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Iterative Learning Control and System Identification of the Antagonistic Knee Muscle Complex During Gait Using Functional Electrical Stimulation

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Abstract: Functional Electrical Stimulation (FES) can be used to support the gait of stroke patients. By measuring joint angles and adjusting the stimulation intensities automatically to the current need of the patient, setup times can be reduced and time-variant effects like muscle fatigue can be compensated. This was achieved in recent publications by using Iterative Learning Control (ILC) on the ankle complex. In this paper we consider FES of the antagonistic knee muscle complex (quadriceps and hamstring muscles) that controls knee flexion/extension. We used a coactivation strategy in order to map the two stimulation channels to a single control input. A large class of dynamic models was obtained by system identification based on data from two experiments: one with standing subjects and one with subjects walking on a treadmill while being stimulated during different time segments of the gait cycle. Time delays, system poles, and in particular the system gains were found to vary largely. Furthermore, large differences were observed between muscle dynamics in standing pose and during walking. We designed an iterative learning controller that is stable for almost all models. In experiments with eight healthy subjects walking on a treadmill, the ILC was found to reduce deviations from a reference trajectory to about five degrees within two strides.

Keywords: iterative learning control (ILC), functional electrical stimulation (FES), stroke rehabilitation, gait, neuroprosthesis, system identification, multichannel, adaptive, knee angle

1. INTRODUCTION

One of the symptoms of stroke is an impairment of gait originating from a partial paralysis of one side of the body. In milder cases, the patients can be actively supported in their movements by using gait-triggered Functional Electrical Stimulation (FES). The first FES-based neuroprosthesis by Liberson et al. (1961) used a foot switch to trigger a stimulation of the tibialis anterior muscle during the swing phase, successfully supporting foot drop patients. Many more foot drop stimulators have been developed since. A review can be found in Lyons et al. (2002). In the 1970s, the single channel stimulation was extended to multichannel stimulation of different muscle groups of the entire gait muscle complex, e.g. gastrocnemius, hamstrings, quadriceps, gluteus maximus, gluteus medius and even shoulder muscles. Each muscle group was then triggered with an individual timing, duration and stimulation intensity. Bogataj et al. (1997) could show that multichannel stimulation had a better effect on rehabilitation than single channel stimulation.

While many studies could show the positive effects of the FES neuroprosthesis, a lot of practical problems still remain. With fixed, triggered stimulation patterns, the clinician or user has to choose the timing, duration and the stimulation intensity of every stimulation channel. From stroke patient to stroke patient there are vast differences in gait due to compensation movements and different severity of paralysis. Hence, highly individualized parameters for the stimulation of each muscle group are needed. Finding a satisfying parametrization is a nontrivial and time-consuming task for the clinician, especially in the often short rehabilitation training sessions. The optimal parameters can also vary within the same individual due to variation of electrode placement, muscle fatigue and bodily changes.

A way to solve this problem is to measure an important parameter of the gait, e.g. the joint angle, and use an automatic algorithm to adapt the stimulation patterns according to the measurement. Since the interaction between stimulation, gait and the human being controlling the gait is a highly nonlinear and time-varying process, robust methods are crucial. One very natural and robust approach is a cyclic adaptation of the stimulation parameters. This means learning from the previous steps to tune the parameters of the current step. Franken et al. (1995) used a cycle-to-cycle control strategy to tune the stimulation duration of the hip flexor muscle at every step by measuring the hip angle range. A more powerful approach is the use of Iterative Learning Control (ILC),

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which is able to not only tune a single parameter but learn an entire input trajectory. ILC was first used together with FES by Dou et al. (1999) to control the elbow angle. Nahrstaedt et al. (2008) were the first to apply ILC during gait on the tibialis anterior muscle. Hughes et al. (2009), Freeman et al. (2009) and Meadmore et al. (2012) further investigated into ILC strategies for the upper limbs. Seel et al. (2016) used ILC to control the tibialis anterior and fibularis longus muscle achieving physiological dorsiflexion and eversion of the foot in walking stroke patients without the need of manual parameter tuning.

So far, ILC was only used in connection with a one or two channel foot drop neuroprosthesis. The tuning and the stability analysis was done by either identifying the dynamics of a sitting subject or by using heuristic tuning methods. During gait the system dynamics are expected to differ from sitting or standing due to the voluntary muscle contractions, the reaction of the subject's movements to the FES and the general complexity of the gait.

