate measurement of uni-axial joint angles based on a combination of accelerometers and gyroscopes. IEEE Transactions on Biomedical Engineering 2005;52:1478–84. doi:10.1109/TBME.2005.851475.

[30] Sabatini AM, Martelloni C, Scapellato S, Cavallo F. Assessment of walking features from foot inertial sensing. IEEE Transactions on Biomedical Engineering 2005;52:487–94. doi:10.1109/TBME.2004.840727.

[31] Jasiewicz JM, Allum JH, Middleton JW, Barriskill A, Condie P, Purcell B, et al. Gait event detection using linear accelerometers or angular velocity transducers in able-bodied and spinal-cord injured individuals. Gait & Posture 2006;24(4):502–9. doi:10.1016/j.gaitpost.2005.12.017.

[32] Lau H, Tong K. The reliability of using accelerometers and gyroscopes for gait event identification on persons with dropped foot. Gait & Posture 2008;27:248–57. doi:10.1016/j.gaitpost.2007.03.018.

[33] Liu T, Inoue Y, Shibata K. Development of a wearable sensor system for quantitative gait analysis. Measurement 2009;42:978–88. doi:10.1016/j.measurement.2009.02.002.

[34] Wu G, Ladin Z. The study of kinematic transients in locomotion using the integrated kinematic sensor. IEEE Transactions on Rehabilitation Engineering 1996;4:193–200. doi:10.1109/86.536775.

[35] Veltink PH, Slycke P, Hemssems J, Buschman R, Bultstra G, Hermens H. Three-dimensional inertial sensing of foot movements for automatic tuning of a two-channel implantable drop–foot stimulator. Medical Engineering and Physics 2003;25(1):21–8. doi:10.1016/S1350-4533(02)00041-3.

[36] Williamson R, Andrews BJ. Sensor system for lower limb functional electrical stimulation (FES) control. Medical Engineering and Physics 2000;22:313–25. doi:10.1016/S1350-4533(00)00038-2.

[37] O'Donovan KJ, Kamnik R, O'Keeffe DT, Lyons GM. An inertial and magnetic sensor based technique for joint angle measurement. Journal of Biomechanics 2007;40:2604–11. doi:10.1016/j.jbiomech.2006.12.010.

[38] Lee S-W, Mase K, Kogure K. Detection of spatio-temporal gait parameters by using wearable motion sensors. In: Proceedings of the 2005 IEEE Engineering in Medicine and Biology 27th Annual Conference. 2005. doi:10.1109/IEMBS.2005.1616075.

[39] Dai R, Stein RB, Andrews BJ, James KB, Wiele M. Application of tilt sensors in functional electrical stimulation. IEEE Transactions on Rehabilitation Engineering 1996;4:63–72. doi:10.1109/86.506403.

[40] Weber DJ, Stein B, Chan KM, Loeb G, Richmond F, Rolf R, et al. Bionic walkaide for correcting foot drop. IEEE Transactions on Neural Systems and Rehabilitation Engineering 2005;13:242–6. doi:10.1109/TNSRE.2005.847385.

[41] Stein RB, Chong S, Everaert DG, Rolf R, Thompson AK, Whittaker M, et al. A multicenter trial of a foot drop stimulator controlled by a tilt sensor. Neurorehabilitation and Neural Repair 2006;20:371–9. doi:10.1177/1545968306289292.

[42] Hanlon M, Anderson R. Real-time gait event detection using wearable sensors. Gait & Posture 2006;24:S127–8. doi:10.1016/j.gaitpost.2006.11.088.

[43] Hanlon M, Andersona R. Real-time gait event detection using wearable sensors. Gait & Posture 2009;24:127–8. doi:10.1016/j.gaitpost.2006.11.088.

[44] O'Connor CM, Thorpe SK, O'Malley MJ, Vaughan CL. Automatic detection of gait events using kinematic data. Gait & Posture 2007;25:469–74. doi:10.1016/j.gaitpost.2006.05.016.

[45] Kirkwood CA, Andrews B, Mowforth P. Automatic detection of gait events: a case study using inductive learning techniques. Journal of Biomedical Engineering 1998;11:511–6.

[46] Sweeney PC, Lyons GM, Veltink PH. Finite state control of functional electrical stimulation for the rehabilitation of gait. Medical and Biological Engineering and Computing 2000;38:121–6. doi:10.1007/BF02344765.

[47] Whittle MW. Gait analysis: an introduction. Butterworth-Heinemann; 2006.