A Reliable Gait Phase Detection System

Ion P. I. Pappas, Milos R. Popovic, Thierry Keller, Volker Dietz, and Manfred Morari

***Abstract*—A new highly reliable gait phase detection system, which can be used in gait analysis applications and to control the gait cycle of a neuroprosthesis for walking, is described. The system was designed to detect in real-time the following gait phases: *stance*, *heel-off*, *swing*, and *heel-strike*. The gait phase detection system employed a gyroscope to measure the angular velocity of the foot and three force sensitive resistors to assess the forces exerted by the foot on the shoe sole during walking. A rule-based detection algorithm, which was running on a portable microprocessor board, processed the sensor signals. In the presented experimental study ten able body subjects and six subjects with impaired gait tested the device in both indoor and outdoor environments (0–25 °C). The subjects were asked to walk on flat and irregular surfaces, to step over small obstacles, to walk on inclined surfaces, and to ascend and descend stairs. Despite the significant variation in the individual walking styles the system achieved an overall detection reliability above 99% for both subject groups for the tasks involving walking on flat, irregular, and inclined surfaces. In the case of stair climbing and descending tasks the success rate of the system was above 99% for the able body subjects and above 96% for the subjects with impaired gait. The experiments also showed that the gait phase detection system, unlike other similar devices, was insensitive to perturbations caused by nonwalking activities such as weight shifting between legs during standing, feet sliding, sitting down, and standing up.**

***Index Terms*—Detection, force sensitive resistor, functional electrical stimulation, functional neuromuscular stimulation, gait cycle, gait phases, gyroscope, identification, walking.**

I. Introduction

SINCE the early 1960s a number of systems that applied functional electrical stimulation (FES) were successfully implemented to help individuals with spinal cord injury or brain trauma to restore the walking function [1]–[3], [11], [16]. Liberson *et al.* [1] unilaterally stimulated the ankle dorsiflexion muscles during the swing phase to compensate for the "drop-foot" problem.[1] Later, various FES systems were designed to help subjects with both disabled legs, to walk. These FES systems provided bilateral leg stimulation. Commercial systems such as MikroFES (Jozef Stefan Institute of Science, Slovenia), Odstock (Salisbury District Hospital, U.K.), WalkAide (Neuromotion, Canada), and Parastep (Sigmetics Inc., USA) were developed and successfully used for unilateral or bilateral leg stimulation in subjects with brain trauma and incomplete or complete spinal cord injury.

Thus far several sensor combinations, including manual switches [2], [4], [16], foot switches [2], [4], [16], force sensitive resistors [7], inclinometers [8], goniometers [5], [7], gyroscopes [12], accelerometers [9], [10], electromyography (EMG) sensors [11], and implanted recording nerve-cuff electrodes [13], [14], were proposed to control the timing of the stimulation sequences generated by FES systems for walking. These sensors were used to distinguish between stance and swing phase during walking or to identify multiple gait cycle phases (events). In what follows, we will briefly review these sensors and their detection performance. The study carried out by Ott *et al.* in [4] and our study in [16] showed that a foot switch, due to its poor detection reliability, is not the appropriate solution for triggering stimulation sequences of a FES system for walking. In particular, the foot switch cannot differentiate foot loading and unloading sequences generated during walking from those that are caused by weight shifting between the legs during standing. In a study by Ng and Chizeck [5], measurements from goniometers at the hip, knee, and ankle joints were used in combination with a fuzzy model classification method to detect five different gait phases. However, the method suffered from frequent detection errors, as shown in the article. Kostov *et al.* in [7] used a machine learning technique to replicate the time-instances when a subject manually triggered the onset of the stimulation sequences while walking with a FES system for "drop-foot" compensation. Inputs to the detection algorithm were measurements from force sensitive resistors placed on the shoe insole and measurements from goniometers that recorded hip adduction/abduction, hip flexion/extension, knee flexion/extension, ankle flexion/extension, and ankle inversion/eversion. This system did not provide any information about the actual gait phase of the subject during walking. In a study by Dai *et al.* [8] different tilt sensors and inclinometers were used to detect the step intention (the moment when the lower leg tilts backward at the end of the stance phase) in order to trigger a FES system for "drop-foot" compensation. Experiments carried out at our laboratory indicate that this approach is not sufficiently reliable. We have discovered that during nonwalking tasks, such as sitting down and standing up, a subject with a "drop-foot" problem often had the shank moving in the range typical for walking, and a tilt sensor or an inclinometer identified this motion as a step intention. Kirkwood *et al.* in their case study presented in [6] combined measurements from goniometers and instrumented shoe insoles to derive four distinct gait phases using inductive learning

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[1]"Drop-foot" is the term commonly used to describe the inability of the subject to contract the ankle dorsiflexor muscles and to lift the foot off the ground, which is essential during the swing phase.

techniques. The experiments performed in their laboratory showed that the detection accuracy of the proposed system was between 70% and 97%. That is insufficient for practical applications. In [12], Tong and Grant showed that with the use of two gyroscopes, one placed on the thigh and the other one on the shank, one can measure the knee angle during walking. In [9], Willemsen *et al.* suggested the use of accelerometers, placed below the knee, to distinguish stance and swing phases during the gait cycles. This system like the one proposed by Dai *et al.* in [8] provided only information about the transition between the stance and swing phases while other gait phases were not treated. In [10], Williamson and Andrews used three accelerometers mounted on the shank of the leg and a machine learning algorithm to detect the transitions between five gait phases during walking. The authors reported that except for occasional signal chattering and small time delays (average time delay was in the range 23–40 ms) the system had 100% detection reliability. This system still remains to be tested in the outdoor environment and during walking on irregular surfaces, slopes, and stairs. Graupe *et al.* in [11] used EMG measurements of the pectoralis muscles' activity to trigger stimulation sequences of a neuroprosthesis that facilitated bipedal walking. Although very effective, the proposed system did not provide information about gait phases. Another interesting method for detecting transitions between the stance and swing phases is based on measurements of afferent nerve activity with implanted nerve-cuff electrodes. Strage and Hoffer [13] and Upshaw and Sinkjaer [14] implanted nerve-cuff electrodes in cats and humans, respectively, and reported that they could detect the transition between the stance and lift-off gait phases with reliability close to 99%. Systems measuring afferent nerve activity are invasive, sensitive to electromagnetic interference, and so far provide information about only one gait event.

Despite the numerous sophisticated techniques for gait phase identification, in practice the most widely used method to trigger the electrical stimulation sequences is still a manual switch (push button), because up to now it was the most reliable method. The main disadvantages of the manual push-button control technique are that it requires the subject's uninterrupted and continuous attention and that with the push-button the subject can only indicate a limited number of gait events. To improve the quality of the leg motion in future FES systems for walking it is crucial to precisely control the timing of the applied stimulation sequences. To achieve this, one possible approach is to develop an automatic, accurate, and reliable gait phase sensor that can provide information about multiple gait phases during walking and to use this information in a closed-loop scheme with the FES system. In this paper a new portable gait phase detection system (GPDS) for closed-loop FES walking applications is presented.

