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An active foot lifter orthosis based on a PCPG algorithm

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Abstract— **Central pattern generators (CPGs) are known to play an important role in the generation of rhythmic movements in gait, both in animals and humans. The comprehension of their underlying mechanism has led to the development of an important family of algorithms at the basis of autonomous walking robots. Recently, it has been shown that human gait could be modeled using a subclass of those algorithms, namely a Programmable Central Pattern Generator (PCPG).**
In this paper, we present a foot lifter orthosis driven by this algorithm. After a learning phase, the PCPG is able to generate adequate rhythmic gait patterns both for constant speeds and acceleration phases. Its output is used to drive the orthosis actuator during the swing phase, in order to help patients suffering from foot drop (the orthosis just follows the movement during the stance phase). The most interesting property of this algorithm is the possibility to generate a smooth output signal even during speed transitions. In practice, given that human gait is not perfectly periodic, the phase of this signal needs to be reset with actual movement. Therefore, two phase-resetting procedures were studied: one standard hard phase-resetting leading to discontinuities and one original soft phase-resetting allowing to recover the correct phase in a smooth way. The simulation results and complete design of the orthosis hardware and software are presented.

I. Introduction

Accidents, wars or diseases are the main reasons for loosing the total control of a leg. For ages, several types of leg prostheses/orthoses have been developed in order to help disabled people and enhance their daily life. The main objective of these prostheses/orthoses is to allow their user to walk as naturally as possible considering the huge complexity of human gait. Active systems have recently become available on the market. Contrary to standard passive systems, these are equipped with self-propulsion capabilities leading to an enhanced locomotion speed, much less fatigue and overall a better comfort of the patient.
Two main categories of active prostheses exist to date. In the first category, a sensing system analyzes the motion of a specific part of the patient's body (e.g. the healthy leg, the upper-body or the hip). Using this kinematics information, the control system identifies the phase of the gait cycle and then adequately triggers an actuator [1]–[5]. The movement of the second type of active prostheses (or orthoses) is guided by myoelectric signals recorded at the surface of the skin, just above the muscles involved in gait [6]–[8].

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Although active prosthetic technology really offers important improvements, the non-natural gait problems of the patients are only partially solved. Indeed, active prosthetic systems seldom adapt to the specificities of the patient's gait and, more specifically, to variations of his walking dynamics at constant or varying speeds. This leads to non energy efficient systems and oblige patients to compensate for these imperfections while walking.

In this general context, we propose to develop an active prosthesis/orthosis system which is inspired by real walking mechanism. Actually, human locomotion is known to be based on a very complex hierarchical system which includes several control networks located at spinal and supraspinal levels [9]. In broad outline, high-level commands are sent by the brain to Central Pattern Generators (CPGs), composed of interneurons and motoneurons located in the spinal cord, which generate periodic gait patterns at the desired speed. This basic principle has widely inspired the field of robotics, particularly in the development of small autonomous walking robots, from multi-legged insect-like robots to humanoids [10].
One of the algorithms developed in this framework is called Programmable Central Pattern Generator (PCPG) [11]. Basically, a PCPG is able to learn and subsequently reproduce any given rhythmic gait pattern. The interest of this particular CPG is twofold: firstly, the parametrization learning is simple and secondly, the learned parameters (i.e. the pattern magnitude and frequency) may be adjusted in order to generate different speeds. During these speed changes, the output signals sent to the walking robot actuators present smooth transitions, which is particularly important for practical applications.

The aim of this paper is to expose the principles and a proof of concept of integrating a human gait model in a foot lifter orthosis using such PCPG algorithm. Noticing that foot kinematics is known to be the most complex signal to model in human locomotion, this work can be easily extended to full prostheses in its principles. In Section 2, the bases of Programmable Central Pattern Generators to learn a gait pattern are detailed. In section 3, the orthosis design and the global control strategy based on a two mode system are explained. In Section 4, the practical needs about frequency estimation and phase-resetting problems when applying this pattern to real orthoses are exposed and solved. In Section 5, most relevant results are displayed in simulation. Those results are available in more details in the enclosed video.

