Control and design of an active prosthesis based on adaptive oscillators

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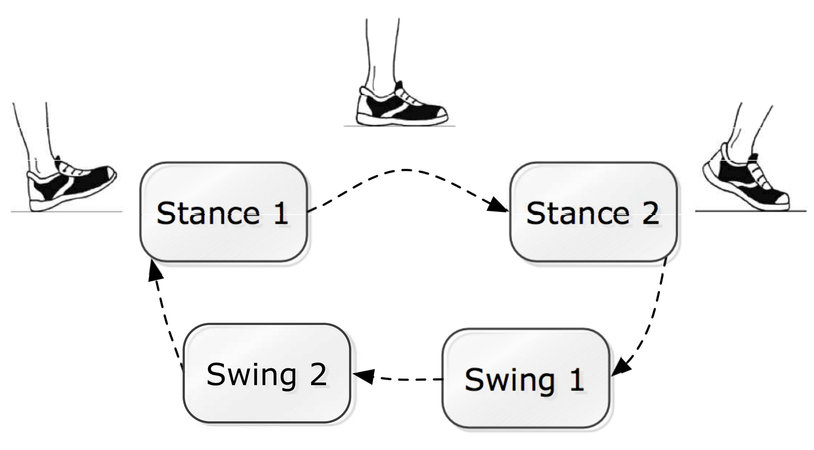
Introduction

Current transfemoral prostheses are almost exclusively passive such that the user spends up to 60% of additional metabolic energy with respect to a healthy person. This contribution introduces the design of an active ankle prosthesis whose controller is based on an adaptive oscillator (AO).

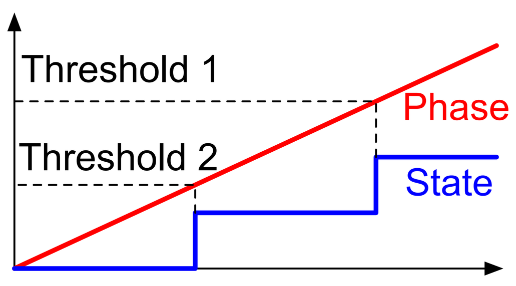
Control Method

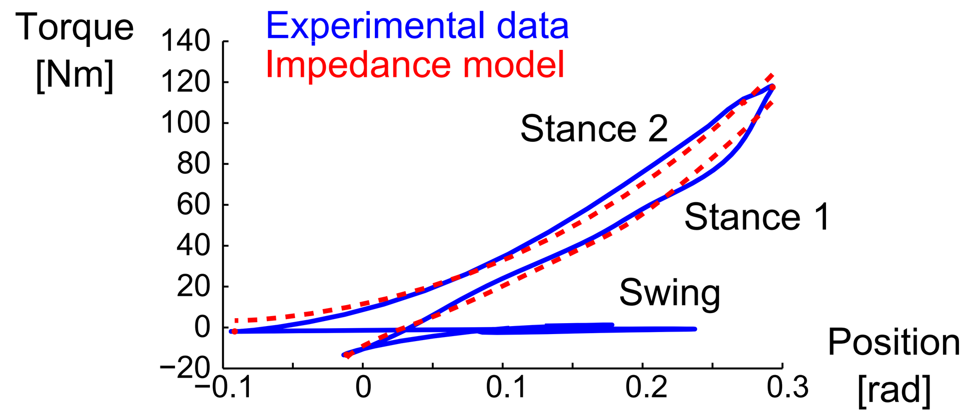


The gait is separated in four states each obeying a different impedance law to match the torque/position relationship of a healthy ankle.

$$\tau = k_1(\theta - \theta_e) + k_2(\theta - \theta_e)^3 + b\dot{\theta}$$

The transition between different states is triggered when the phase of an oscillator reaches a given threshold.