The GPDS uses a different (new) sensor combination consisting of three force sensitive resistors that measure the force load on a shoe insole and a miniature gyroscope-chip that measures the rotational velocity of the foot. The system detects accurately and reliably in real time the four gait phases: *stance*, *heel-off*, *swing*, and *heel-strike* (definitions are provided in Appendix A). Experiments that involved ten healthy subjects and six subjects with impaired gait showed that the GPDS reliably identified the above gait phases both in indoor and outdoor walking tasks, despite the large variation in walking styles and walking conditions. The experimental study included walking tasks such as walking on flat and irregular ground, walking on slopes, stepping over small obstacles, and climbing and descending stairs. Furthermore, the GPDS has the ability to distinguish "true" walking from feet sliding or shifting of the weight from one leg to the other while standing. Finally, the GPDS has the potential to be miniaturized to fit together with the microcontroller unit inside a 1-cm-thick shoe insole. Although the GPDS was developed for close-loop FES walking applications, it could also be used in gait analysis applications, as a diagnostic and screening tool for assessing gait pathologies, as a tool to provide biofeedback in rehabilitation applications, and for various virtual reality or computer games' applications.

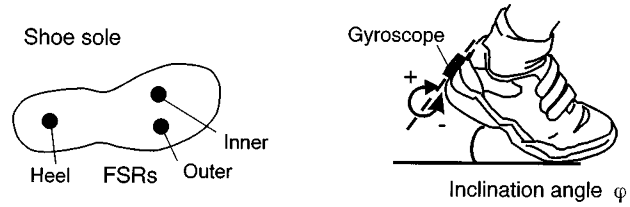


Fig. 1. Placement of the sensors used by the GPDS.

II. Methods and Materials

A. Hardware Description

The GPDS relies for the detection of the gait phases on two types of "off-the-shelf" sensors: 1) three force sensitive resistors (FSRs) that are used to measure forces exerted by the foot on the shoe insole during walking and standing, and 2) a gyroscope that measures the rotational velocity of the foot (see Fig. 1). The sensor signals were sampled at a frequency of 100 Hz with a resolution of 10 bits and processed on a 20-MHz microcontroller board (Hitachi SH7032). No calibration of the sensors was needed prior to the experiments. The FSRs (Interlink El. Inc. FSR 152 NS) were small (∅ 1 cm) flat resistors whose resistance changes nonlinearly with applied force. One of the FSRs was placed underneath the heel and two underneath the first and fourth heads of the metatarsal bones. Two (instead of one) FSRs were used underneath the metatarsal heads since the foot is not always loaded symmetrically (irregular ground or asymmetric gait style). For the experiments the FSRs were taped with a masking tape on a 3-mm-thick shoe insole. The same insole and FSR sensors were used for each subject, but the insole size and positions of the FSRs were adjusted according to the individual's foot size and form. The FSRs are not precision sensors (specified $\pm 25\%$ part-to-part repeatability), therefore they were only used as two-state switches to indicate when weight was applied to them and when not, which was achieved by measuring the voltage drop across each FSR connected in a voltage divider circuit. Their specified switching time delay was 1 ms. The FSRs alone neither can distinguish between true walking and weight shifting from one leg to the other, nor can they provide any information about the foot condition during the swing phase.

The miniature gyroscope (Murata ENC-03J, size $15.5 \times 8.0 \times 4.3$ mm, weight 10 g) was attached to the posterior aspect (heel) of the shoe with its sensing axis oriented perpendicular to the sagittal plane to measure rotations of the foot in that plane (see Fig. 1). The Murata ENC-03J gyroscope measured the rotational velocity by sensing the mechanical deformation caused by the Coriolis force on an internal vibrating prism. The gyroscope signal was filtered by a third-order bandpass filter (0.25–25 Hz) with a 20-dB gain in the pass band. The frequencies outside the passband were filtered out because they were not related to the walking kinetics. The filtered gyroscope signal was used to directly estimate the angular velocity of the foot and it was integrated to estimate the inclination of the foot relative to the ground. A resetting mechanism was built in the algorithm to avoid accumulation of drift errors in the integrated signal. The foot inclination (integrated gyro signal) was reset to zero during the stance-phase when all three FSRs were loaded. A detailed discussion about the gyroscope signal processing and the effect of varying ambient temperature on the gyroscope performance was presented in [15].

B. The Gait Phase Detection Algorithm

The GPDS divided the gait cycle in four different gait phases: *stance, heel-off, swing*, and *heel-strike*. These gait phases were represented in the form of a state machine with four distinct states. The loop frequency of the state machine was 100 Hz, i.e., equal to the sensor sampling frequency. The transitions between the states were governed by a knowledge-based algorithm, which was derived after off-line processing and testing of numerous experimental data sets. The algorithm allowed a total of seven different transitions (*T1–T7*) between the four gait phases, as illustrated in Fig. 2. A summary of the rules governing these transitions is given below.

Note: In the following discussion we assume that the subject is viewed from the right lateral side and clockwise rotations are considered positive. The symbol φ represents the inclination angle of the heel with respect to the ground and $\varphi_{\rm th}$ is the threshold for the detection of the heel-off phase. The variables *heel, metat1*, and *metat4* represent the status of the FSRs underneath the heel, and the first and fourth metatarsal heads, respectively. Finally, ε_ω and ε_α represent small threshold values for the detection of close to zero angular velocities and accelerations, respectively.

Transitions Between States:

T1: stance $\rightarrow$ *heel-off)* In the stance phase, the algorithm awaits the beginning of the heel-off phase. The heel-off phase is detected when the heel FSR is not pressed and the inclination of the heel with respect to the ground φ exceeds a given threshold angle $\varphi_{\rm th}$ (3°). The inclination of the heel is obtained by integrating the gyroscope signal.
Transition condition: $\varphi \geq \varphi_{\rm th}$ AND *(heel = not pressed)*.

T2: heel-off $\rightarrow$ *swing)* In the heel-off phase the algorithm anticipates the transition to the swing phase. The condition for the transition to the swing phase is that none of the FSRs is pressed and that the rotation of the heel changes from positive (in the heel-off phase) to negative direction.
Transition condition: $(\varphi' < 0)$ AND *(heel = metat4 = not pressed)*.

T3: swing $\rightarrow$ *heel-strike)* In the swing phase the algorithm awaits the transition to the heel-strike phase, which begins with the first contact of the foot with the ground. Thus, the heel-strike phase is detected as soon as any of the FSRs are pressed. In normal gait, the heel touches the ground first. However, when climbing stairs or in pathological gait styles, the first contact can be established with the front part of the foot.
Transition condition: *(heel = pressed)* OR *(metat1 = pressed)* OR *(metat4 = pressed)*.