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II. Human Gait Modeled by PCPG

This section aims at describing the PCPG principles to model gait as well as the way to extend the current study of ankle orthosis to full prostheses.

A PCPG is able to learn a standard gait pattern. This standard gait pattern was obtained by averaging about 50 gait cycles performed on a treadmill, determined and synchronized by a peak detection algorithm able to locate all the relevant maxima in the kinematics recordings. Ideally, the standard pattern could be derived from the subject himself before his accident or, more probably, from a similar subject in terms of his gait parameters such as age, height, etc [5]. As defined in [11], a PCPG is a kind of Fourier series decomposition and is composed of several adaptive oscillators. This algorithm is governed by the following equation system:

$$\begin{cases} \dot{x}_i = \gamma(\mu - r_i^2)x_i - \omega_i y_i + \epsilon F(t) + \tau_i sin(R_i - \phi_i) & (1) \\ \dot{y}_i = \gamma(\mu - r_i^2)y_i + \omega_i x_i & (2) \\ \dot{\omega}_i = -\epsilon F(t)\dfrac{y_i}{r_i} & (3) \\ \dot{\alpha}_i = \eta x_i F(t) & (4) \\ \dot{\phi}_0 = 0 & (5) \\ \dot{\phi}_i = sin(R_i - sgn(x_i)cos^{-1}(-\dfrac{y_i}{r_i}) - \phi_i), \forall i \neq 0 & (6) \end{cases}$$

with

$$R_i = \frac{\omega_i}{\omega_0} sgn(x_0)cos^{-1}(-\frac{y_0}{r_0}) \tag{7}$$

and

$$F(t) = P_{teach}(t) - \sum_{i=0}^{N} \alpha_i x_i \tag{8}$$

As depicted in Figure 1, oscillators are coupled between each other. The instantaneous phase of the fundamental oscillator R_0 is scaled at the frequency ω_i through R_i and the phase difference with the fundamental oscillator is given by ϕ_i. In the standard approach, the coupling constant τ_i is chosen constant over all oscillators. They are each composed of one adaptive magnitude coefficient α_i and one frequency parameter ω_i. μ has a role of normalization of the learned pattern with respect to $r_i = (x_i^2 + y_i^2)^{\frac{1}{2}}$. The other parameters γ and ϵ aim at accelerating the convergence while limiting stability problems [11]. The $Q_{learned}(t)$ signal resulting from the sum of oscillator outputs is compared to the $P_{teach}(t)$ gait pattern target and the error value $F(t)$ is computed. Throughout the learning step consisting in integrating the differential equations by a 4th order Runge-Kutta method with a fixed step size, all the parameters of the PCPG are modified in order to minimize $F(t)$ (the typical learning time is less than five minutes on a standard laptop). When this learning step is finished, $F(t)$ is close to zero and the system is generating the right pattern as depicted in Figure 2.

Properties of PCPGs make them suitable for trajectory generation in robotics and also for prosthesis applications. In fact, the pattern learned by a PCPG can be easily controlled in magnitude and in frequency thanks to a simple linear

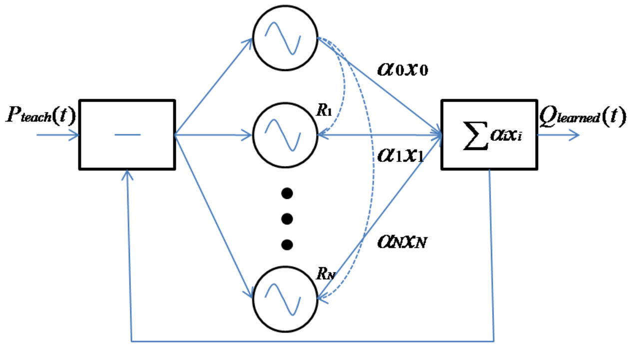

Fig. 1: The PCPG is able to learn the frequency components of a periodic signal as well as the various phases and magnitudes. The main interest of PCPGs is the possibility to modify a learned pattern in amplitude or frequency in a smooth way. This Figure is inspired from [11].

