A Reliable Gyroscope-Based Gait-Phase Detection Sensor Embedded in a Shoe Insole

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

***Abstract*—This paper presents results of patient experiments using a new gait-phase detection sensor (GPDS) together with a programmable functional electrical stimulation (FES) system for subjects with a dropped-foot walking dysfunction. The GPDS (sensors and processing unit) is entirely embedded in a shoe insole and detects in real time four phases (events) during the gait cycle: stance, heel off, swing, and heel strike. The instrumented GPDS insole consists of a miniature gyroscope that measures the angular velocity of the foot and three force sensitive resistors that measure the force load on the shoe insole at the heel and the metatarsal bones. The extracted gait-phase signal is transmitted from the embedded microcontroller to the electrical stimulator and used in a finite state control scheme to time the electrical stimulation sequences. The electrical stimulations induce muscle contractions in the paralyzed muscles leading to a more physiological motion of the affected leg. The experimental results of the quantitative motion analysis during walking of the affected and nonaffected sides showed that the use of the combined insole and FES system led to a significant improvement in the gait-kinematics of the affected leg. This combined sensor and stimulation system has the potential to serve as a walking aid for rehabilitation training or permanent use in a wide range of gait disabilities after brain stroke, spinal-cord injury, or neurological diseases.**

***Index Terms*—Biomedical signal processing, gait-phase detection, gyroscope, locomotion, microcontroller, neuromuscular stimulation.**

I. Introduction

SINCE the early 1960s, it has been demonstrated by many research centers that functional electrical stimulation (FES) can be used effectively to assist individuals with walking deficiencies who have suffered damage in the central motor nervous system [1]. Such electrical stimulation systems consist of an electrical stimulator, which sends electrical stimulation pulses via self-adhesive surface electrodes or implanted electrodes to selected muscles of the leg. The applied electrical stimulation produces artificial contractions of the stimulated muscles during the gait cycle.

Crucial for the functional effectiveness is the correct timing of the applied stimulation within the gait cycle. The simplest practice to control the timing of the stimulations is by manually pressing a push-button once for each step [2]. Although very simple, this control method requires the subject's uninterrupted attention and coordination, and it results in an irregular synchronization with the gait cycle events and is practically limited to a single event indication per gait cycle. Therefore, to discharge the patient, several automatic triggering methods have been proposed based on different sensors systems ranging from simple foot switches placed in the shoe insole to inclinometers, goniometers, gyroscopes, accelerometers, electrodes for electromyography, and even implanted nerve cuff electrodes for afferent nerve signal recording [3]–[6]. Despite these efforts, the available triggering methods are still insufficiently reliable in everyday use and many patients who benefit from an FES system during their rehabilitation in the clinic, when they return home stop using it. Progress must, therefore, be pursued on two fronts: improving the functional outcome as well as the user-friendliness and reliability of the FES equipment.

In this paper, we present a new reliable real-time gait-phase detection system (GPDS) for FES walking applications which has been miniaturized to fit inside a shoe insole. An important practical aspect of a gait-phase detection system is that the system must be insensitive to disturbances caused by nonwalking activities. In daily activities, walking is interrupted by short nonwalking activities, such as standing, sitting, shifting the weight from one leg to the other, sliding of the feet, etc. It would be very impractical if the gait-phase detection system needed to be continuously turned on and off to avoid stimulations during nonwalking activities. Systems relying on force sensors alone or inclinometers attached to the shank do not comply with the above demand. Various designs have been proposed in the past. In particular, Skelly *et al.* [3] presented a rule-based gait event detector with fuzzy logic and concluded that two force sensitive resistors (FSRs) per insole are sufficient for gait event detection during walking. The robustness however to nonwalking activities (shifting the weight from one leg to the other) is questionable. Williamson *et al.* [4] reported excellent detection reliability by using three accelerometers attached to the shank and a machine-learning algorithm to detect in real time the transitions between five gait phases during walking, but no results have been presented for a use of this system with an FES system. The Salisbury Group (U.K.) has administered to several hundreds patients the Odstock dropped foot stimulator (ODFS). This stimulator is a single channel stimulator controlled by a simple foot switch placed usually beneath the heel for the correction of dropped foot in the last seven years [5]. The foot switch indicates the heel-off and the heel-strike phases. The subjects learn to keep the foot switch pressed when they stop walking in order to avoid false stimulation triggers.