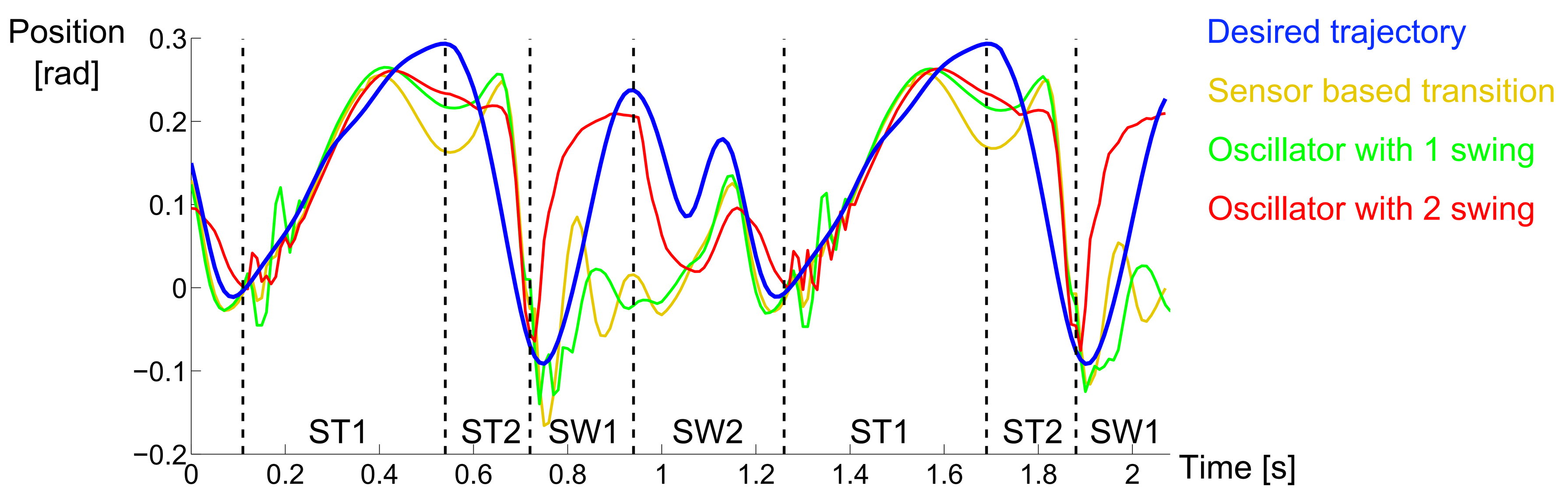


Our controller has two different adaptation mechanisms:

- **Continuous adapation** of the oscillator frequency to the user gait frequency using an adaptive oscillator.
- **Discrete adaptation** of the thresholds to the user gait profile using foot sensors.

Simulation results

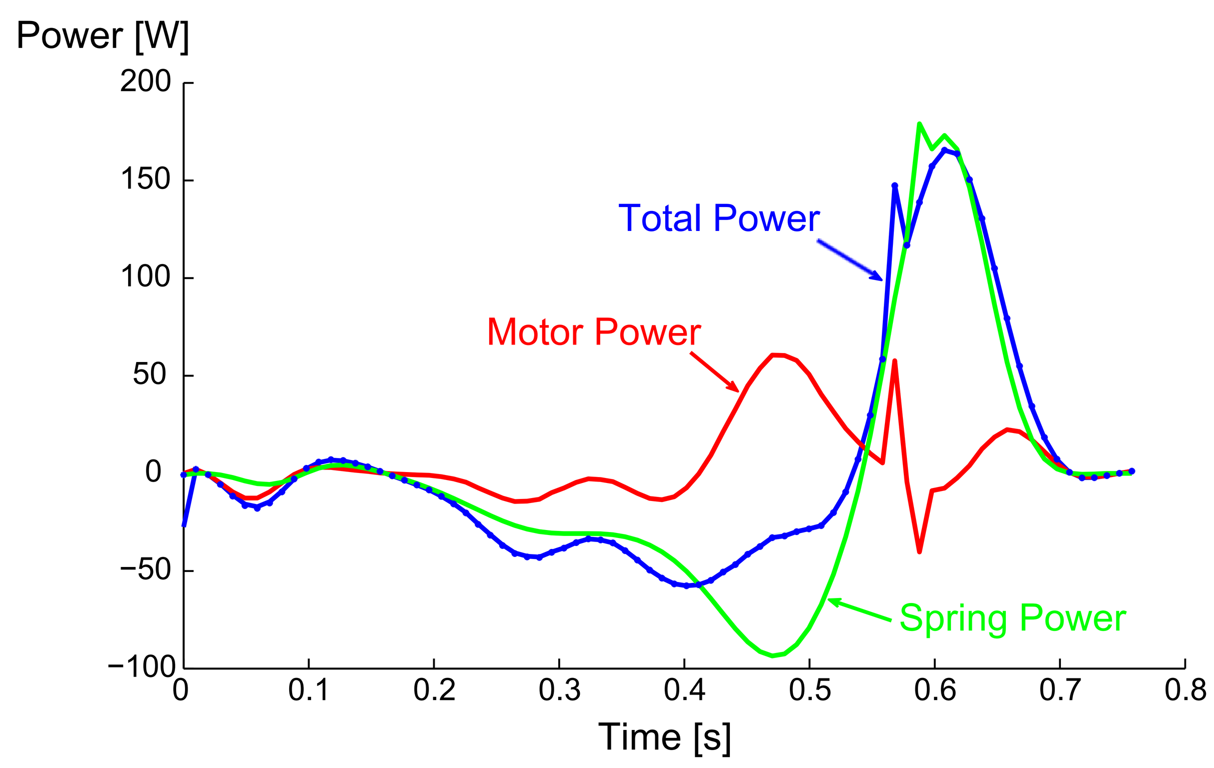
Simulations using Webots compared our controller to a more classical finite state machine. It showed that the fine threshold placement made the torque profile smoother as compared to the FSM-based controller.



