# Design and Control of an Adaptive Peroneal Stimulator with Inertial Sensor-based Gait Phase Detection

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***Abstract*— This contribution is concerned with the design and control of a novel drop foot stimulator. Unlike almost all other drop foot stimulators, the present device uses a combination of gyroscopes and accelerometers attached to the foot, and optionally to the shank. On the one hand, the inertial sensor on the foot is used for a detailed gait phase detection which allows to precisely synchronize the stimulation with the gait events. On the other hand, the accelerometers and gyroscopes are used to directly measure the success of the stimulation, i.e. the angle of the foot with respect to the ground, and optionally with respect to the shank. Based on this information, the device adapts the stimulation intensity profile from step to step, thus achieving two objectives: The first is that the system automatically compensates changes in the stimulation dynamics caused, e.g., by muscular fatigue or varying spasticity. The second is that a very natural foot motion is achieved by feeding physiological angle profiles, or alternatively the angle profile of the contralateral side, as a reference to the iterative learning controller. The effectiveness of this approach is demonstrated by experimental results with both healthy subjects and stroke patients walking on a treadmill.**

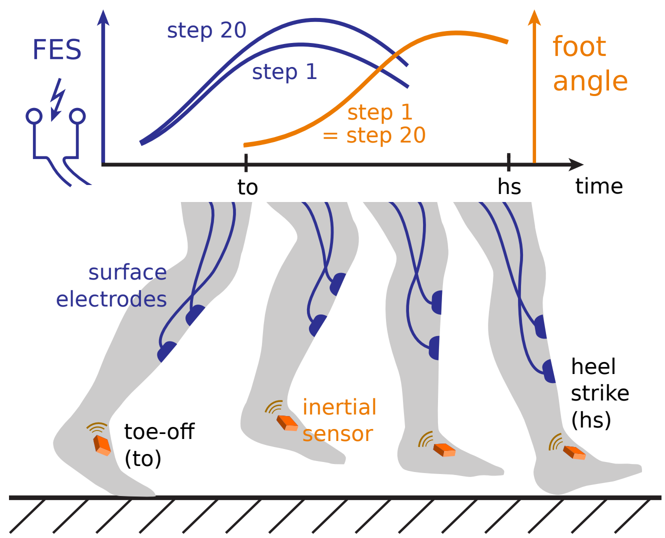


Fig. 1. Hardware setup of the novel drop foot stimulator. An inertial sensor is used for gait phase detection and angle measurement. A constantly natural foot motion is achieved through adaption of the stimulation intensity profile.

## I. INTRODUCTION

The drop foot syndrome is characterized by the limited ability or inability to dorsiflex the foot by voluntary muscle activation. Drop foot stimulators are neuroprostheses that support the foot lifting during the swing phase of gait through functional electrical stimulation (FES). They are used, e.g., in rehabilitation of sub-acute stroke patients and in the treatment of chronic drop foot patients. Most devices use heel switches to determine when the foot is on the ground [4] and apply a predefined stimulation intensity profile whenever the heel is lifted. Due to muscular fatigue, the height (or a similar measure) of that profile must be adapted periodically by the user. The alternative of choosing a higher value from the very start leads to increased fatigue. This problem can be avoided if the actual outcome of the stimulation is detected by suitable sensors. In that case, closed-loop control can be applied to adjust the stimulation intensity, which was demonstrated to yield improved performance in [5], [7], and [1]. However, while the control methods used therein yield sufficient foot clearance, they do not guarantee a natural foot motion. That this additional objective might be achieved by the use of Iterative Learning Control (ILC) was demonstrated in [6] through simplified experiments with a healthy subject.

All mentioned closed-loop control approaches require a measurement of the foot-to-ground angle or of the dorsiflexion angle of the ankle joint, i.e. the angle between the foot and shank. The former can be determined using an inertial measurement unit[1] on the foot [7]. In contrast, determining the joint angle requires a second IMU on the shank, or alternatively a goniometer [5] or bioimpedance measurements [6]. In case inertial sensors are used, the heel switch can be replaced by a gait phase detection that uses the measured accelerations and angular rates instead. In [3] and [2], it has been demonstrated that an inertial sensor attached to the shank yields as accurate results as a heel switch. But without a second sensor on the foot, the amount of foot lifting cannot be determined. In [7], it is demonstrated, that a single inertial sensor on the foot yields a more detailed gait phase detection and enables the determination of the foot-to-ground angle. However, the sensor must be mounted in a predefined orientation and the algorithms are limited to walking on level ground.