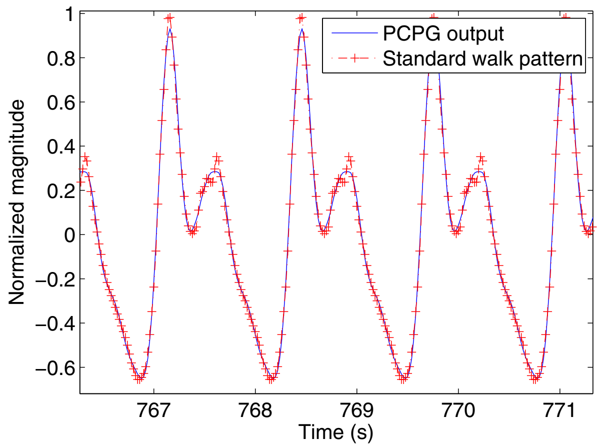

Fig. 2: The PCPG is able to learn quasi-perfectly an average normalized pattern of foot relative angle by means of 5 oscillators determined by the frequency complexity of the signal.

change of the $\vec{\omega}$ and $\vec{\alpha}$ vectors representing the $\Re^N$ PCPG states (N is the number of oscillators). This linearity leads to a smooth change of the global system behavior. For instance, if the $\vec{\omega}$ vector is divided by two, the underlying frequency of the standard temporal pattern is divided by two. The same effect occurs for the $\vec{\alpha}$ vector, which governs the pattern magnitude.

Finally, as proposed in [12], it is possible to couple several PCPGs to model different joint angles. This is performed thanks to equations of coupling between the fundamental oscillators of each PCPG and by learning the phase difference:

$$\begin{cases} \dot{x}_{0,k} = \gamma(\mu - r_{0,k}^2)x_{0,k} - \omega_{0,k}y_{0,k} + \tau sin(R_{0,k-1} - \phi_{0,k}) & (9) \\ \dot{\phi}_{0,k} = sin(R_{0,k-1} - R_{0,k} - \phi_{0,k}) & (10) \end{cases}$$

where $(0, k)$ denotes the first oscillator of the kth PCPG (frequencies of joint angles are the same).

III. Orthosis design and control

This section first describes the orthosis hardware and design. Then, the focus is on the gait cycle control strategies.

A. Orthosis hardware

As shown in Figure 3, the orthosis is made of several components: two custom-fit plastic shells, two commercial flexure joints, a linear actuator, a ball-link transmission, a load cell to measure the actuator force, and two force sensors installed in the orthosis sole, under the heel and the toes. The plastic shells were designed using a 3D scan of the right foot and leg of a healthy subject, adding mounting surfaces for the actuator, the flexure joints, and the mechanical transmission. The actuator includes a position control unit based on a PID controller that can be driven by an external analog signal in the range of 0 to 10 V.
However, this first prototype is quite cumbersome. One of the problems that still represent an obstacle in the development of active orthoses is the weight of the commercial actuators. For this preliminary prototype, a commercial actuator including the motor, the transmission, and the control electronics was the first choice to facilitate the construction in the shortest time. The weight of the actuators found in the market that satisfied the mechanical requirements for developing a complete gait cycle was above 3.5 kg. In order to reduce the actuator weight, the selection was done according to an estimate of the root-mean-square torque during the swing phase and the maximum velocity developed by the ankle in one gait cycle, considering the maximum intrinsic ankle stiffness and damping coefficients. The chosen actuator weight is about 1.6 kg and its maximum power is around 117 W, which corresponds to a third of the peak power developed by a healthy ankle [13]. In the current stage of the project, a lightweight custom-fit actuator with passive energy-storage elements is being developed, powerful enough for the stance phase as well. Control strategies specific to an active stance phase are explored in parallel. These developments will be included in future prototypes.

The control and the PCPG algorithms reside in a DsPIC 30F4013 microcontroller running at 120 MHz. Both algorithms are calculated at each time step at a sampling frequency of 500 Hz but the output of one or the other is chosen according to the orthosis state, as detailed in the next section. The differential equations of the PCPG are solved by a simple explicit Euler integration method. The microcontroller manages three analog (two sole force sensors and one load cell) inputs for the signals coming from the force sensors. The position command is generated using Pulse Width Modulation (PWM) with a 9-mV resolution after amplification. Initialization commands for the actuator are generated using four digital outputs whereas the actuator state (home sequence completion) is monitored using one digital input.