Moreover, the mean square error between the real and desired ankle trajectory decreased by 45% and the foot clearence was significantly improved.

Prototype

To test our controller, we developed a prototype exploiting the principle of series elastic actuator (SEA).



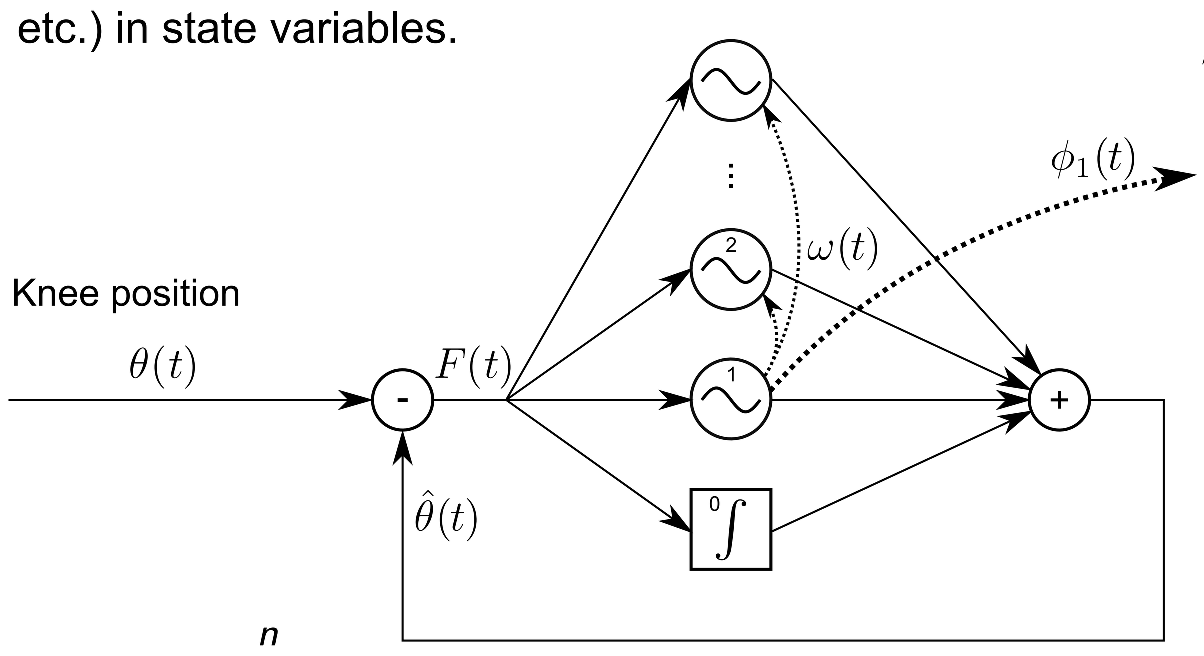
SEA consists in introducing a compliant element (e.g. a spring) between the actuator and the load (i.e. the foot). As such, the spring stores energy during early stance and releases it during late stance.

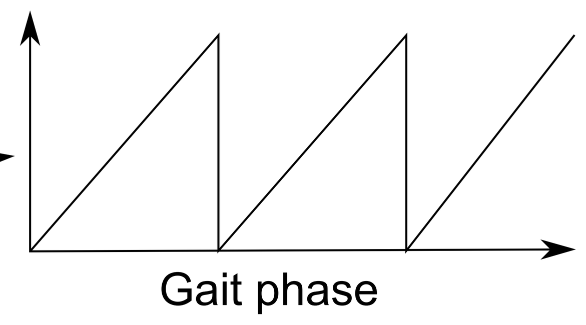
$\Rightarrow$ smaller actuator requiring smaller power peak.