T4: heel-strike $\rightarrow$ *stance)* After the heel-strike the next phase is stance, which begins when both the front and rear part of the foot touch the ground. This event is detected when the heel FSR and at least one of the front FSRs are pressed. It is not required that both front FSRs are pressed, because on irregular ground (or if the subject has a pathologic walking style) only one side of the foot may be loaded. A special case is when a subject climbs stairs and places only the front part of the foot on the step. According to the above rule the GPDS could not detect the stance phase. However, we know that during the stance phase the rotational velocity of the foot is close to zero. Therefore, the transition from the heel-strike phase to the stance phase is also detected if the rotational velocity and its derivative are close to zero.
Transition condition: [*(heel = pressed)* AND (*(metat1 = pressed)* OR *(metat4 = pressed)*)] OR [$(|\varphi'| < \varepsilon_\omega)$ AND $(|\varphi''| < \varepsilon_\alpha)$].

The following state transitions are allowed as well:

T5: heel-off $\rightarrow$ *stance)* If the subject lifts the heel and then places it back onto the ground (without going into a swing phase, as for normal walking) this event is detected in the gait phase detection algorithm by a transition from heel-off to stance (T5). If during the heel-off phase the status of the heel FSR is pressed, the algorithm transits to stance phase.
Transition condition: *heel = pressed*.

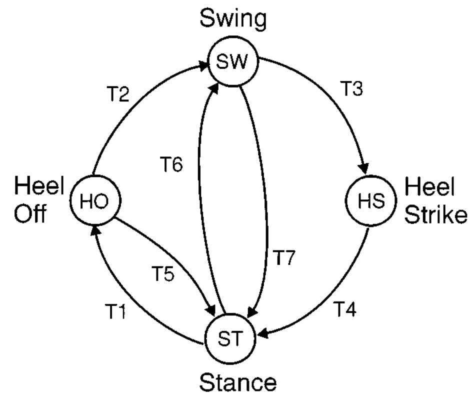


Fig. 2. The GPDS divided the walking cycle into four gait phases: stance, heel-off, heel-strike, and swing. The arrows *T1–T7* illustrate the possible transitions between the gait phases.

TABLE I
SUBJECTS' DATA FOR BOTH GROUPS A AND B

	male	female	av. age [range]	av. Weight [range]	av. height [range]
group A	7	3	31.5 [25-35]	68 [50-86] kg	1.74 [1.62-1.83]
group B	5	1	42 [19-60]	64 [53-70] kg	1.69 [1.64-179]

T6: stance → swing) In certain pathological gait-styles and occasionally in the first of a sequence of steps the foot does not go through a true heel-off phase (no detectable heel rotation) but moves directly from stance to swing phase. Such a transition (T6) takes place under the condition that none of the three FSRs is pressed and that the gyroscope signal indicates a negative rotational velocity of the foot.
Transition condition: *same as for T2.*

T7: swing → stance) In certain pathological gait styles the heel-strike phase is too short and it can not be detected as a distinct phase. In this case the algorithm transits from the swing phase directly to the stance phase (T7). The condition for this transition is that the heel FSR and at least one of the front FSRs is pressed or that the rotational velocity and its derivative are close to zero.
Transition condition: *same as for T4.*

C. The Optical Motion Analysis System

To validate the GPDS' performance we have compared the GPDS output with the measurements obtained with the commercial optical motion analysis system Vicon 370 (Oxford Metrics Ltd., U.K.). Vicon tracked and measured the three-dimensional (3-D) trajectories of retro-reflective markers placed on the subject's body, as shown in Figs. 3 and 4(b). Three Vicon cameras with sampling frequency 50 Hz were used to track the marker positions with accuracy of ± 1 mm. The markers' trajectories were used to extract a "reference" gait phase signal that was later used to validate the accuracy of the GPDS output signal (how the reference signal was generated out of raw Vicon measurements is described in Appendix B).

TABLE II
ADDITIONAL INFORMATION ABOUT SUBJECTS IN GROUP B

Subject	Lesion/ Disability	Walking-Aid
1.	incompl. C4	Crutches
2.	incompl. C5/C6	None
3.	Cauda equina lesion L1	Crutches
4.	Cauda equina lesion L5	Crutches
5.	Brain-Stroke	Ankle foot orthosis (right).
6.	Cauda equina lesion L1	Crutches

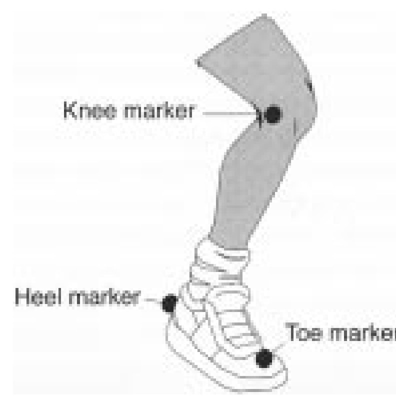


Fig. 3. Positions of the retroreflective markers that were placed on the knee, heel, and toe. The Vicon measurement system tracked and measured the 3-D trajectories of these markers during the experiments.