A typical gait cycle for ground-level walking has a fundamental frequency around 1 Hz. To minimize the effects of noise coming from the sensors and to smooth the command signal, the analog inputs and the PWM output are filtered by low-pass first-order filters with a cutoff frequency of 15 Hz, which mainly determines the bandwidth. The microcontroller architecture is shown in Figure 4.

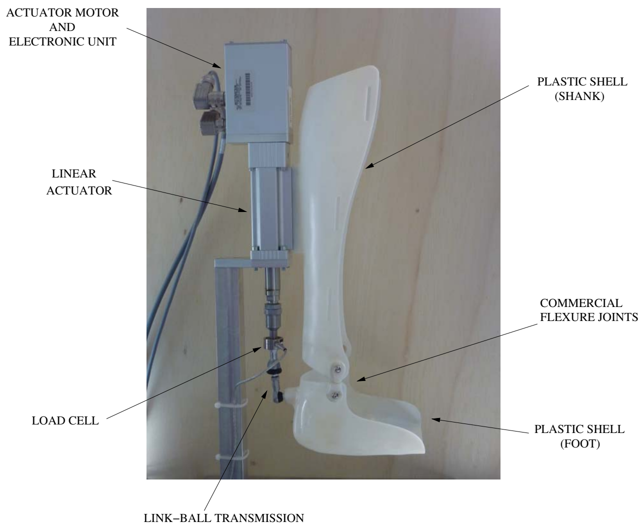

Fig. 3: The orthosis prototype is made of several important components.

B. Gait cycle control strategies

In gait, there are two main events: the Heel Strike (HS) and the Toe Off (TO) for each foot, which correspond to the time of the initial contact of the heel with the ground and the time of the last contact of the toes respectively. There are also two different phases: the stance phase, i.e. between the HS and the TO, and the swing phase, i.e. between the TO and the HS when the foot is in the air. As the objective of the orthosis in this preliminary stage is to help people with foot drop problems, the orthosis comprises two different control modes, one for the stance phase when the subject entirely drives the orthosis and another one for the swing phase when the PCPG output governs the system.

The first mode is active during the stance phase, allowing the free motion of the foot around an equilibrium point (approximately at a foot angle of $90°$ with respect to the tibia) but, at the same time, providing a certain level of stability through a virtual stiffness element. This part of the control algorithm is based on the following expression

$$\dot{x} = c\,(f - k\,(x - x_r)) \tag{11}$$

where $\dot{x}$ is the time derivative of the actuator position, denoted by x; $c = 6.6\frac{\mathrm{mm}}{\mathrm{Ns}}$ is the proportional gain; f is the actuator force, measured by the load cell; $k = 3.8\frac{\mathrm{N}}{\mathrm{mm}}$ is the desired virtual stiffness; and $x_r = 39.1$ mm is the desired equilibrium position. The actuator possesses an internal controller that regulates the rod position taking an external