In the present contribution, we overcome these restrictions by introducing the design of a novel drop foot stimulator that uses two inertial sensors mounted to the foot and shank in arbitrary orientation and position. In Sections II-A and II-B,

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[1]i.e. typically a combination of accelerometers, gyroscopes, and magnetometers

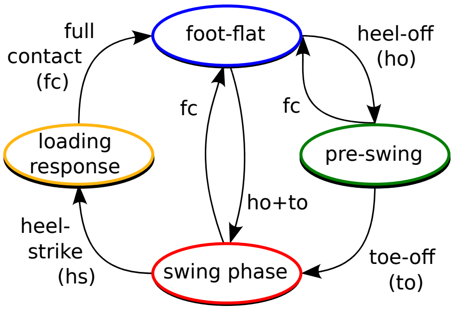


Fig. 2. Finite state automaton for detailed gait phase detection. Four typical gait events (representing the transitions of the automaton) are detected from characteristics in the measured acceleration and angular rate.

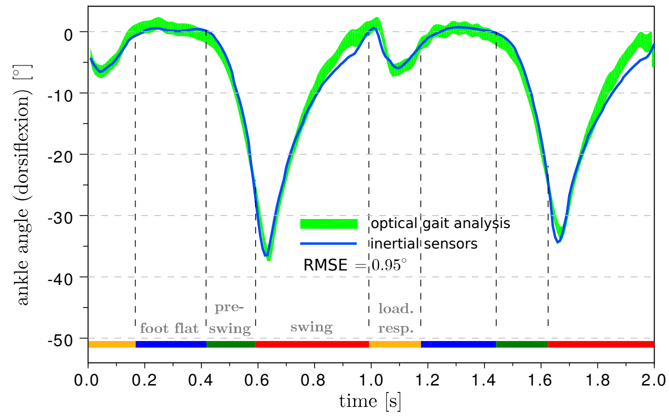


Fig. 3. Comparison of inertial sensor-based joint angle measurements with the results of an optical gait analysis system. The root-mean-square deviation is about 1°. Vertical dashed lines indicate the gait phase transitions that are detected by analyzing the inertial measurement signals of the foot sensor.

we explain how these are used for gait phase detection and angle measurement, respectively. In the subsequent Section II-C, we use the aforementioned method of ILC to adjust the stimulation intensity profile periodically. In Section III experimental results with healthy subjects and, for the first time, with stroke patients are presented.

## II. MATERIAL AND METHODS

### A. Gait Phase Detection

For gait phase detection, a single inertial measurement unit, consisting of a three-dimensional accelerometer and a three-dimensional gyroscope, is used. While the IMU may incorporate magnetometers as well, we refuse to use their measurement information, since it is known to be less reliable for indoor use and in the presence of magnetic disturbances. The IMU is attached to the foot or shoe using adhesive tape, elastic straps, or by putting it between the shoe tongue and shoelace. The orientation and position of the sensor with respect to the foot or shoe is assumed to be unknown, thus allowing for maximum freedom of mounting and yielding more robustness.

The gait is modeled by a finite state automaton, as depicted in Figure 2, and the transitions are automatically detected by recognizing certain characteristics in the measured accelerations and angular rates. More precisely, the beginning and end of the foot flat phase is detected by the norms $||a(t)||_2$ and $||g(t)||_2$ entering and leaving the proximity of $9.8\,\frac{\mathrm{m}}{\mathrm{s}^2}$ and $0\,\frac{\mathrm{rad}}{\mathrm{s}}$, respectively. At the end of every foot-flat phase, the vertical axis is determined by averaging the measured accelerations, and a strap-down integration is (re-)started. Subsequently, the foot's pitch axis is identified from the pre-swing motion and the toe-off is then detected by a major increase in the norm of the horizontal velocity and by a proper[2] sign change in the pitch rate. Finally, the heel strike is detected by a major decrease in the norm of the horizontal velocity and a spike in the time-derivative of the acceleration. All further details shall be omitted at this point for the sake of brevity. Figure 3 shows exemplary results of the gait phase detection, which agree well with corresponding features in the joint angle.