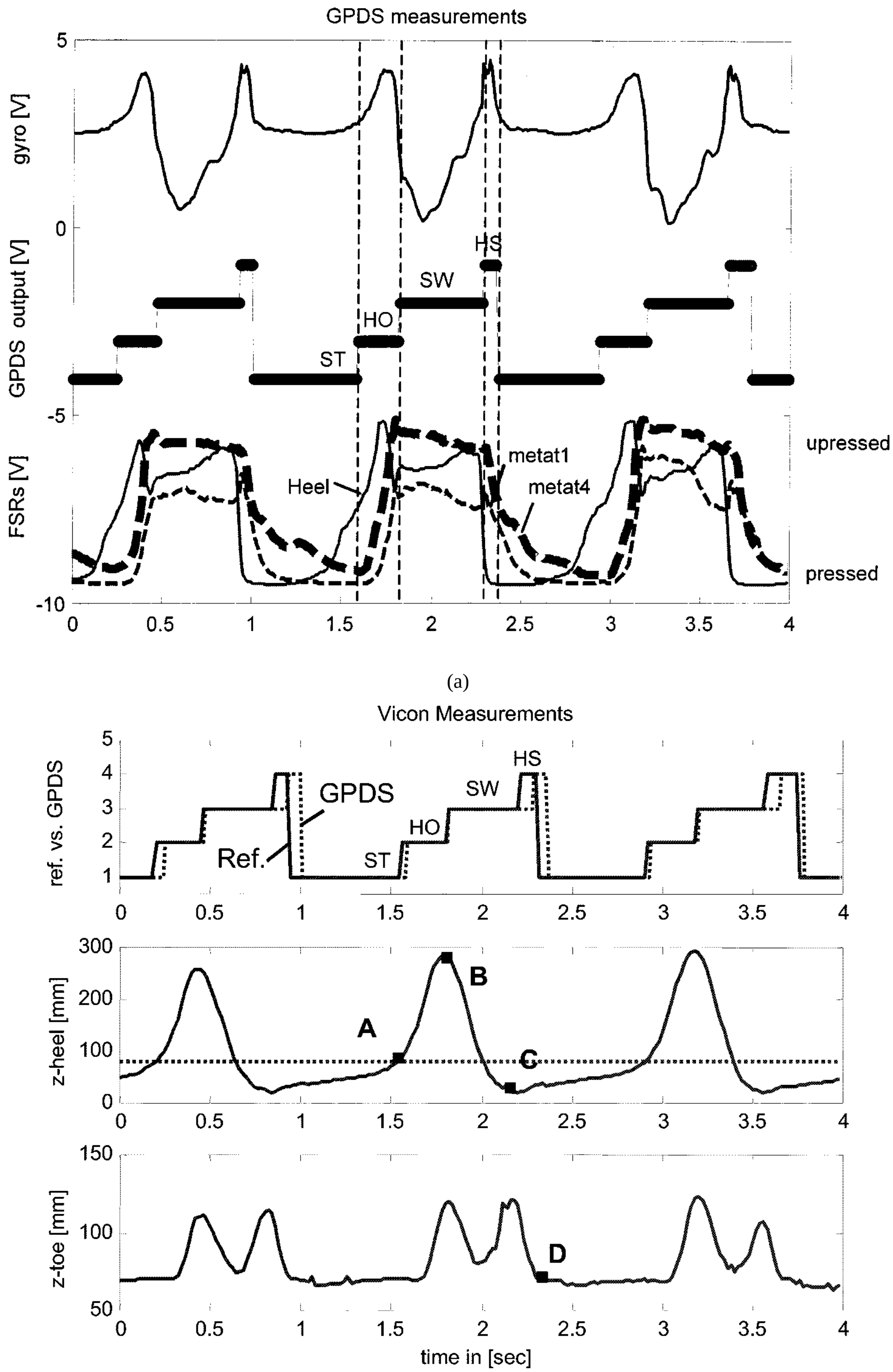
III. EXPERIMENTAL STUDY

An experimental study was carried out in order to quantify the detection success ratio of the GPDS on a wide variety of gait styles. The study involved a group of ten healthy adults (group A) and a group of six adults with various gait pathologies (group B). The subjects from the group A had no known orthopedic, metabolic, or neurological impairments or pain that could modify or influence their natural walking patterns (see Table I). Despite their gait pathologies, the subjects from the group B all were able to walk short distances with or without crutches (see Tables I and II). All subjects were informed about the purpose of the experiment and written consent was obtained from each subject prior to the study.

The experimental study consisted of four parts. In Part I, the delay and accuracy of the GPDS output were evaluated using the reference gait phase signal obtained from the Vicon 370 system measurements. In Part II, performance of the GPDS was tested on a variety of walking tasks such as walking on level ground, walking on slopes, walking on irregular surfaces, and climbing stairs. Part III consisted of a number of nonwalking tasks to verify that the GPDS does not produce false gait detection during these tasks. Finally, the purpose of Part IV was to determine the range of walking speeds for which the GPDS yields reliable results.

A. Experiments Part I—Comparison of the GPDS with a Motion Analysis System

The goal of these experiments was to validate the GPDS output with an external reference measurement system and to quantify the delay time of the GPDS. For this purpose the 3-D optical motion analysis system Vicon 370 was used. Three randomly selected able-bodied subjects were asked to walk at least 20 steps at walking speeds of 3 and 5 km/h on a treadmill



(b)

Fig. 4. (a) The gyroscope signal (top), the FSR signals (bottom), and the GPDS output signal (middle) for a sequence of three gait cycles of a subject walking on the treadmill. (b) Synchronized Vicon measurements of the heel marker (middle) and toe marker (bottom) trajectories in the vertical direction. From the marker measurements (reference points A, B, C, and D) we extracted a *reference* gait phase signal (top, solid line), which was used to evaluate the delay time of the GPDS output signal (top, broken line). (Note: ST = stance, HO = heel-off, SW = swing, HS = heel-strike, heel-FSR = solid line, front-FSRs = broken lines).

with retro-reflective markers placed on their heel, toe, and knee as shown in Fig. 3. Three Vicon cameras were used to track the trajectories of the body markers while the GPDS output signal was recorded. Based on the marker trajectories the reference gait phase signal was generated according to a set of rules given in Appendix B. Then the reference signal was compared to the measured GPDS output.

B. Experiments Part II—Four Walking Tasks

The purpose of the second set of experiments was to test the performance of the GPDS on a large number of subjects during different walking tasks and under real environment conditions. The experiments were carried out for the largest part outdoors during winter time (temperatures: 0 °C–10 °C) and consisted of the following tasks:

TABLE III
DETECTION RESULTS OF THE EXPERIMENTS PART II

Exp II	walking tasks:	level ground	Slope	irregular surface	stairs upwards	stairs downward
group A	# of recorded steps	608	1303	946	517	424
	# of correctly detected steps	608	1303	946	517	422
group B	# of recorded steps	306	298	314	64	58
	# of correcty detected steps	306	297	306	62	55

1) walk on level ground (length $= 100$ m);
2) walk up and down a steep cobblestone road (length $= 2 \times 50$ m, 15% inclination);
3) walk on grass, snow, earth, step over small obstacles, and step on and off the pavement (length $= 100$ m, maximum obstacle height $= 10$ cm);
4) ascend and descend the stairs (2×50 steps).

All subjects of groups A and B participated in the experiments. Some subjects of the group B were not able to complete all of the tasks due to their disability. The subjects were asked to walk at their preferred speed carrying on their back a portable computer (3 kg), which recorded the GPDS output signal and the FSRs and gyroscope signals. To test if the system is robust to changes in ambient temperature the first half of task 1) was completed indoors and the second half outdoors. In these experiments the success rate of the GPDS was determined by comparing the GPDS output signal and raw sensor signals. The recorded sensor signals were plotted as shown in the example in Fig. 5(a) and the consistency between the GPDS output signal and the sensor signals was verified visually. Video recordings were employed in some cases to validate the described evaluation procedure.

C. Experiments Part III—Nonwalking Tasks

The purpose of the third set of experiments was to verify that the GPDS does not falsely identify any gait phases during nonwalking activities. Nonwalking activities such as standing up, sitting down, standing still, shifting the weight from one leg to the other during standing, and sliding of the feet, belong to our daily locomotion activity. Therefore, the GPDS must be robust against perturbations caused by these and similar nonwalking activities. All subjects of groups A and B participated in these experiments and were asked to perform the following tasks:

1) stand up from a chair and sit down (five repetitions);
2) stand, then bend the knees and touch the floor with the fingers (five repetitions);
3) stand upright and rotate in clockwise direction for 360° and in counter-clockwise direction for 360° around the subject's own vertical axis (sliding of the feet was allowed but lifting of the feet was forbidden).