The constructive parameters (lever arm, stiffness, etc.) were optimized using a genetic algorithm, with the objective to minimize the motor peak power, i.e. to maximize the energy transfer between the foot and the spring.

Continuous adaptation

The controller phase is the output of an AO. This is a set of differential equations having the capacity to synchronize to an external signal $\theta(t)$ by learning its features (frequency, etc.) in state variables.

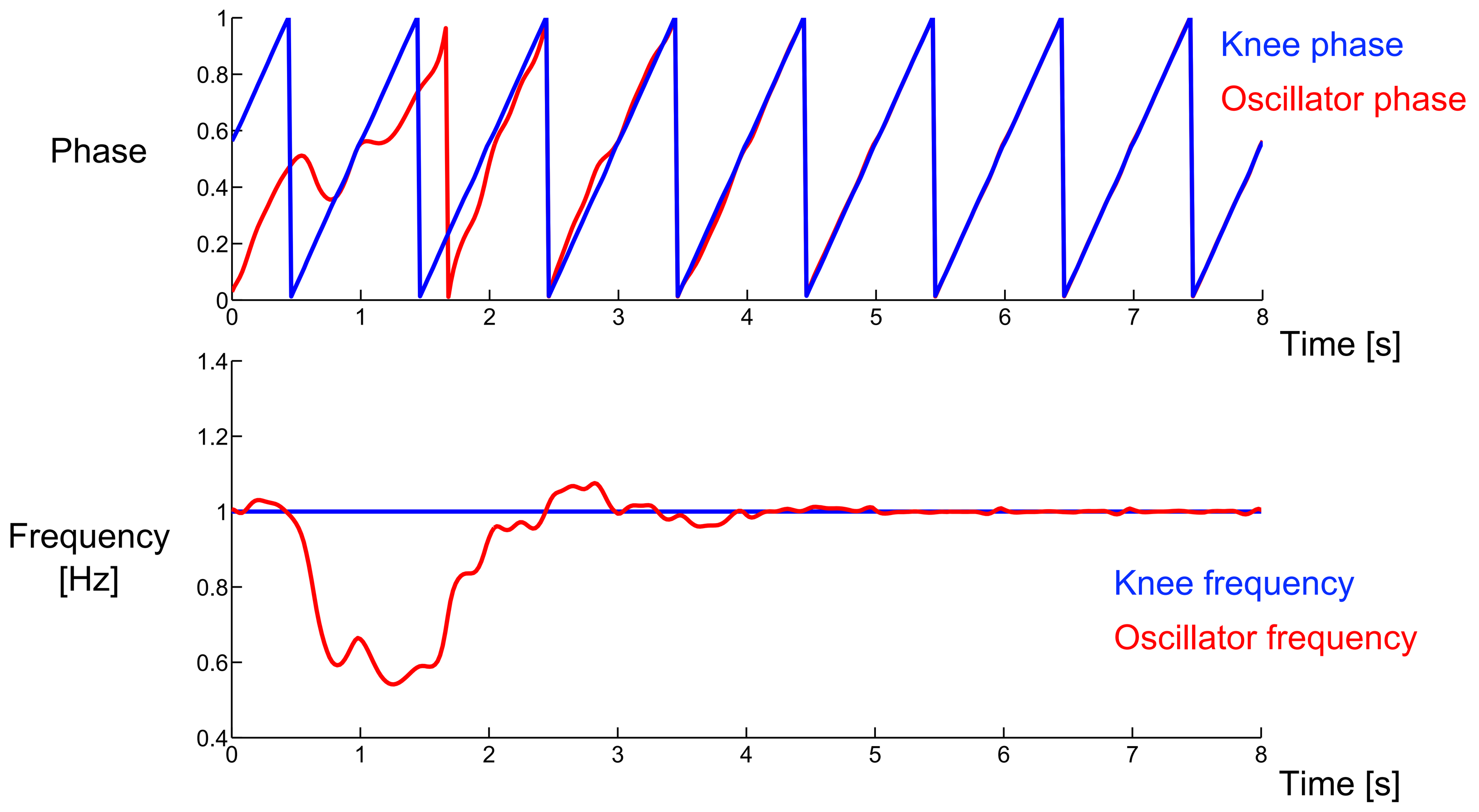




$$\begin{aligned}
\dot{\phi}_i &= i\omega + \epsilon F(t)\cos(\phi_i)\\
\dot{\omega} &= \epsilon F(t)\cos(\phi_1)\\
\dot{r}_i &= \nu F(t)\sin(\phi_i)\\
\dot{\Theta} &= \nu F(t)
\end{aligned}$$