[2] i.e. the signal must exceed a certain limit first

### B. Angle Measurement

As pointed out in the introduction, the closed-loop control of a drop foot stimulator can either employ foot-to-ground angle measurements or measurements of the dorsiflexion angle of the ankle joint. While the former can be obtained by using a single IMU attached to the foot, the latter requires a second IMU attached to the shank. In the following, we briefly discuss both options under the assumption that, again, the IMUs are attached in arbitrary unknown position and orientation to the foot and shank.

The foot-to-ground angle is calculated from the angular rates measured on the foot. As explained in Section II-A, a strap-down integration is (re-)started at the end of each foot-flat phase. Therefore, the orientation of the foot with respect to the ground, and thus the respective two-dimensional angle, is known up to the small amount of drift that accumulates between two foot-flat phases. As soon as full contact is detected, this drift can be removed by using the acceleration averaged over the first few samples of the new foot-flat phase[3]. Although this means that the measured angle profile can only be transmitted batch-wise for each step, the method works well for a drop foot stimulator that uses ILC to update the stimulation intensity profile in the foot-flat phase between two steps[4]. Finally, please note that, unlike previous approaches, this method is robust with respect to changes in the slope of the ground and always yields the two-dimensional angle between the foot and the ground from which the foot took off.

A second IMU is attached to the shank with a special mounting strap in order to determine the dorsiflexion angle

[3] For drift elimination, we assume at least stepwise constant bias

[4] For applications which require driftless angle estimates in real time, it is possible to perform immediate (approximate) drift removal using the bias that is estimated from the measurement data of the previous step.

of the ankle joint. As a first step, the methods explained in [9] are used to calculate the dorsiflexion axis coordinates $j_1, j_2$ (in the first and second sensor's local coordinates) from the gyroscopic measurement data of a few steps or a few seconds of free dorsiflexion and plantarflexion movements. Subsequently, the dorsiflexion angle is computed for each moment in time by sensor fusion of the gyroscope-based angle $\int j_1^T g_1(t) - j_2^T g_2(t) dt$ and an accelerometer-based angle estimate, see the methods in [10] for details. Unlike the previously explained foot-to-ground angle, this calculation of the dorsiflexion angle does not require periodic phases of rest. The experimental results in Figure 3 demonstrate that this method has an accuracy of about 1° when compared to optical systems.

### C. Iterative Learning Control

The gait phase detection described in Section II-A is used to apply a stimulation intensity profile in every step via surface electrodes that were carefully placed over the muscles associated with dorsiflexion. Here, a stimulation profile refers to the values of stimulation intensity[5] for each sample instant from heel-rise to full contact. In the first step, this profile is chosen, based on heuristic values, to guarantee sufficient foot clearing even for weak muscular responses. In general, this means that the intensity is higher than required. Therefore, the complete profile of the foot-to-ground angle and the dorsiflexion joint angle is determined in each step using the methods from Section II-B. In each foot-flat phase, the angle profile is compared to a reference profile that is either predetermined from standard gait data or determined from the very last step of the contralateral side using the same instrumentation and methods as on the paretic side.

Whether it is more beneficial to use the foot-to-ground angle or the dorsiflexion joint angle and whether to use generic reference profiles or an online reference from the contralateral side, is an issue of current research and is not further discussed at this point. However, in all four possible combinations, the observed difference between the actual angle profile and the reference profile is used to update the stimulation intensity profile before the next heel-rise. If this update is conducted properly, then the deviation between both profiles will be reduced to a small value and the desired foot motion will be achieved within a few steps. Appropriate control design methods for this application have been developed in [8]. Furthermore, please note that, even after convergence, this update is consecutively performed in each foot-flat phase. Thereby, the stimulation profile is always adapted to yield the desired motion, even when muscular fatigue or similar slow variances in the system dynamics occur.

