Control of the elbow joint angle in the vertical plane using Functional Electrostimulation (FES)

Final Report

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ANSYMB II - Real-time control of actuated system

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1 Abstract

Functional electrostimulation offers a way of creating movement in limbs whose functional motor innervation is no longer given. In this project, the accuracy of joint angle control in the vertical plane, thus considering the effects of gravitation on the system, was tested. A crucial difficulty regarding this task, is the nonlinearity of the muscles torque gain relationship, making it complicated to model the muscles behaviour as an actuator. After successfully developing a model and estimating further parameters needed, like the test person specific torque gain value by trial, a control loop with PI control only, as well as one with PI- and an additional pilot control were tested.

2 Background

There are several diseases, that can lead to mobility restriction, with probably the best-known being (partial) spinal cord injuries - mainly due to accidents – or furthermore, stroke induced damages. In both cases the defect is to be found on level of the central nervous system. As a result, the motoric innervation, especially of the extremities, but also of inner organs such as the bladder, might be affected to varying degrees, depending on the extend of the injury in question.
In those body regions, in which the capacity for motoric actuation is no longer given, muscle mass and strength will gradually decrease. To counteract this process of muscle atrophy, and to maintain whatever mobility is left, Functional Electrostimulation (FES) is commonly used for rehabilitation purposes.

Functional electrostimulation works by triggering muscle contractions through controlled application of electrical stimuli from an external voltage source. Generally, FES can be applied in one of two ways, depending on the location of the lesion. In case of upper motor neuron lesion – as is the case in spinal cord injuries – either stimulation of the peripheral motor neuron or direct muscle stimulation via surface electrodes is possible. Whereas, in the case of lesion in the lower motor neuron, only direct muscle stimulation delivers results. [1]
However, external triggering of partly impaired – or completely paralyzed – muscles not only offers a way to reduce and counteract muscle atrophy, but makes it possible for the muscle to be used as an actuator of the locomotion system again, despite its impairment. Depending on the purpose of the application single muscles, as well as muscle groups or antagonist muscle pairs can be addressed via FES.

Gadgets like the „MyGait", developed by the medical technology company Otto Bock, already use FES technology for motion support. "MyGait" is used in patients with weak foot dorsiflexion, also known as foot drop. The system works by stimulating the Peroneal nerve, which triggers the contraction of the anterior tibial muscle and further foot extensors, and thus supports the foots dorsiflexion. [2]

Fig. 1 - MyGait

Prior research in the field of joint angle control with functional electrical stimulation (FES) [3] shows, that the elbow angle can be regulated using FES by implementing a cascaded feedback control. To simplify the control complexity, a wooden swing was used on which the lower arm could rest, allowing arm movements in the horizontal plane. In this way, the impacts of gravitational force on the arm could be neglected. For their control *Raisch et al.* used the antagonist muscle pair of biceps- and triceps brachii as actuators, since both where needed, as there was no external force bringing the arm back to the initial position. All measured signals (accelerations, velocities and positions) were used fed back to the control loop an observer was implemented.