D. Experiments Part IV—Speed Range

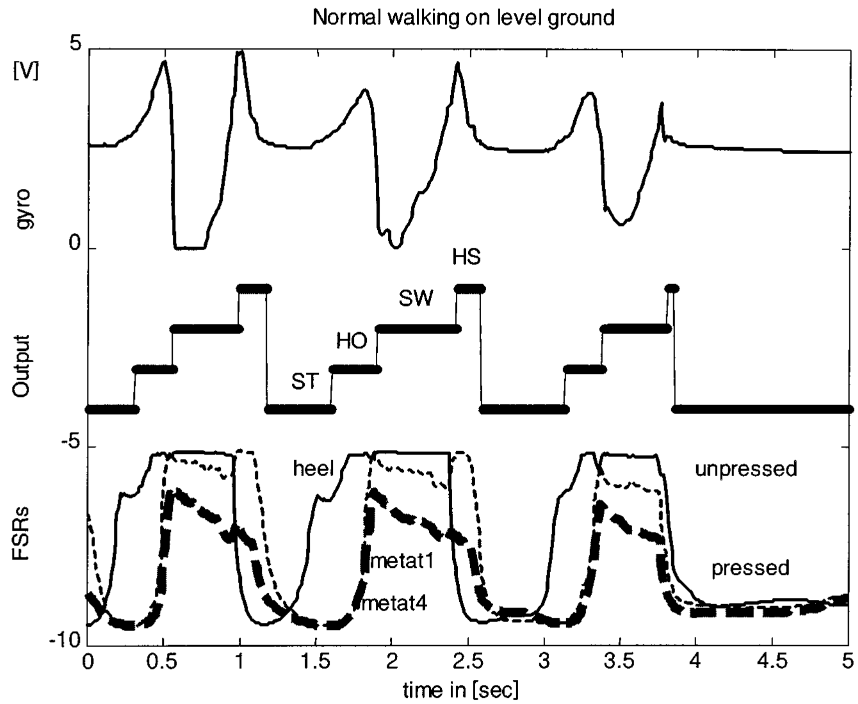
The purpose of the last set of experiments was to determine the range of walking and running speeds for which the GPDS yields reliable results. Three randomly selected able body subjects were asked to walk/run on the treadmill. The treadmill speed was gradually increased from a slow speed of 0.5 km/h to a maximum speed of 13 km/h in steps of 0.5 km/h for the first step and 1 km/h for all consecutive steps ($0.5, 1, 2, 3, \ldots, 13$ km/h). Speed of 13 km/h corresponds to a fast jogging speed. For each subject, we recorded the sensor signals and the GPDS output signal for ten steps at each of the above speeds.

IV. EXPERIMENTAL RESULTS

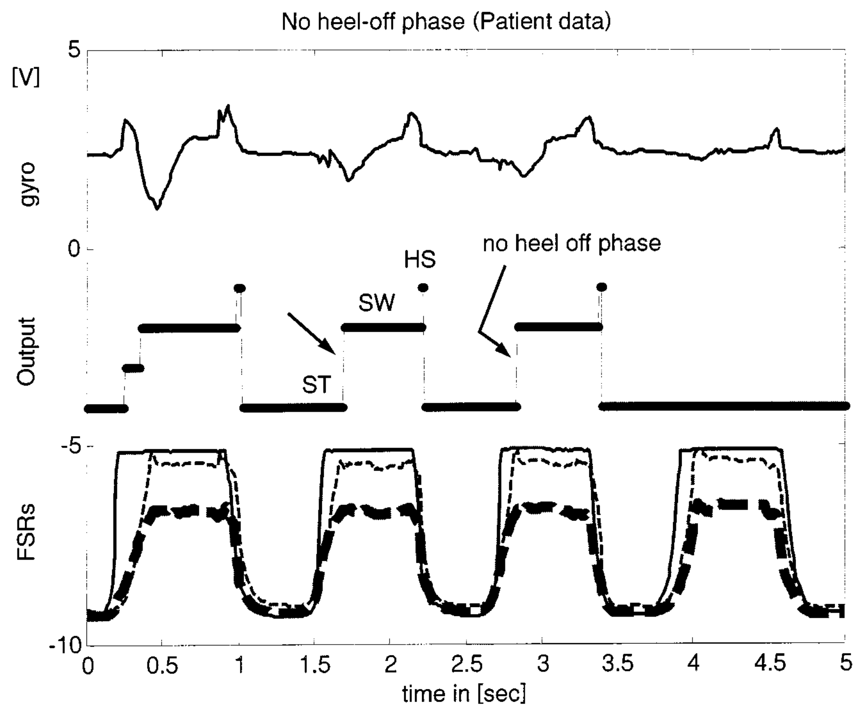
A. Results Part I—Comparison of the GPDS with a Motion Analysis System

In these experiments we validated the GPDS performance with Vicon measurements. A typical example of the gyroscope signal (top), FSRs signals (bottom), and GPDS output signal (middle) recorded during the walking of an able body subject is shown in Fig. 4(a). Synchronized Vicon measurements of the (heel and toe) marker positions in the vertical direction are shown in Fig. 4(b). A comparison between the reference gait signal (solid line), which was generated based on the heel and toe markers' trajectories, and the GPDS output signal (broken line) is shown in Fig. 4(b)-top. The GPDS output correlated well with the reference gait phase signal for all trials. However, a time delay of the GPDS signal relative to the reference signal was observed, in particular in the detection of the heel-strike and stance phases. Averaged measurements of 60 gait cycles (three subjects and 20 steps per subject) at the walking speed 3 km/h indicated that the GPDS delay for the detection of the four gait phases relative to the reference signal was approximately 40 ms for heel-off, 35 ms for swing, 70 ms for heel-strike, and 70 ms for stance. Given that the Vicon sampling frequency was 50 Hz, the reference signal itself may lag additional 20 ms from the actual gait event. Therefore, in the worst case the gait phase detection delay was less than or equal to 90 ms.

The time lag of the GPDS can be explained by the fact that the reference signal was based on a video signal, while the GPDS was partly based on force measurements (FSRs). For example, at the heel-strike phase, first the heel contacted the ground and then it was loaded with weight. Some authors differentiate during the heel-strike phase between "initial contact" and "weight acceptance" in order to emphasize the time-difference of the two events. In our case, the reference signal switched from swing to heel-strike at the "initial contact," whereas the GPDS did it at the "weight acceptance" event. The other observed time delays can be all explained in the similar fashion.



(a)

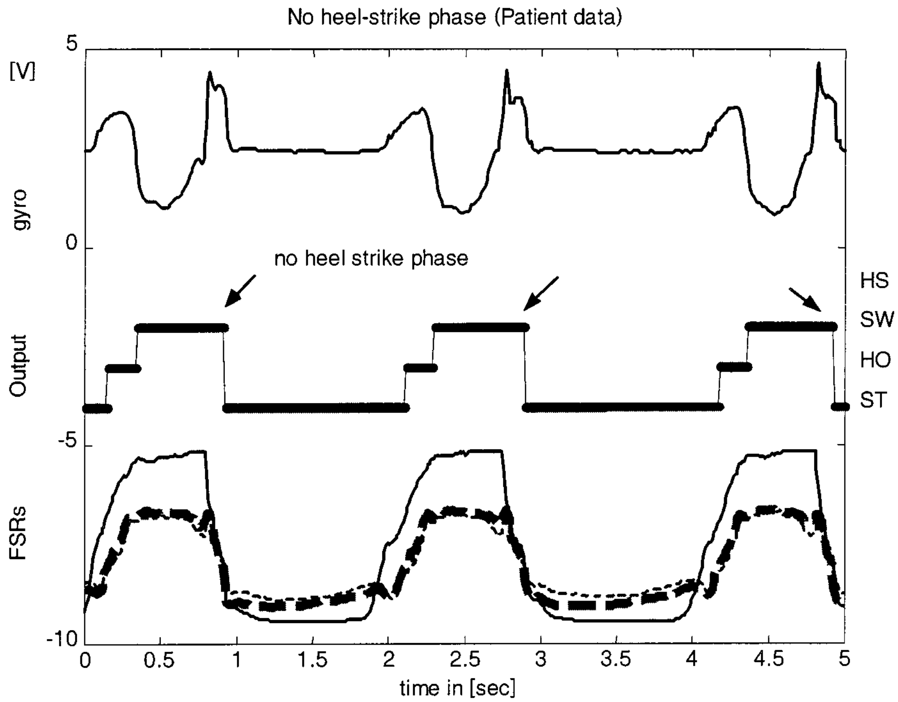


(b)