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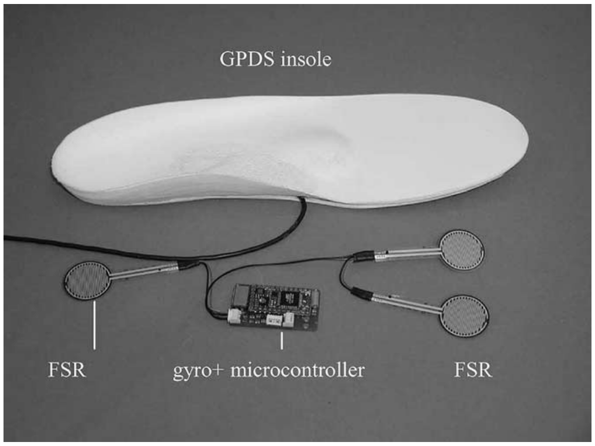

Fig. 1. Instrumented GPDS insole (white) and the embedded components displayed next to it.

After a short period of inactivity, the stimulator shuts down and must be turned on again to continue walking.

Our concept for the gait-phase detection was is based on the use of a miniature gyroscope sensor in addition to three force sensitive resistors placed on the shoe insole. In this paper, we present a new miniaturized gait-phase detection system, in which the force sensitive and gyroscope sensors, as well as the microprocessor have been entirely embedded in a shoe insole. We discuss its clinical use as walking neuroprosthesis in combination with a functional electrical stimulator. The system has been used in our clinic in combination with the Compex-Motion FES stimulator [8] to assist incomplete (hemiplegic) spinal cord injured subjects to improve their walking performance. Kirtley in [9] presented a similar instrumented insole with an embedded microcontroller, a gyroscope, five FSRs, a three-axis accelerometer, and a radio-frequency transmitter. The insole does not extract locally in real time the gait phases, but transmits the raw data to a PC, which analyzes the gait dynamics and kinematics.

II. Gait Phase Detection Sensor

A. Hardware

The GPDS is embedded in an anatomically shaped shoe insole (Bauerfeind AG. no. 212 70/5), shown in Fig. 1. For the detection of the four gait phases, the system relies on the combination of two types of "off the shelf" sensors:

1) three flat, circular (diam. 2.5 cm) force sensitive resistors (FSR Interlink El, Inc. 152NS), whose resistance changes nonlinearly with applied force;
2) a miniature gyroscope (ENC-03JA Murata, Japan, size $15.5 \times 8.0 \times 4.03$ mm, weight 1.0 gr), which measures the angular velocity of the foot in the sagittal plane.

The FSRs were fixed on the bottom side of the insole, one 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, in case of irregular ground or asymmetric (pathologic) gait style. The FSRs are not precision sensors (specified 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 cannot 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 gyroscope measures the angular velocity by sensing changes in the mechanical oscillations of a vibrating beam caused by the Coriolis force. The gyroscope signal was filtered by a third-order bandpass filter (0.25–25 Hz) with a 20-dB gain in the passband. The frequencies outside the passband were filtered out because they were not related to the walking kinematics. The filtered gyroscope signal was used to directly estimate the angular velocity of the foot and at the same time it was integrated to estimate the angle or inclination of the foot relative to the ground. Both the angular velocity and the inclination of the foot signal are used in separate parts of the gait-phase detection algorithm. 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. Thus, no calibration of the sensors was needed prior or during use. A detailed discussion about the gyroscope signal processing and the effect of varying ambient temperature on the gyroscope performance was presented in [10].

The miniature gyroscope and the band-pass filter was integrated together with a miniature microcontroller board on a small electronic circuit board (dimensions: $30 \times 49 \times 7.7$ mm), which was embedded in the anatomically shaped foot-arch of the insole, after some material was removed from this location using a carpet cutter. The sensing axis of the gyroscope was oriented perpendicular to the sagittal plane, in order to measure rotations of the foot in that plane. To protect the components of the circuit board from external loads, a slim custom aluminum housing was used as a cover. The FSRs were connected to the circuit board using flat micro-connectors (Molex, Inc.).