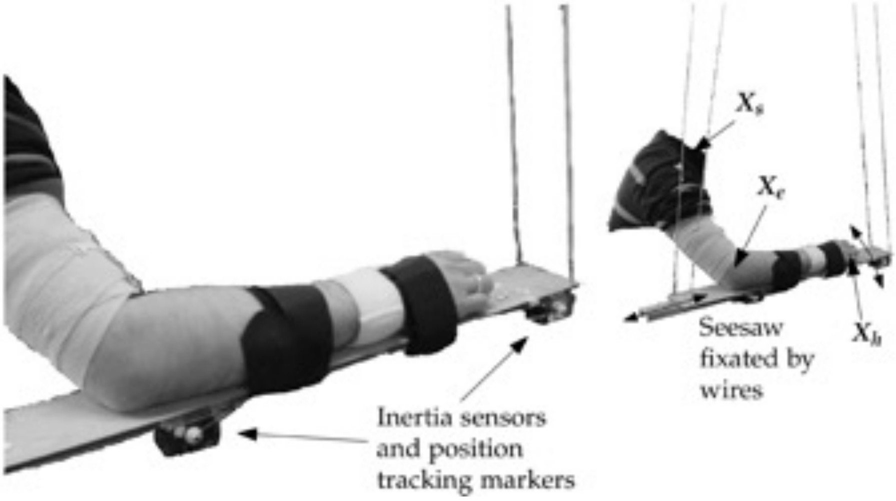

Fig. 2 - Elbow angle control in the horizontal plane

The aim of our project was to implement a control loop like that of *Raisch et al.,* only with the important distinction that ours should be able to control the elbow angle not in the horizontal, but in the vertical plane, with the arm hanging in parallel to the body in a natural position, thus considering the influence of gravitation to our system.
For this system only one muscle, the biceps brachii, needs to be triggered, as gravitation will bring the arm back to its starting position without the need for external triggering. The possibility to regulate the vertical arm movement brings the possible advantages of enabling the wearer of the system to perform tasks like i.e. lifting (grocery) bags, picking up a ball or zipping a jacket.
However, for our project, the development of an all in all stable control for the arm movement itself stands in the foreground. Like in the before mentioned research by *Raisch et al.* the RehaStim FES stimulator by HASOMED was used; with a stimulation frequency of 50 Hz and a current of 15 mA.

The main problem in designing a stable control lies in the nonlinearity of the force-length relationship of the muscle. Because of that, the torque that can be created by contracting the biceps varies, depending on the elbow joint angle. At the same time, the angle also affects the force with which the lower arm is influenced by gravitation.

Firstly, to estimate the parameters needed to establish a control – despite those nonlinearities mentioned – we will start by designing a basic Simulink model. Using those parameters and experimentally determining the biceps' torque gain characteristic a control loop with an additional pilot control will be implemented, to increase the controls stability and optimize the arm angle follow-up behaviour.

3 Methods

3.1 Experimental Setup

For implementation only one muscle, the biceps brachii is stimulated by FES. This will flex the elbow joint. To stretch the elbow joint, this method is to take the arm's and hand's mass and the effect of gravity into account. The task will be to archive desired elbow angles.
The subject (female, 24) is to stand upright, having its arm hanging loose on the side. With no visual contact to the computer screen, the subject will not be noticed of the angle aimed or the current angle measured by the system. The subject will hold an E-Stop button in its other hand.
The angle is defined between the forearm orientation and the gravitation vector. Note that the orientation of the upper arm is considered fixed. A 9 DOF IMU from Adafruit (chip LSM303) is used to measure the angle. The IMU is attached to the wrist by a Velcro bandage.