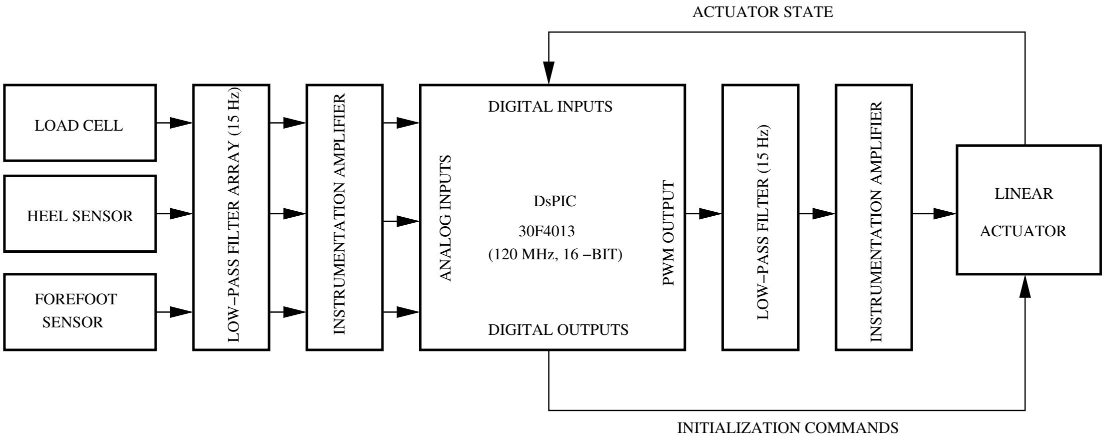

Fig. 4: The block diagram shows the microcontroller architecture.

analog signal as a reference. In our scheme, x represents this external signal. The internal controller has a relatively fast response considering the time scale in which a normal stride develops so the displacement command x is assumed to be equivalent to the actual actuator displacement. Because of the stable linear system structure, if f remains bounded, the stability at the output x is insured. A stability analysis of the whole system and tests using the foot sensors are part of future work.

The second mode is associated to the swing phase and is intended to help the patient to achieve enough foot clearance to initiate the next gait cycle. As a preliminary approximation, this part of the algorithm will mainly consist in a trajectory tracking scheme to follow the PCPG position pattern similar to that developed by a healthy foot during the swing phase. This tracking is performed by the internal controller.

IV. Practical needs

This section gives more details on the needs for frequency adaptation of the PCPG output and phase-resetting in order to deal with practical problems. Firstly, to adapt the gait pattern frequency at each step, the frequency estimation method is based on gait events and a simple method to correct the PCPG frequency shift is proposed. Secondly, it introduces two approaches of phase-resetting: a hard one and a novel scheme to achieve a smooth and robust adaptation.

A. Frequency adaptation

Given that the only available information is the TO and HS events, they have to be used to determine the gait period. In this paper, this period estimation T_{est} is obtained by the time between two successive HS of the same foot. To consider when the subject is stopping walking, if the next HS does not appear after a predetermined time T_{max}, the orthosis switches to the follower mode. Because the PCPG has to be controlled in angular frequency, the estimated angular frequency is given by $\omega_{est} = \frac{2\pi}{T_{est}}$. For easier frequency control purpose, the Normalized Frequency Control (NFC) parameter F_ω is obtained by $F_\omega = \frac{\omega_{est}}{\omega_0}$ where ω_0 is the fundamental angular frequency of the learned standard pattern.

By experiments, it was noticed that the observed frequency at the output of the PCPG F_{obs} and the command frequency F_{com} were not equal, probably due to the integration method (Euler method is known to modify the period of integrated signals). A linear regression was sufficient to perfectly model this frequency shift:

$$F_{obs} = 1.0294 * F_{com} - 0.04 \tag{12}$$

This easily leads to:

$$F_\omega = \frac{(\omega_{est} + 0.08 * pi)}{(1.0294(\omega_0))} \tag{13}$$

B. Phase-resetting

Obviously, at constant speed, gait cycles are not perfectly identical [14]. This fact and numerous perturbations can provoke phase mismatch between the perfectly periodic PCPG output and the real gait pattern in addition to change in frequency. If this mismatch is too important, the subject has to compensate for it leading to a non-natural gait. The aim of this phase-resetting is to pave the way to allow the orthosis to adapt to the patient as quickly and smoothly as possible aiming at increasing the subject comfort.
The phase-resetting consists in resynchronizing the PCPG state according to special events. Therefore, the PCPG will be phase reset on the HS to allow the system to recover the correct phase in a smooth way at the time of the TO. Two approaches are available: a *hard* and a *soft* phase-resetting.