For the signal acquisition, processing and implementation of the gait-phase-detection algorithm, we used the low-cost BX-24 microcontroller board (NetMedia, Inc.), which contains a low-power ATMEL microcontroller with floating point math capability, 16 standard I/O pins, eight of which can be used as ADC with 10-bit resolution. The microprocessor can be programmed conveniently in Visual Basic. The sensor signals were sampled at a frequency of 100 Hz. To protect the FSRs and the electronic circuit board from direct contact with sharp or potentially hazardous objects, a protective thin plastic layer (1 mm) was glued to the bottom of the insole. The total thickness of the fully instrumented insole was less than 6 mm, at the heel and below the metatarsal bones. Twelve instrumented insoles were fabricated in different sizes: small, medium, large, and extra-large (36, 38, 41, and 45, respectively, in European sizes).

The detected gait phases, which were used to trigger FES sequences, were transmitted via a direct cable connection as discrete states from the embedded microcontroller to the programmable electrical stimulator Compex-Motion (Compex SA,

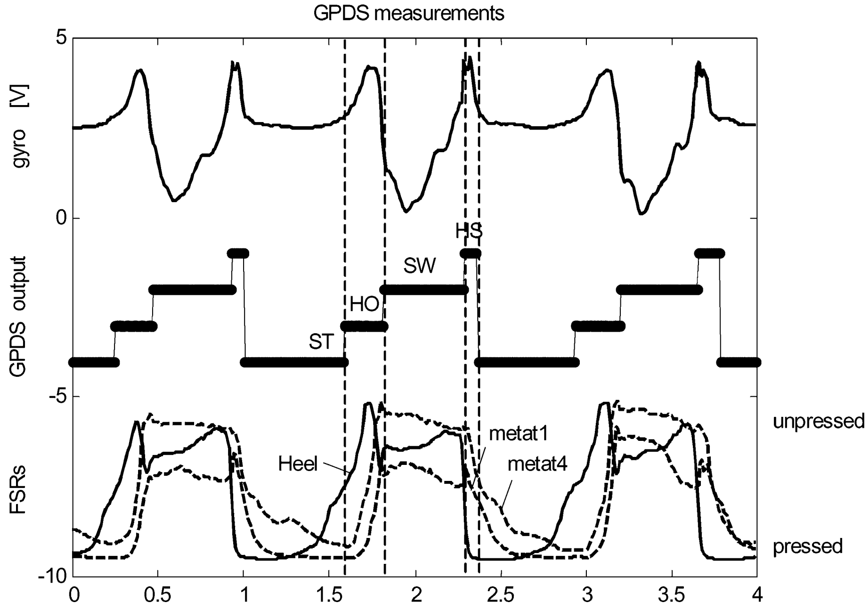

Fig. 2. Sensor signals and GPDS output during three steps. Top: gyroscope. Middle: GPDS. Bottom: FSRs. ST: stance. HO: heel off. SW: swing. HS: heel strike (figure lended from our previous publication [7], where the same sensors and processing algorithm were used).

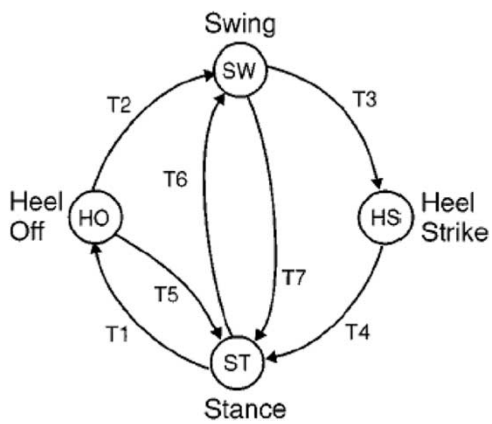

Fig. 3. GPDS divides the walking cycle into four gait phases: stance, heel off, swing, and heel strike. The arrows T1–T7 illustrate the possible transitions between the gait phases.

Switzerland) [8]. The same cable was also used to provide power from the stimulator to the GPDS.

B. Gait-Phase Detection Algorithm

The gait-phase detection algorithm detects in real-time transitions between the following four phases of the gait cycle: *stance, heel off, swing, and heel strike* (see Figs. 2 and 3). The loop frequency of the algorithm is 100 Hz, i.e., equal to the sensor sampling frequency. The algorithm[1] is a knowledge-based and rule-based algorithm, which allows a total of seven different transitions (T1-T7) between the four gait phases, as illustrated in Fig. 3. It was programmed in Visual Basic and runs on the BX-24 microcontroller, embedded in the insole.