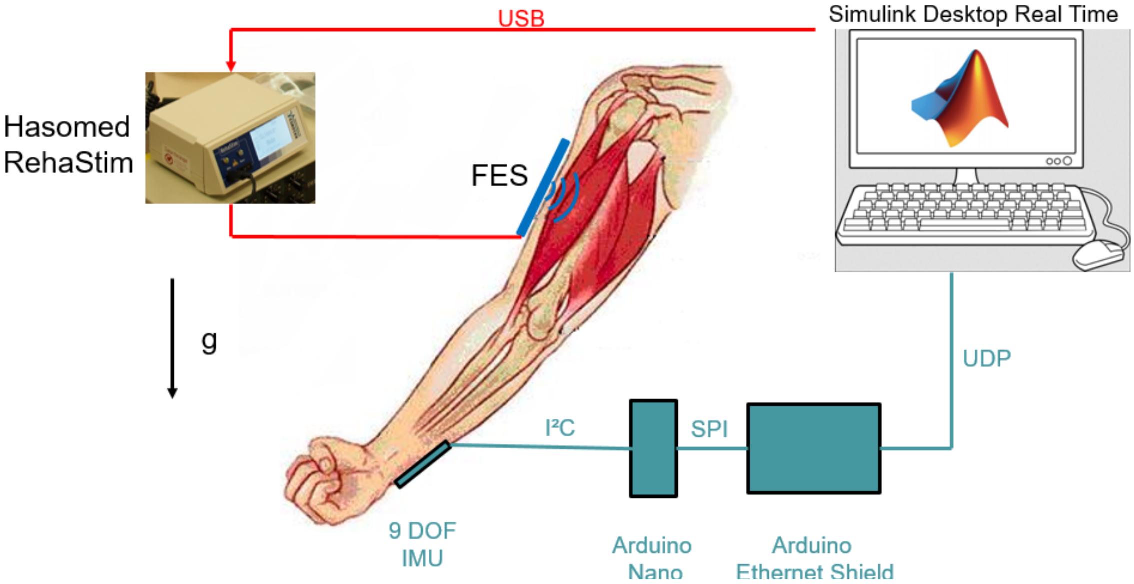

Fig. 3 - Setup

An Arduino Nano is connected by an I²C bus to read the IMU data. The Arduino runs a sensor fusion algorithms supplied by the manufacturer (Adafruit) to merge the chip's internal magnetometer and gyro accelerometer for more precision. The Arduino then sends the angle position (accuracy of 0.1 degree) to an Arduino Ethernet Shield by an SPI bus. The Ethernet Shield then sends the measurement via UDP packages and an Ethernet cable to a PC. The PC runs a Matlab Simulink Desktop Real Time Target. The pin connections are shown in Figure 8. The Stimulation device (RehaStim, HASOMED GmbH, Magdeburg) connects via USB to the PC and offers a Matlab Simulink interface when used in ScienceMode. The electric stimulation is conducted by 2 surface electrodes (8x5 cm) attached to the biceps brachii. The stimulation frequency is 50Hz and the current set to

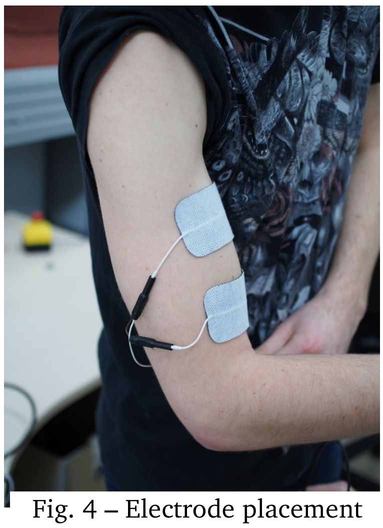

Fig. 4 – Electrode placement

constant 15mA. Stimulation intensity is controlled by the pulse width reaching form 0 to 500μs.

3.2 Derivation of control algorithm

To not cause unnecessary pain to the subject, a control strategy is found using a model based control simulation prior to the experiment. Both the arm and the muscle behaviour were modelled.

The arm is modelled by its weight and centre of mass location (in radial direction). The values for mass = 1.5kg and centre of mass location = 0.162m were taken from [4].

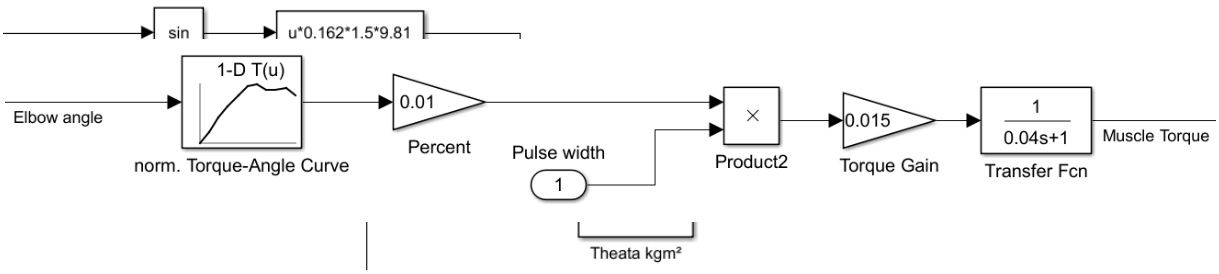

Fig. 5 – Arm Model

In [3] the muscle behaviour is described by a nonlinear recruitment curve in combination with a dynamic system. In this paper, the recruitment curve is represented by a torque-angle curve described by [5], a torque gain k and a low-pass transfer function 1st order to represent the dynamic behaviour (Figure 6).

The time coefficient of the transfer function is given according to *Raisch et al.* with 0.04s. The torque gain value of 0.015 is experimentally determined by having the subject's forearm resting at 45° against an object, ramping the stimulation pulse width until the forearm just lifts off the object and backwards calculating the required torque with help of the arm model.

The overall system model also accounts for sensor noise (white noise power 1e^{-6}), sensor time delay (0.1s rise time), stimulation quantizer (10μs – same as RehaStim tech sheet) and a stimulation saturation (500μs as pain protector).

The system was not stable with a PID controller; therefore, a pilot control was implemented. The pilot control accounts for the nonlinearity of the arm model caused by the sin function. The pilot control (Figure 7) is therefore the inverted arm model divided by the torque gain k. The torque angle curve is not being accounted for by the pilot control, as stability was already reached.