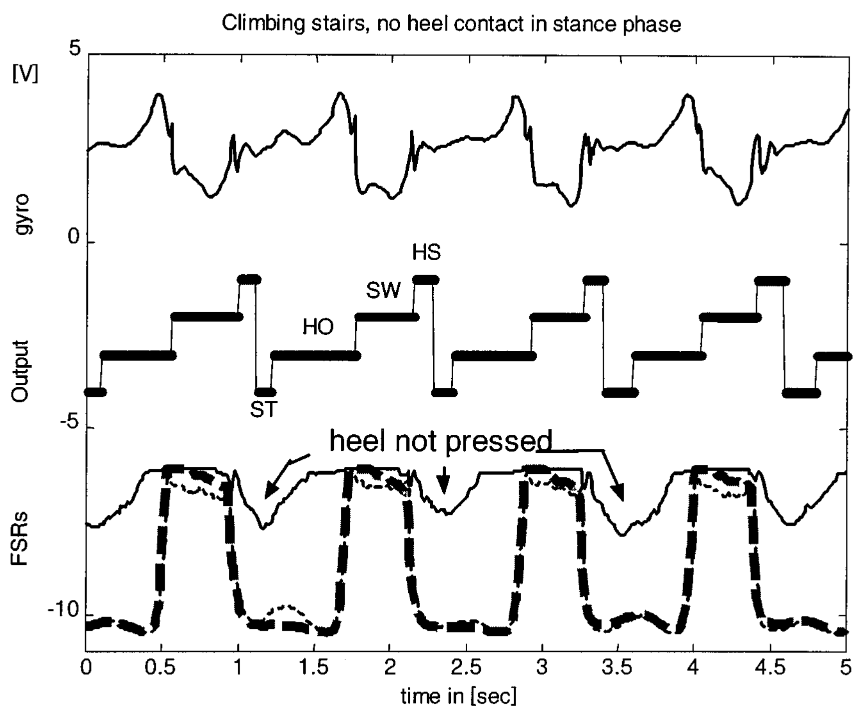
Fig. 5. Examples of different walking conditions. (a) *Normal walking on level ground*. (b) *A subject with a weak gastrocnemius muscle.* In the second and third step the excursion of the gyroscope signal in the heel-off phase was too low to be detected by the GPDS, which switched from the stance phase into the swing phase, skipping the heel-off phase. (Note: ST = stance, HO = heel-off, SW = swing, HS = heel-strike, heel-FSR = solid line, front-FSRs = broken lines.)

B. Results Part II—Four Walking Tasks

The results of this set of experiments showed that the GPDS detected the gait phases with excellent reliability for many different walking tasks such as walking on level ground, on slopes, on irregular terrain, and on stairs (see Table III). During the first three tasks (walking on level ground, on slopes, and irregular terrain), we recorded a total of 2857 $(=608 + 1303 + 946)$ steps for group A and 918 $(=306 + 298 + 314)$ steps for group B. The GPDS detected correctly all four phases in all recorded steps of the group A and failed in 9 steps of the group B, which yields a success rate of 100% and 99%, respectively. A typical example of the GPDS sensor signals and output signals is shown in Fig. 5(a). The walking style of the subjects of group B was much more irregular and unpredictable compared to the able body subjects' walking. For example, the subjects in group B



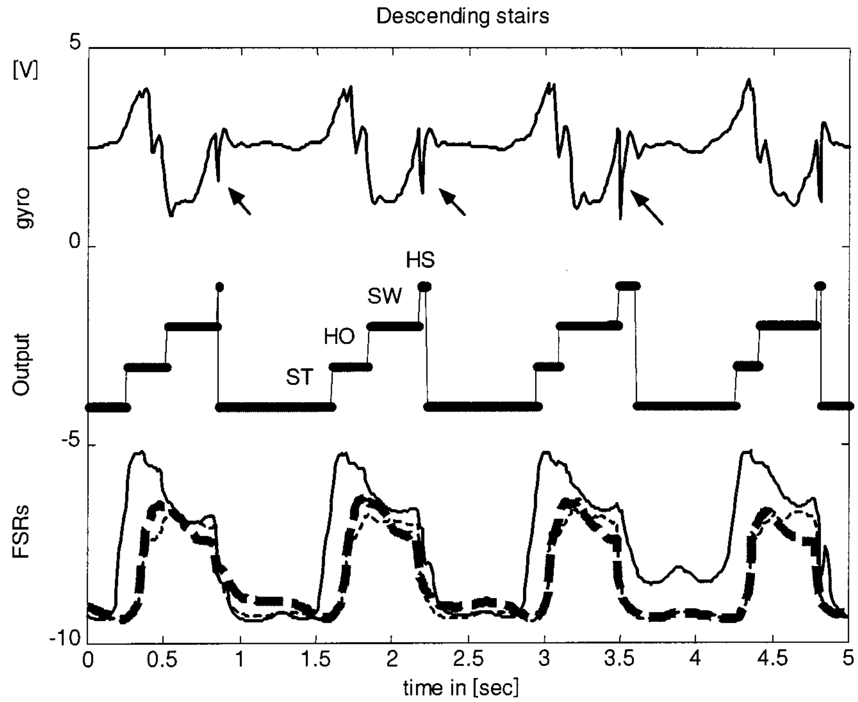
(c)



(d)

Fig. 5. (Continued.) (c) *A subject with a weak tibialis anterior muscle.* As shown by the FSR signals, the front and rear parts of the foot were pressed simultaneously causing transition from swing into stance phase without going through heel strike phase. (d) *Walking up stairs.* Many subjects climbed the stairs "on their toes" without placing the heel on the ground. In this case to detect the stance phase the algorithm looked for angular velocity that was equal to 0. (Note: ST = stance, HO = heel-off, SW = swing, HS = heel-strike, heel-FSR = solid line, front-FSRs = broken lines.)

often hesitated which leg to move forward when stepping over an obstacle. They sometimes paused, initiated a step, and terminated it abruptly. In some cases, usually at the first or last step of a sequence of steps, the subject performed steps with very poorly pronounced heel-off phases (i.e., the heel was not lifted at all, or was lifted below the detection threshold). In such a case, shown in Fig. 5(b), the GPDS skipped the heel-off phase switching directly from the stance phase to the swing phase. Similarly, as shown in Fig. 5(c), some group B subjects occasionally did not perform a well-pronounced heel-strike phase, but placed the foot flat on the ground without first performing the heel contact. In such cases the GPDS skipped the heel-strike phase switching directly from the swing phase to the stance phase.