C. Robustness of Algorithm

The gait-phase detection algorithm reliably identifies the transitions between *stance, heel off, swing*, and *heel strike* for a wide range of normal and pathological gait styles. The reliability of the gait-phase detection algorithm was evaluated by ten able-bodied subjects and six subjects with walking impairments in a preliminary experimental study presented in [7]. It was shown that the employed gait-phase detection algorithm was very reliable under very diverse walking conditions such as walking on flat and rough terrain (grass, earth, and snow), walking on inclinations and on stairs. The algorithm was also very robust against nonwalking actions such as shifting the weight from one leg to the other, sliding of the feet, standing up, and sitting down. 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 a detection rate above 99% for able-body subjects and above 96% for subjects with impaired gait. The performance of the GPDS was verified using a standard Vicon 370 (Oxford Metrics, Ltd., U.K.) or based on comparison of the GPDS output signal and raw sensor signals. The system was tested in both indoor and outdoor environments and it was shown that its detection performance did not depend on the ambient temperature that ranged from 0 °C to 25 °C. Additionally, the GPDS was tested at different walking and running speeds (0.5 to 13 km/h fast jogging) and detected the four gait phases with the same reliability.

III. Experimental Case Study

The purpose of the experimental study presented here was to quantitatively measure the benefit of the combined GPDS system and FES system, which is presented here for the first time. In particular, we were interested in determining if the timing of the stimulation is appropriate for different walking speeds and ground inclinations. Two subjects, with incomplete spinal cord injuries, participated in the study (level of lesion: T10 subject A; C6 subject B). As a result of their spinal cord injuries, the subjects suffered from a unilaterally dominated para-

[1]More details on the algorithm are given in [7].

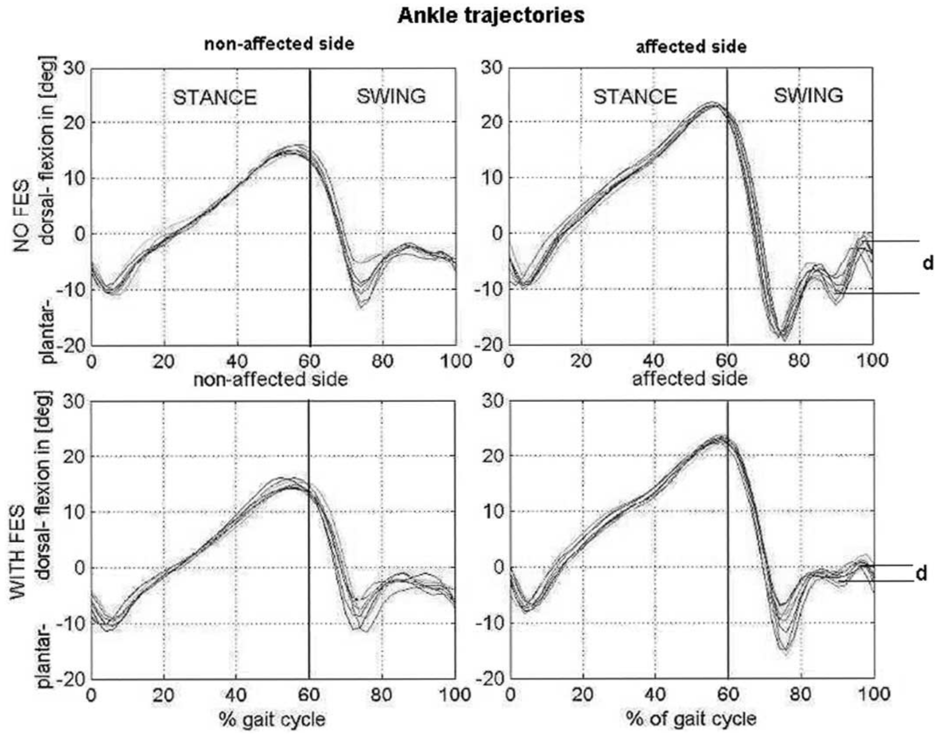

Fig. 4. Ankle trajectories of ten steps at 2.0 km/h. Left: *non*-affected side. Right: affected side. Top row: *without* FES. Bottom row: *with* FES.

plegia (one leg affected). Without FES, subject A could walk independently using crutches but with a typical dropped-foot gait pattern, while subject B due major deficits in hip and knee flexion used almost exclusively the wheel-chair in daily life. An informed consent to participate in this study was obtained from the subjects and the local Ethics Committee approved the experimental protocol.