The hard phase-resetting relies on a direct modification of the integrated values: in each oscillator i, x_i and y_i are put to standard values corresponding to the HS event. The main advantage of this approach is the quick phase-locking whereas the disadvantages are (1) a more sensitive reaction to noise in the frequency estimation due to small variations

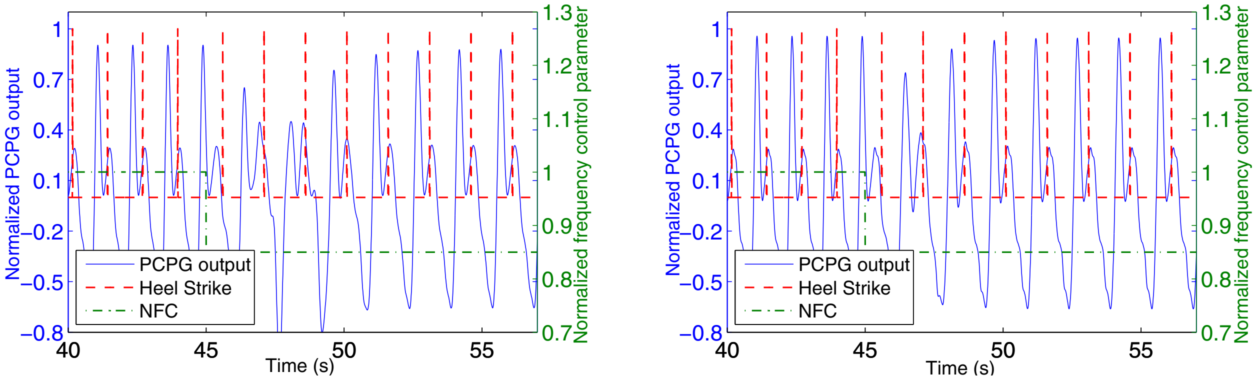

Fig. 5: On the left: without enhancements, the soft phase-resetting leads to an important and long transient. On the right: this problem is strongly mitigated.

in gait cycles at constant speeds or in the measurement itself and (2) important modification of the actuator state leading to a delay of recovery because of the low-pass filter. In the case of a foot lifter orthosis, during the stance phase, the actuator is not commanding the system and thus, the latter disadvantage is mitigated. However, it is a real problem in the case of a complete prosthesis.

In the soft phase-resetting, the original PCPG algorithm was slightly modified. To control the phase of the first PCPG oscillator, a coupling with a reference oscillator at instantaneous phase $R_{0,r}$ was established. This allows to modify the phase difference $\phi_{0,r}$ between the reference oscillator and the first oscillator of the PCPG. Formally, the reference oscillator is as follows:

$$\dot{x}_{0,r} = \gamma(\mu - r_{0,r}^2)x_{0,r} - \omega_{0,r}y_{0,r} + \tau sin(R_{0,r}) \quad (14)$$

whereas the coupling with the PCPG (subscripted by p) is shown in:

$$\dot{x}_{0,p} = \gamma(\mu - r_{0,p}^2)x_{0,p} - \omega_{0,p}y_{0,p} + \tau sin(R_{0,r}k - \phi_{0,r}) \quad (15)$$

where $k = \frac{\omega_{0,p}}{\omega_{0,r}}$. The coupling with the other oscillators of the PCPG is identical to the previous description. Because the phase of higher order oscillators had more difficulties to follow a phase change in experiments, coupling constant was defined as $\tau_i = \tau \frac{\omega_{i,p}}{\omega_{0,p}}$. Figure 5 shows how this modification can produce a smooth and robust kinematics output when a phase reset is applied. In our experiments, we chose $k = 1$.

V. Results

In this section, results about the system response to two realistic perturbations for both phase-resetting techniques are presented. These were obtained using the microcontroller connected to the orthosis (the PCPG output shows the command of the orthosis, not the actual position of it). Gait events were generated by simulation. First, a steep speed change is considered. Then, a speed change according to a chirp function is described in order to expose the behavior of each technique to a highly changing environment.

To model a rapid change of frequency, i.e. of the subject's speed, the heel strike period is suddenly modified. To be coherent with practical applications, speed is instantaneously decreased by 1 km/h from the speed of 4 km/h (from $F_\omega = 1.13$ to $F_\omega = 1$). As depicted in Figure 6, the hard phase-resetting is recovering the correct phase quasi-instantaneously at the price of a non-continuous modification (with a small transient due to the low-pass filtering when connected to the orthosis) whereas the soft phase-resetting takes much more time for recovering (less than two gait cycles) with a very smooth transition. Similar results are obtained when the speed is suddenly increased.