## III. EXPERIMENTAL RESULTS

The closed-loop control from Section II-C is evaluated experimentally with sub-acute stroke patients walking on

[5] i.e. pulsewidth or current, or a combination of both

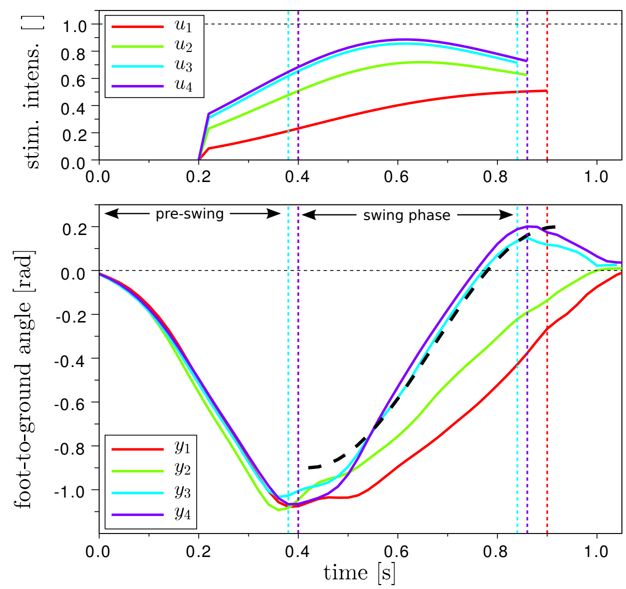


Fig. 4. Iterative Learning Control of the foot lift during swing phase of four consecutive steps. The foot-to-ground angle profile $y$ between toe-off (left vertical dashed lines) and heel-strike (right vertical dashed lines) converges to a predefined reference profile (bold dashed) within only two steps.

a treadmill at the hospital Medical Park Humboldtmühle, Berlin. Prior to this, all functions are tested in experiments with healthy subjects who simulated a drop foot by walking without foot dorsiflexion in steppage gait on a treadmill. All experiments are conducted using FES surface electrodes that are carefully placed on the lateral side of the lower leg to provide proper dorsiflexion with little eversion and inversion. The foot-to-ground angle and respective reference profiles were fed to the ILC, while the timing of the stimulation was driven by the gait phase detection.

Figure 4 demonstrates how the measured angle profile approaches the reference from step to step, as the stimulation intensity profile is adapted by the ILC algorithm. While the foot touches the ground with toes first in step one, a proper heel strike is observed in steps three and four. In this experiment, the initial stimulation profile was, on purpose, chosen insufficiently small in order to demonstrate the rapid learning of the ILC. Furthermore, please note that variances in the length of the swing phases do not affect this convergence, since they were taken into account during controller design [8].

In a series of experiments, which were approved by the Ethics Committee at Charité Berlin, we further evaluated the long-time behavior of the closed-loop drop foot stimulator with stroke patients. Figure 5 gives an example for adaption to slow changes in the stimulation dynamics. An increasing stimulation intensity is required to achieve a constantly small deviation between measured and reference angle profile. For each step, the given quantities are determined by averaging over all sample values of the applied or obtained profiles. Therefore, the stimulation intensity appears to be smaller in

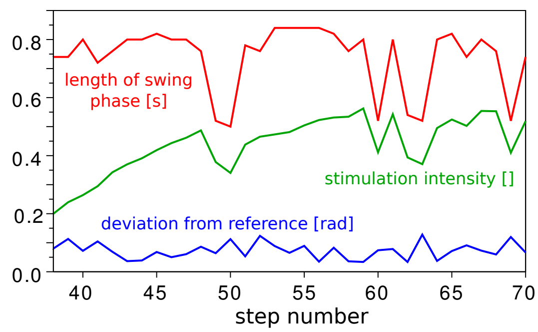


Fig. 5. Course of characteristic quantities over thirty steps of a hemiplegic patient walking with the novel drop foot stimulator at constant speed on a treadmill. The stimulation intensity is slowly increased by the ILC to compensate for fatigue and to keep the deviation between measured foot-to-ground angle and reference small.

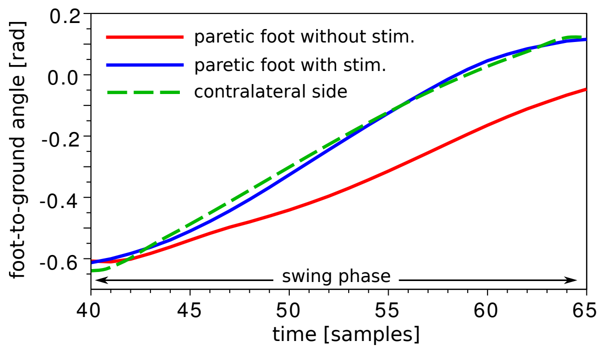


Fig. 6. Comparison of the foot-to-ground angle profiles from three steps of a hemiplegic patient walking with the novel drop foot stimulator at constant speed on a treadmill. Without stimulation, the patient achieves (just about) sufficient foot clearance. The additional support of the FES (adjusted by ILC) yields a better angle profile and a symmetric gait.

steps with a shorter swing phase. However, as before, the ILC achieves small tracking errors despite these variations.