The experiment consisted of the following tasks, which were carried out on a treadmill:

1) walking horizontally at three speeds: slow, normal, and fast. (subject A: 1.2, 2.0, and 2.5 km/h; subject B: 0.6, 0.9, and 1.4 km/h);
2) walking downhill (− 15% inclination), normal speed;
3) walking uphill (15% inclination), normal speed;
4) non-walking activities: shifting the weight from one leg to the other, standing up, and sitting down.

The "normal" walking-speeds were initially self-selected by the subjects and were then enforced by the treadmill speed. We also asked the patients to verbally express their opinion about the comfort/usefulness of the system. Subject A used two stimulation channels for a balanced dorsal flexion of the ankle (stimulated muscles: ext. dig. longus and peroneus tertius). Subject B used two stimulation channels for the stimulation of the peroneal nerve (to elicit the flexion reflex) and the activation of the tibialis anterior muscle. The stimulation started at the detection of the heel-off phase and terminated at the detection of the heel-strike phase. Both subjects had previous walking training with a manually or GPDS triggered FES system once a week during the previous three months, but had no particular muscle training.

The bilateral hip, knee, and ankle joint trajectories, as well as the foot clearance, were recorded using the optical motion analysis system Vicon (Vicon Motion Systems, Ltd., U.K.) with five cameras and fifteen reflecting markers, placed on standard body locations. Synchronously to the Vicon measurements, we recorded the output signal of the gait-phase detection system (GPDS) in order to measure the phase detection delay.

IV. Experimental Results

The combination of the GPDS and the FES system worked successfully and offered great functional benefit to both subjects, in all of the above-listed conditions [1)–4)].

A comparison of the joint trajectories of the affected and non-affected sides based on the Vicon measurements is presented in Fig. 4.

A. Joint Trajectories

Fig. 4 displays ankle joint trajectories obtained from subject A during ten sequential steps of horizontal walking. The two graphs on the top show the ankle joint trajectories *without* FES, the nonaffected side on the left and the affected side on the right. When comparing the both sides in the swing phase (∼ 60%–100% of the gait cycle), the foot's plantar flexion on the affected side is larger by almost 10°. Additionally, during the swing phase, an oscillation of the foot can be observed, which is explained by the paralysis of the distal leg muscles, which fail to stabilize the foot. The bottom graphs of Fig. 4 show the ankle joint trajectories during walking *with* FES. In the nonaffected side, no significant change is observed. On the affected side, two benefits can be noted. a) The excessive plantar flexion of the affected foot during the swing phase has been reduced to normal levels and b) the undesired oscillations of the foot during the swing have been greatly reduced, and, although not