To show the behavior of both techniques in quasi-continuous speed adaptation, a chirp function was used to decrease the normalized frequency control parameter F_ω. This parameter was decreased from 1.3 to 0.7 by 0.1 step every two heel strikes of the same foot, which corresponds to a deceleration from of 6 km/h to 1.5 km/h in 12 gait cycles for the studied subject. As depicted in Figure 7, at first, when the frequency is changed, the system can not detect this information and is not adapting leading to a certain phase delay depending on the frequency difference. Then, at the following heel strike, the system detects the correct phase and frequency. Regarding the hard phase-resetting, the phase adaptation is done quasi-immediately as in the steep speed change whereas the soft resetting method takes more time and does not converge quickly enough to totally recover the phase when the next modification occurs. The same conclusion about the discontinuities can be reported. In a smooth approach, our frequency estimation scheme implies an intrinsic delay, namely the system can not predict the actual speed before the HS. Enhancements could be brought by using for instance additional gait events or sensors.

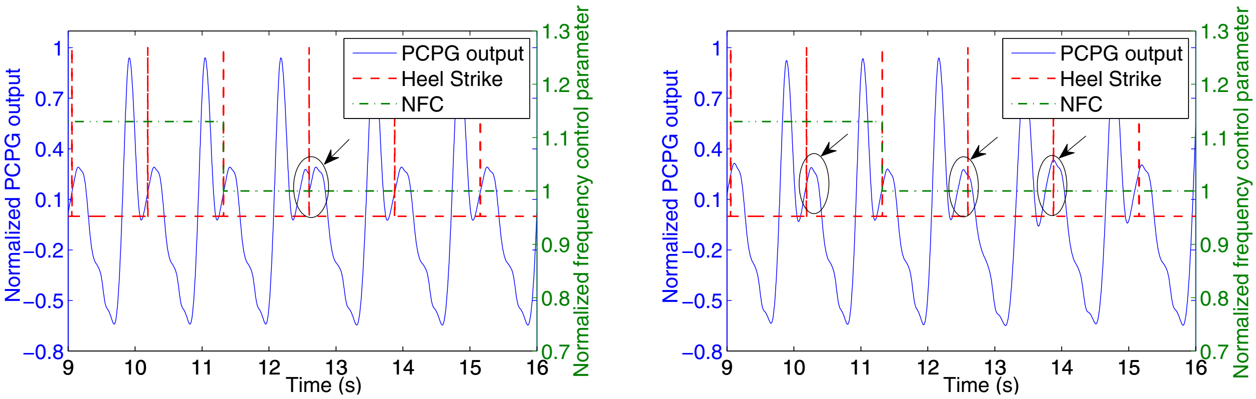

Fig. 6: On the left: the hard phase-resetting is able to quasi-immediately recover the correct phase at the price of discontinuities shown by the arrow. This clearly leads to be more sensitive to frequency estimation noise. On the right: the soft phase-resetting is recovering the phase quite slowly but in a perfectly continuous way. Indeed, the first arrow on the left is showing the correct place of the HS event. The next ones indicate that there is a decreasing phase offset.