Finally, the option of using the contralateral foot-to-ground angle as a reference to the ILC was evaluated. A hemiplegic patient that was able to walk a few steps without FES support was equipped with one IMU on each foot. Results are presented in Figure 6. The ILC has learned to support the patient in such a way that the angle profiles, and thus the foot motions, on both sides are almost equal. This yields two advantages: One is safety, because the foot clearance is larger than in the step without the stimulation. The second is a symmetric gait, which is desirable from both a medical and an aesthetic point of view.

## IV. DISCUSSION AND CONCLUSIONS

The experimental results of Section III prove the effectiveness of the present approach. Inertial sensors combined with suitable algorithms were demonstrated to provide highly accurate gait phase detection and useful angle information. Measuring the foot-to-ground angle and using this information to adapt the stimulation profile via ILC yields a constantly physiological and symmetric gait. Muscular fatigue and similar variances in the stimulation dynamics are compensated automatically.

However, certain aspects of the system can be improved: For some patients, the electrode position that yields good foot lifting in seated pose was found to yield undesired eversion or inversion while walking. Therefore and in order to reduce the effort of electrode placement in general, array electrodes should be used and algorithms should be developed for both initial estimation and online adaption of proper stimulation sites. As a further method of improvement, the measured dorsiflexion joint angle should be incorporated in the control scheme, since it yields valuable additional information on the motion of the lower leg that strongly influences the foot motion. Finally, all experimental results were obtained at constant walking speed. Consequently, speed changes will be subject of further research, as well as climbing stairs and slopes.

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## REFERENCES

[1] M. Hayashibe, Q. Zhang, C. Azevedo-Coste, "Dual predictive control of electrically stimulated muscle using biofeedback for drop foot correction", *IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, pp. 1731-1736, September 2011.

[2] R. Héliot, K. Mombaur, C. Azevedo-Coste, "Online CPG-Based Gait Monitoring and Optimal Control of the Ankle Joint for Assisted Walking in Hemiplegic Subjects", *Modeling, Simulation and Optimization of Bipedal Walking (Cognitive Systems Monographs)*, vol. 18, pp. 53-69, 2013.

[3] D. Kotiadis, H.J. Hermens, P.H. Veltink, "Inertial Gait Phase Detection for control of a drop foot stimulator inertial sensing for gait phase detection", *Medical Engineering & Physics*, vol. 32, no. 4, pp. 287-297, 2010.

[4] G. M. Lyons, T. Sinkjær, J. H. Burridge, D. J. Wilcox, "A Review of Portable FES-Based Neural Orthoses for the Correction of Drop Foot", *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 10, no. 4, pp. 260-279, 2002.

[5] N. Mourselas, M.H. Granat, "Correction of Drop Foot Using a Fuzzy Logic Controlled Miniature Stimulator", *Proc. of the 5th Annual Conference of the International Functional Electrical Stimulation Society*, pp. 140-141, July 2000.

[6] H. Nahrstaedt, T. Schauer, R. Shalaby, S. Hesse, J. Raisch, "Automatic Control of a Drop-Foot Stimulator Based on Angle Measurement Using Bioimpedance", *Artificial Organs*, vol. 32, no. 8, pp. 649-654, 2008.

[7] N.-O. Negård, "Controlled FES-assisted gait training for hemiplegic stroke patients based on inertial sensors", *Doctoral Thesis*, TU Berlin, 2009.

[8] T. Seel, T. Schauer, J. Raisch, "Iterative Learning Control for Variable Pass Length Systems" *Proceedings of the 18th IFAC World Congress*, pp. 4880-4885, 2011.

[9] T. Seel, T. Schauer, J. Raisch, "Joint axis and position estimation from inertial measurement data by exploiting kinematic constraints", *Proc. of the IEEE International Conference on Control Applications (CCA)*, Pages 45-49, October 2012.

[10] T. Seel, T. Schauer, "IMU-based Joint Angle Measurement Made Practical", *Proc. of the 4th European Conference on Technically Assisted Rehabilitation*, March 2013.