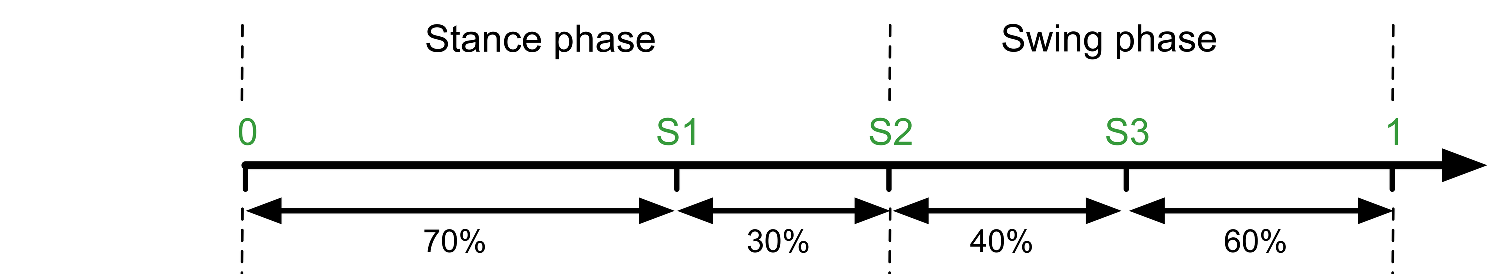
$$\hat{\theta} = \Theta + \sum_{i=1}^{n} r_i \sin(\phi_i)$$

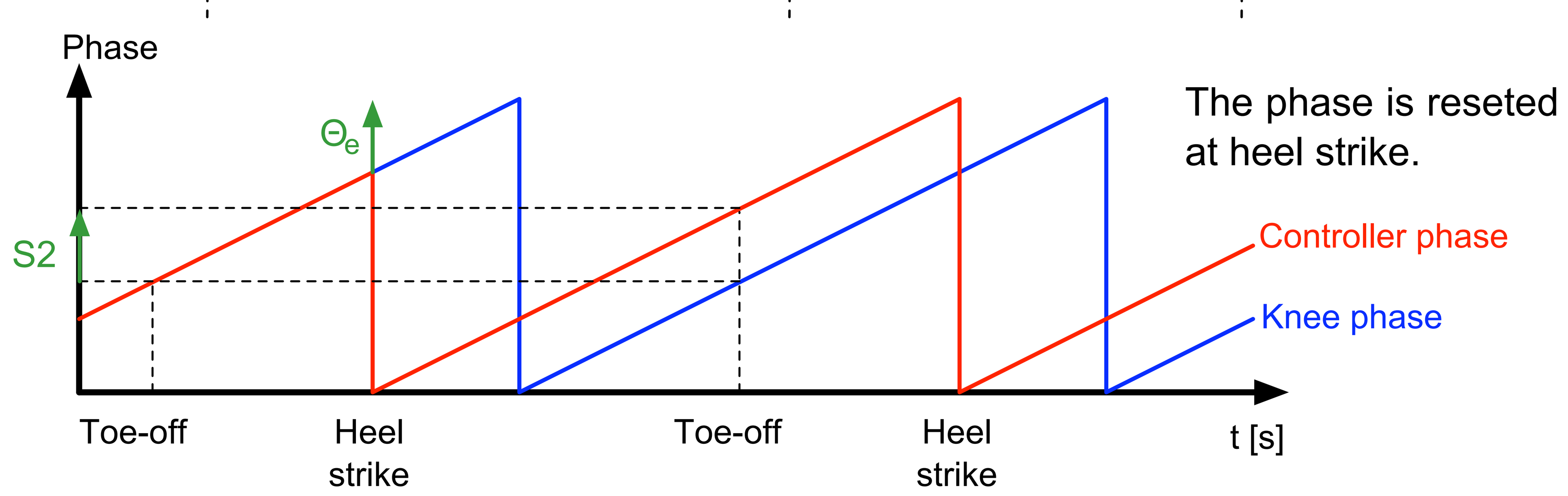
In this application, the external signal is the knee position. Once this knee signal is learned by the AO, the output phase $\phi_1(t)$ is synchronized to the actual gait phase. Adaptation to changes in the user gait features (e.g. cadence) is guaranteed by the AO convergence properties.