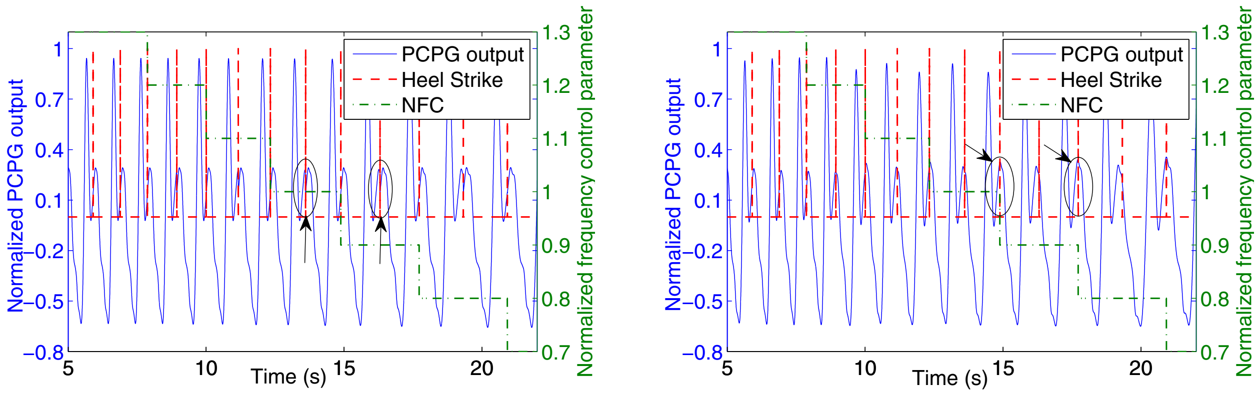

Fig. 7: On the left: the hard phase-resetting is working as in the steep speed change. On the right: the soft phase-resetting does not totally recover the right phase when a speed change occurs as indicated by the arrows. However, the phase delay is quite small.

VI. Conclusion and Future Work

A. Conclusion

In this paper, the fundamentals and principles of an original foot lifter orthosis integrating a human gait model are exposed. The system is composed of three main parts: a PCPG-based model of human locomotion, a sole sensor-based gait control switching strategy and a heel strike-based phase-resetting approach.

Modeling human gait by PCPG is greatly facilitated by its controllability in frequency and magnitude. Moreover, smooth transitions produced by this model are highly relevant for this purpose.
A twofold gait control strategy was developed. Sole sensors provide information about gait events. When the stance phase is detected, the follower mode allowing the patient to entirely govern the orthosis is activated. Otherwise, the PCPG human model is fed in the orthosis, which aims at mimicking as closely as possible the gait pattern and increasing the patient's comfort.

Two phase-resetting methods are detailed and their advantages and disadvantages are discussed. A hard version consists in directly modifying the values of the PCPG at the heel strike event, which could lead to discontinuities in the pattern. This problem can be mitigated in a foot lifter orthosis because of the specific control strategy (during the stance phase, after a HS, the orthosis is in the follower mode) but has to be improved in full prostheses. Therefore, a second and original soft phase-resetting scheme based on a reference oscillator and a harmonic-adaptive coupling constant is described leading to a smooth PCPG output transition. This scheme is obviously more robust to slight frequency estimation variations at constant speed. However, when based on heel strike events, it is at this time not sufficiently rapid to recover the phase.

B. Future work

Future work will be dedicated to increasing the speed of phase recovery in the soft phase recovering while keeping the smooth aspect. In addition to parameter optimization, this could be done by using other gait events or sensors in both frequency estimation and phase-resetting. An alternative could be to combine the hard and soft phase-resettings to increase reactivity while remaining quite smooth. For instance, heel strikes of the injured foot could be used with hard phase-resetting, while toe off of the injured foot could be used with soft phase-resetting to slowly recover the phase and facilitate the hard phase-resetting work. This procedure would avoid steep magnitude modification just before the injured foot leaves the ground.

From experimental data, the evaluation of metrics such as the settling time, i.e. the time the system needs to recover the phase given a certain error band, could be interesting to precisely characterize the recovery speed of soft phase-resetting. When combining both phase-resettings, the determination of the hard phase reset step distribution with and without soft phase-resetting in realistic application will make it possible to judge the relevancy of this combination to reduce the magnitude of the hard phase-resetting step.

Pattern magnitude is known to vary according to the speed. Such magnitude adaptation could be envisaged aiming at further increasing the patient's comfort [15]. Similarly, the frequency- and magnitude-adjusted gait pattern is also known to slightly differ as a function of speed. As proposed in [15], this could be solved by combining two PCPGs whose learned frequencies are quite different and correspond to different ranges of gait speeds.

Regarding the orthosis itself, energy-storage elements powerful enough for the stance phase will be implemented. Control strategies specific to an active stance phase are explored in parallel. Also, a main point of interest will be the inclusion of a lightweight custom-fit actuator with passive energy storage elements.

Finally, the feedback from the patients will make it possible to correctly design the final orthosis.

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