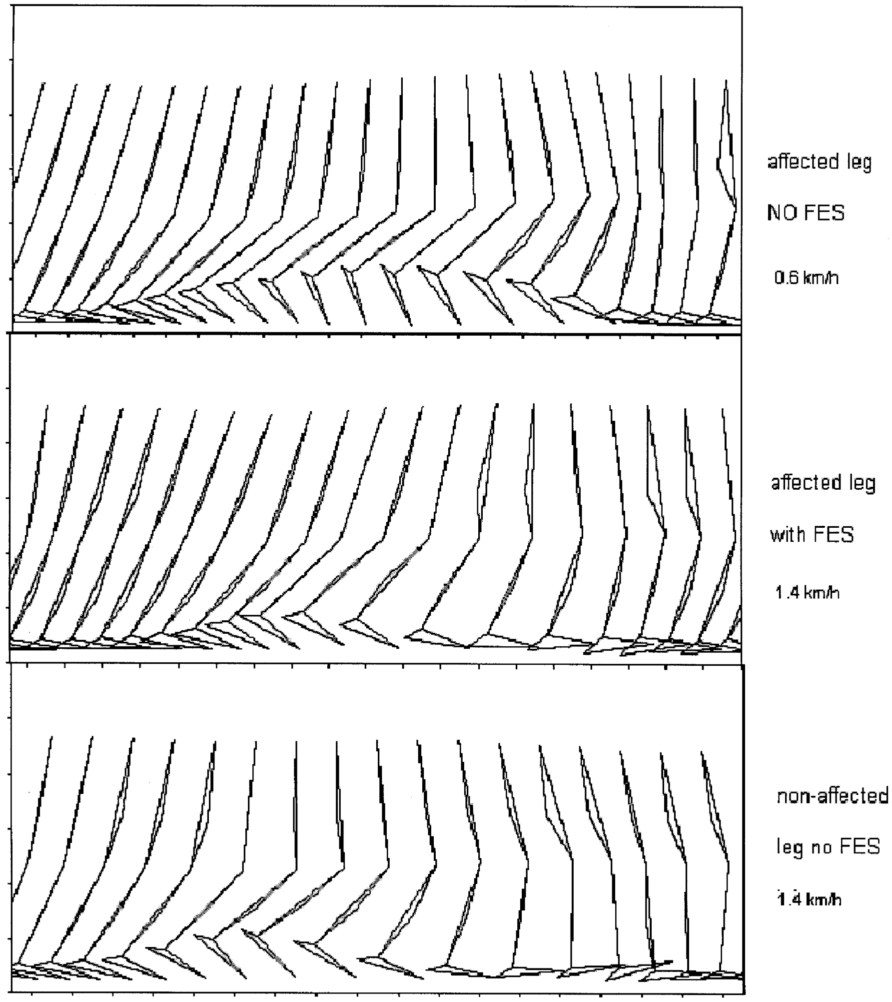

Fig. 5. Measured leg motion for subject B; top = no FES; middle = with FES; bottom = non-affected leg. The top graph shows that during the swing phase without FES there is excessive plantar flexion of the ankle and the clearance is small. In the middle graph the electrical stimulation reduces the excessive plantar flexion to normal levels and the clearance increases.

completely eliminated, their peak-to-peak amplitude is reduced from an average of 10° to 3°. This is explained by the FES-induced activation of the dorsal extensors of the foot. The comparison of the graphs shows that the application of FES leads to a greater similarity between the affected and nonaffected sides and therefore to an overall more "physiological" gait pattern.

For subject B, the application of the GPDS + FES system offered similar benefits. As shown by the illustration of Fig. 5, the application of FES reduced the excessive plantar flexion of the foot during the swing phase (from 10° to 0°) and provided a better clearance. Overall, the application of FES led to a quantitatively greater similarity of the affected and nonaffected sides and the subject could increase his comfortable walking speed from 0.6 to 1.4 km/h.

Both subjects reported that walking with the FES system was less tiring and safer than without FES. For both, the use of FES stimulation allowed them to significantly increase their walking speed (subject A: from 1 to 1.5 km/h; subject B: from 0.6 to 1.4 km/h). The subjects further expressed a clear preference for the automatic, GPDS triggered FES system in comparison to a manually triggered, which they had used earlier in our hospital, because, they reported, their mind and hands did not need to concentrate on giving precise trigger timing. Most importantly, the GPDS did not generate false triggers during standing, shifting the weight from one leg to the other, standing up or sitting down, where simpler systems consisting only of force sensors would fail. Thus, the subjects did not need to worry about turning off and on the system every time they stopped or started walking. The gait-phase detection delay of the GPDS (< 70 ms) proved to be sufficiently small for the above-described applications and walking speeds.

V. Discussion and Future Directions

The positive results obtained in the above-described experiments and our clinical experience indicates that the GPDS provides increased comfort to the FES user. The stimulation being triggered automatically leaves the hands and the mind free to be used for other things. The fact that the GPDS works robustly on flat, rough, or inclined terrain and the fact that it does not generate false triggers increases the comfort and the confidence the user has in the system. Faster walking speeds overcoming double-stance phase directly result from it. We hope that in the near future, the GPDS system may be commercialized[2] and may

[2]Further information at http://control.ee.ethz.ch/~fes/.

gain wide acceptance among the experts for FES walking applications.

Future developments are expected to continue in the following directions: adding wireless communication between the instrumented insole and the functional electrical stimulator, studying long-term effects on patient walking performance, and, finally, using the gait-phase signal and the gyroscope signal not only as a trigger to time pre-programmed stimulation sequences, but as a direct feedback signal in a closed-loop FES control scheme. As suggested by Veltnik *et al.* [11], two orthogonal gyroscopes could serve to control the plantar- and dorsi-flexion as well as the eversion and inversion of the foot.

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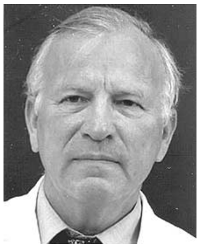

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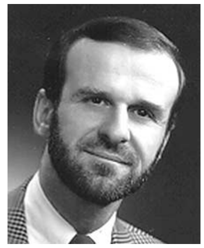

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