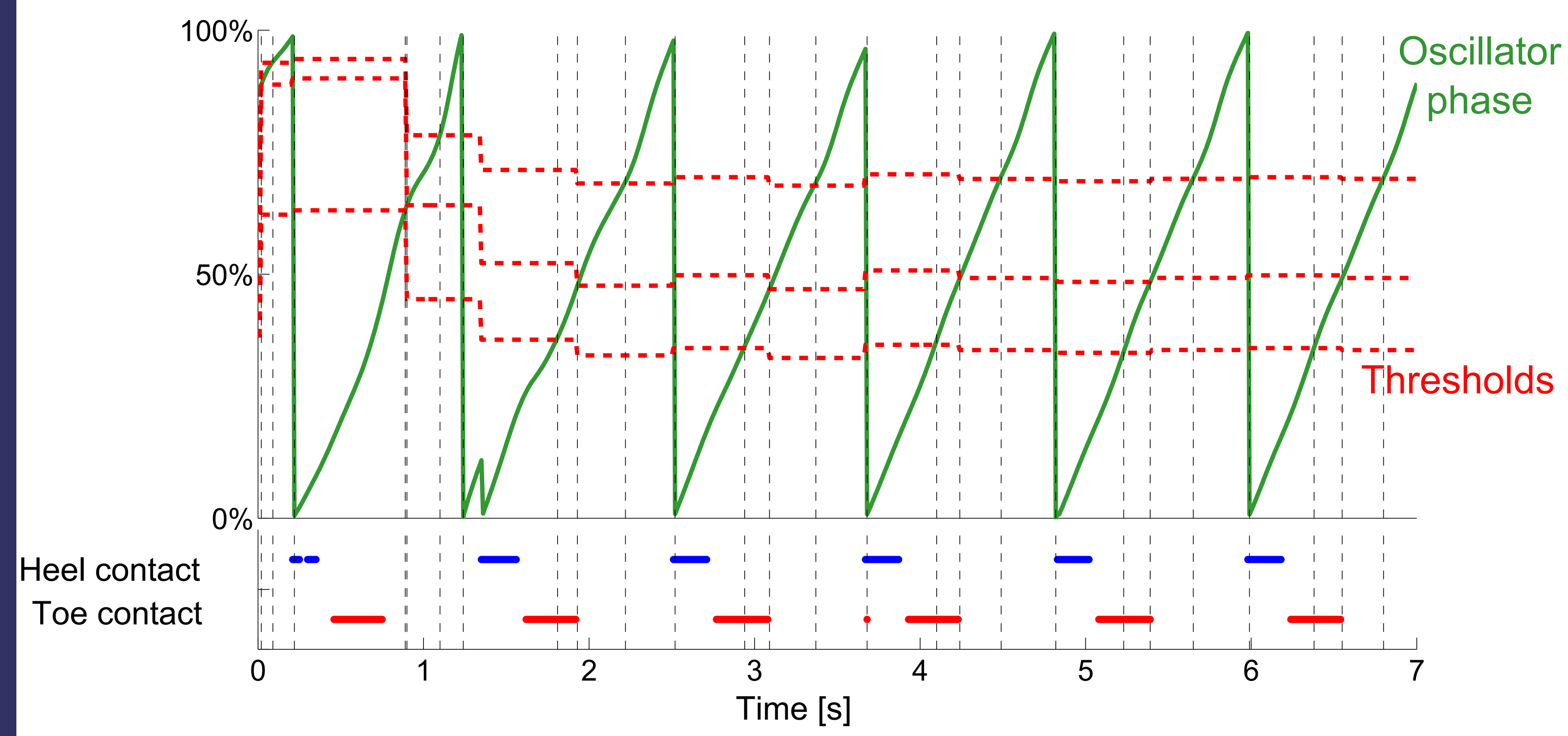
Discrete adaptation

Thresholds governing the transition between the controller states are further adapted to the user gait profile.





This is done by using foot sensors detecting heel strike and toe-off, and by adapting the phase thresholds accordingly.



Future work

The first prototype is assembled. The next steps will be :

- to embed our controller and put it to test with patients.
- to adapt the controller to the patient gait.
- to improve the prototype.