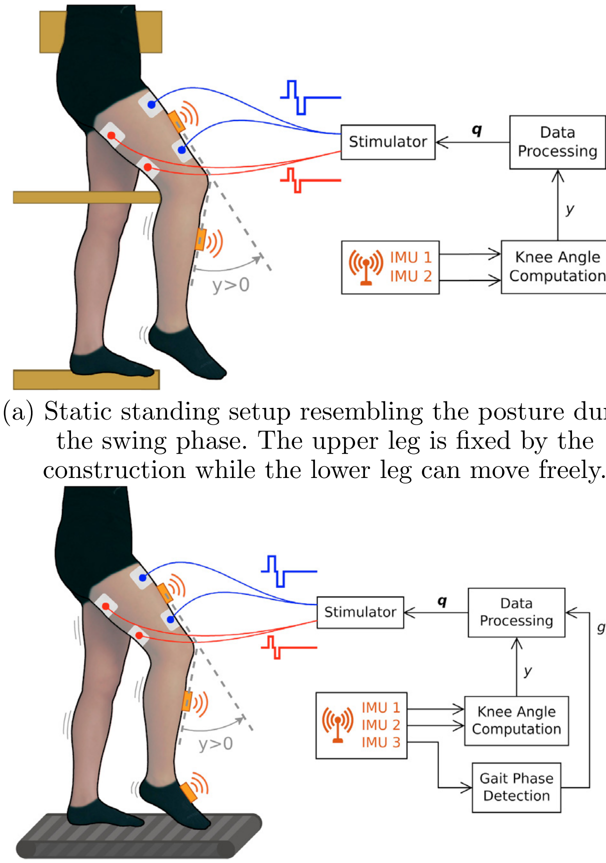
In this paper we want to move closer towards an ILC-based multichannel neuroprosthesis. In order to achieve this, we designed an ILC for the antagonistic knee muscle complex. This ILC could be later used together with an ILC of the ankle complex. One of our main goals was to investigate into the dynamics of stimulation and knee angle during gait. We used a coactivation strategy in order to map the two stimulation channels to a single control input. 5 healthy subjects were asked to walk on a treadmill while being stimulated at different times of gait. From this data we identified simple dynamic models and compared them to models that we identified on standing subjects. In a second experiment 8 healthy subjects were asked to walk on a treadmill while the ILC was tested.

2. METHODS

2.1 Experimental Setup

In order to measure the joint angles and detect gait phase events, we used three wireless Inertial Measurement Units (IMUs) sampling at 100 Hz (MTw wireless units, Xsens Technologies B.V., Netherlands). An eight-channel stimulator was used for the electrical stimulation with a frequency of 50 Hz (Rehastim, Hasomed GmbH, Germany). The placement of the electrodes and the IMUs is depicted in Fig. 1. The stimulation intensity of each channel was controlled by a parameter q proportinal to the stimulation charge. A $q = 0$ would mean a pulse width of 0 and a current of 0, both were linearly increased so that a $q = 1$ corresponds to a pulse width of 500 μs and a current of 50 mA.

For the IMUs on the upper and lower leg, an orientation estimation algorithm was used to estimate the absolute orientation from the gyroscope, accelerometer and magnetometer data. The real-time knee angle was then calculated using Euler decomposition and downsampled to 50 Hz. The IMU mounted to the foot was used to detect real-time gait events, using a threshold-based approach (Müller et al., 2015). Four distinct events could be detected: initial contact, full contact, heel-off and toe-off (also downsampled to 50 Hz).



(a) Static standing setup resembling the posture during the swing phase. The upper leg is fixed by the construction while the lower leg can move freely.

(b) Walking setup. The subject walks on a treadmill at a fixed speed.

Fig. 1. Experimental setup for the standing and walking experiments

For all experiments, two different setups were used. The standing pose resembling the swing phase is shown in Fig. 1a, here the upper leg was fixed by a construction. In the second and main setup, the subject were asked to walk on a treadmill at a constant speed of 1.5 km/h (Fig. 1b). The three following experiments were conducted in the scope of the paper:

- System identification while standing (5 healthy subjects)
- System identification while walking (same subject group)
- ILC while walking, preceded by a brief system identification while standing (8 healthy subjects)

2.2 Coactivation Strategy

Most simple control problems are Single-Input Single-Output (SISO) systems. In our case there are two control inputs, the stimulation intensity of the quadriceps and of the hamstring muscles. However, there is only one system output, the knee flexion angle. A straightforward way to use a standard ILC controller is by mapping the two stimulation inputs to one virtual control input, creating a SISO problem.

The basic idea of mapping both stimulation inputs to one virtual input is that a positive input leads to an increase of the knee angle, while a negative input causes a decrease. When controlling antagonistic muscle pairs, the human