(e)

Fig. 5. (Continued.) (e) *Descending stairs.* Compared to level ground walking during the heel-strike phase the gyroscope signal had a downward peak instead of an upward peak. The reason was that the foot touched the ground first with the front part of the foot and then with the heel. (Note: ST = stance, HO = heel-off, SW = swing, HS = heel-strike, heel-FSR = solid line, front-FSRs = broken lines.)

During the stair climbing task, we recorded from group A a total of 517 ascending and 424 descending steps. All gait phases, except for two (both during descending stairs), were correctly identified yielding overall success rate of 99.78%. From group B only three subjects were able to climb stairs and their steps during this exercise were very irregular. In total 64 ascending and 58 descending steps were recorded out of which 96% were correctly identified. One significant difference between the stair climbing task and the normal walking is that at the end of the swing phase the first contact with the ground (heel-strike) is in fact established with the front part of the foot, and not with the heel. Furthermore, some subjects climbed or descended the stairs using only the front part of their foot, i.e., only the front part of the foot was placed on the step while the heel remained "in the air." As a result, the heel FSR was often not pressed at all during the entire gait cycle as shown in Fig. 5(d). To overcome this problem the algorithm also looked at the gyroscope signal in addition to the FES signals, since during stance the rotational velocity of the foot is always zero. Thus the algorithm detected the onset of the stance phase as soon as the gyroscope became still, and its rotational velocity and acceleration were close to zero.

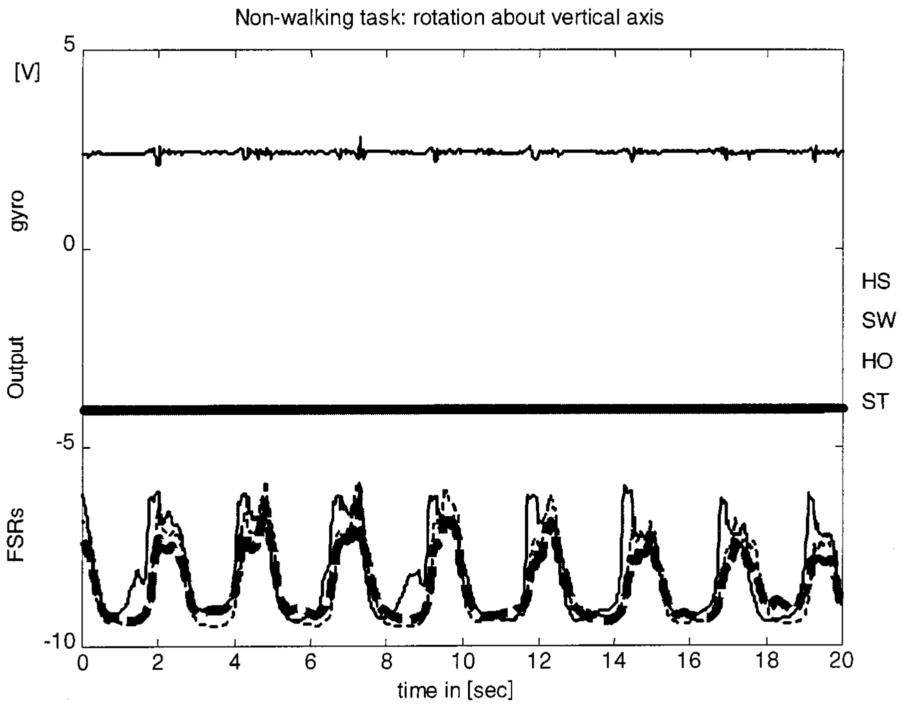
We have also observed a fundamental difference in the gyroscope signals of normal walking and the stair descending task. When walking on level ground (seen from the right lateral side) the foot rotates clockwise in the heel-off phase, then anti-clockwise in the swing phase, and again clockwise in the heel strike phase (+-+). During stair descending the situation is different. First the foot rotates clockwise, then anti-clockwise, and then again anti-clockwise (+--), because the toes contact the ground first and then the heel. In spite of this difference the GPDS identified the gait phases correctly as shown in Fig. 5(e).

In this set of experiments no specific information was obtained about the timing of the GPDS output relative to an external measurement system. The timing of the gait phases was explicitly examined in the previous set of experiments (Part I) and, since the hardware was not modified, the timing properties should remain the same.

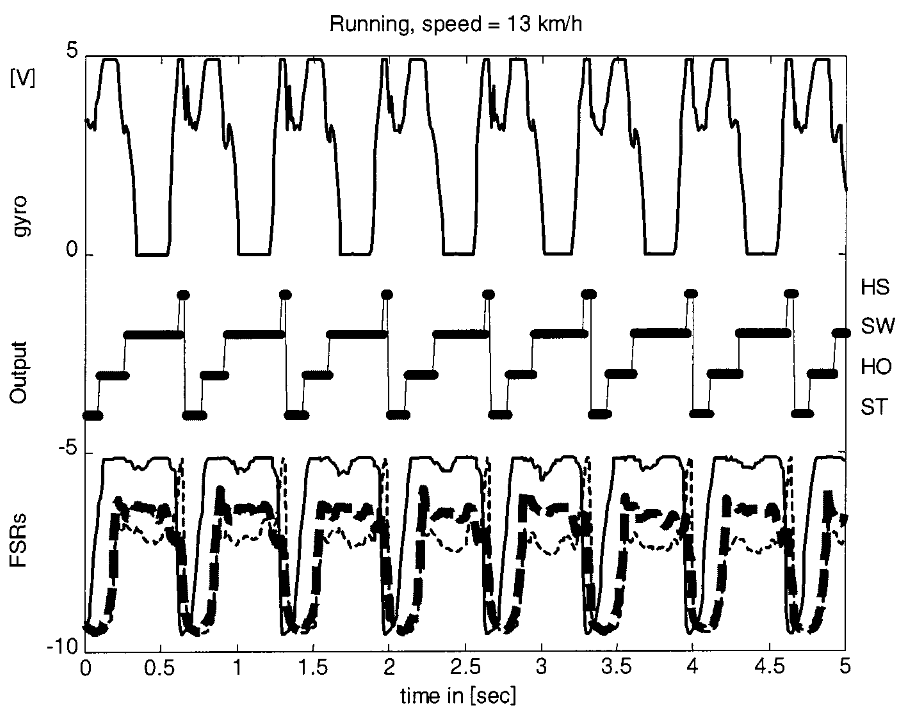
C. *Results Part III—Nonwalking Tasks*

The results of these experiments showed that the GPDS was not only reliable during walking but was also robust (did not wrongly detect gait phases) during nonwalking tasks such as sitting down, standing up, bending the knees, and sliding the feet. During the nonwalking tasks, even though the FSR signals appeared identical to those of walking tasks (due to loading and unloading of the foot) the gyroscope signal amplitude was close to zero, indicating that the foot was not rotating, i.e., remained on the ground [see Fig. 6(a)]. Thus an advantage of the GPDS is that it does not need to be turned on and off every time the subject starts or stops walking but can remain in operation continuously. As shown by the summarized experimental results in Table IV the GPDS did not detect any gait phases during nonwalking tasks for both subject groups.

One should note that there is a tradeoff between the detection robustness and the detection delay in the case of the GPDS. For example, in the stance phase inclinations of the heel larger then the selected threshold of 3° triggered the transition of the GPDS from stance into heel-off phase. To make the system more robust to perturbations during stance a larger threshold can be applied which would in turn cause greater delay in the detection of the heel-off phase during walking.



(a)



(b)

Fig. 6. (a) Illustrates the sensor signals and the GPDS output during a nonwalking task where the subject rotated around his/hers own vertical axis. The FSR signals were similar to walking FRS signals because the subject shifted his/hers weight from one leg to the other, but the gyroscope signal amplitude remained minimal, since there was no foot rotation in the sagittal plane. The GPDS correctly identified that there was no gait-phase change. (b) Running at 13 km/h. Due to the high angular velocity of the foot the gyroscope signal saturated in the heel-off, swing, and heel-strike phases. However, all four gait phases were correctly identified.

D. Results Part IV—Speed Range

The aim of these experiments was to determine the range of walking/running speeds for which the GPDS yields reliable detection results. The experimental results showed that the GPDS detected the gait phases correctly (100% reliability) for all three test-subjects in the entire length of the experiment (ten steps at each of the speeds from 0.5 to 13 km/h). As shown in Fig. 6(b) for the running speed of 13 km/h the gait cycle and in particular the stance phase are quite short 0.6 and 0.1 s, respectively. Despite larger rotational velocities of the foot that caused the gyroscope signal to saturate during the heel-off, swing, and

TABLE IV
DETECTION RESULTS OF THE EXPERIMENTS PART III

Exp III	Tasks	stand up and sit down	bend knees	rotation ± 360°
group A	# of recorded repetitions	50	50	20
	# of correctly detected steps	50	50	20
group B	# of recorded repetitions	32	30	20
	# of correctly detected steps	32	30	20

heel-strike gait phases, the GPDS maintained its detection reliability at 100%.

V. Conclusions and Discussion

A new gait phase detection sensor that reliably identified the transitions between *stance, heel-off, swing,* and *heel-strike* gait phases, was presented. This real-time system was based on a simple set of off-the-shelf sensors including three force sensitive resistors placed on a shoe insole and a miniature gyroscope placed at the posterior aspect (heel) of the shoe. The gyroscope's output signal was used to estimate the rotational velocity of the foot in the sagittal plane as well as the foot's inclination relative to the ground (integration of the gyroscope's signal). A resetting mechanism of the integrated signal was applied during the stance phase to avoid unwanted drifts of the inclination measurements. The sampling frequency of the sensor signals and the loop frequency of the gait phase detection algorithm were 100 Hz and were carried out on a portable 20-MHz microcontroller board. The performance of the GPDS was verified using a standard Vicon 370 optical motion analysis system and with respect to raw FSRs and gyroscope measurements. Both able body subjects and subjects with walking impairments tested the GPDS by performing diverse walking and nonwalking tasks. The detection success rate for both groups of subjects for walking on level ground, slopes, and irregular terrain was above 99%. In the case of the stair climbing and descending tasks the GPDS achieved detection rate above 99% for able body subjects and above 96% for subjects with impaired gait. It is important to mention that the GPDS was challenged with a set of nonwalking tasks such as sliding of the feet, standing up and sitting down, and shifting weight during standing from one leg to the other. The experimental results have shown that the GPDS was very robust against such perturbations mainly due to the use of the gyroscope sensor which was insensitive to foot loading and unloading, but was sensitive to foot rotations in the sagittal plane. Another set of experiments showed that the GPDS detected the selected gait phases with the same reliability for all walking and running speeds from 0.5 to 13 km/h (fast jogging). This sensory system was tested in both indoor and outdoor environments and was shown that its detection performance did not depend on the ambient temperature that ranged from 0 °C to 25 °C. The experimental results also showed that the GPDS detected the gait phase events with a time delay that did not exceed 90 ms. The effect of the gait phase detection delays in a FES walking application yet need to be explored.

Besides its potential use in prosthetic and neuroprosthetic applications, the GPDS could be useful to analyze the subject's gait dynamics, to monitor or screen gait pathologies, to trigger physiological experiments in the field of human locomotion or as a biofeedback device that trains patients to improve their gait pattern. Currently our team is attempting to integrate the GPDS into a shoe insole and to manufacture it as a stand-alone system that can be interfaced with any data acquisition system or a prosthetic device.

Appendix I

A. Terminology

The following is the terminology used in this document for describing the gait phases:

stance phase: period when the foot is with its entire length in contact with the ground (angular velocity $= 0$);

heel-off phase: period following the stance phase during which the front part of the foot is in contact with the ground and its heel is not;

swing phase: period when the foot is in the air (not in contact with the ground) and swings forward;

heel-strike phase: period following the swing phase which begins with the first contact of the foot with the ground (usually the heel, but not necessarily) and which ends when the entire foot touches the ground;

walking cycle: period from one stance phase of the foot to the next stance phase of the same foot.

B. Rules for Reference Signal Generation

The following set of rules was used to generate the *reference* gait phase signal from the Vicon measurements. The heel marker reached its highest position at the beginning of the swing phase and it reached its lowest position at the initial contact of the heel with the ground (toes pointed upwards); see Fig. 4(b). The vertical position of the toe marker reached its first maximum at the end of the heel-off phase (in synchrony with the maximum vertical position of the heel marker), its second maximum was at the end of the swing phase, and it reached its lowest position during the stance phase.

stance → heel-off: The reference signal switched from the stance phase to the heel-off phase when the vertical position of the heel marker exceeded the threshold of 80 mm, i.e., 40 mm above its position during quiet standing, marked by the point A in Fig. 4(b).

heel-off → swing: The reference signal switched from the heel-off phase to the swing phase when the vertical position of the heel marker reached its maximum, marked by the point B in Fig. 4(b).

swing → heel-strike: The reference signal switched from the swing phase to the heel-strike phase when the vertical position of the heel marker reached its minimum, marked by the point C in Fig. 4(b).

heel-strike → stance: During the heel-strike phase the toe marker was continuously lowered until the foot was flat on the ground, i.e., until it reached the stance phase. The reference signal switched from the heel-strike phase to the stance phase when the toe marker stopped descending and settled around its minimum value, marked by point D in Fig. 4(b).

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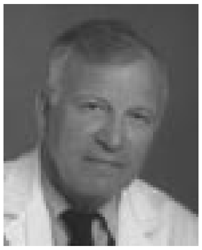

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