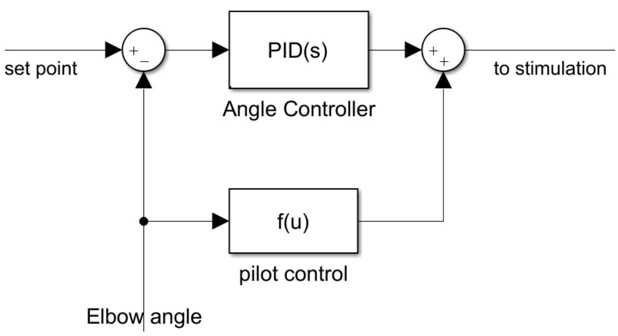

Fig. 7 - PDI- and Pilot control

3.3 Implementation to Real-Time-System

The Matlab Simulink Real Time Target runs in “normal” mode at a fixed step size of 0.01s (100Hz). Simulink requests the IMU measurements in every time step. The PID and pilot control are adopted from the simulation. IMU noise is reduced by a 10Hz low-pass filter 1st order. Real time set point values are also low pass filtered for a smoother curve (2Hz 1st order).

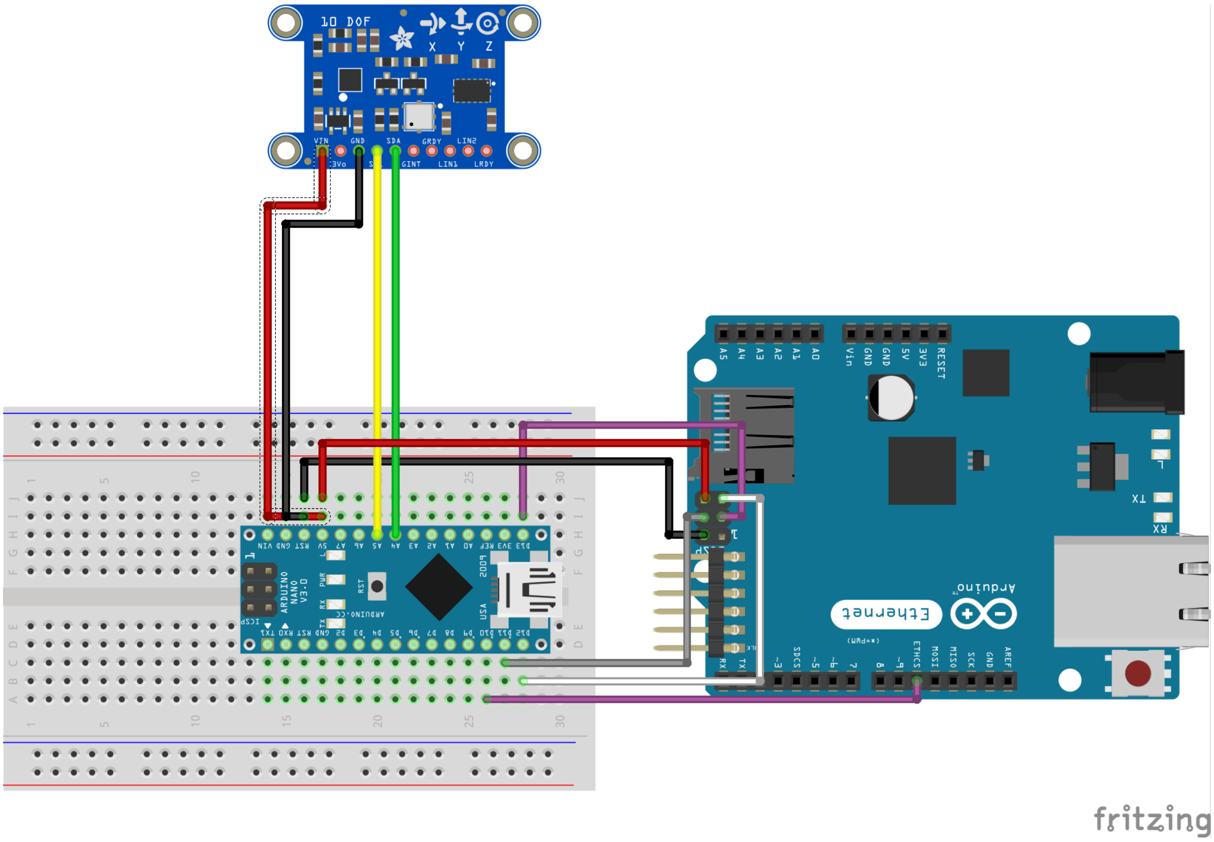

Fig. 8 - Arduino Nano, Ethernetshield and Adafruit IMU circuit

4 Results

4.1 Simulation Results

Figure 9 shows the simulation results for a PID control and a PID control combined with the pilot control. The PID + pilot control behaved stable however PID control only caused the system to alternate.

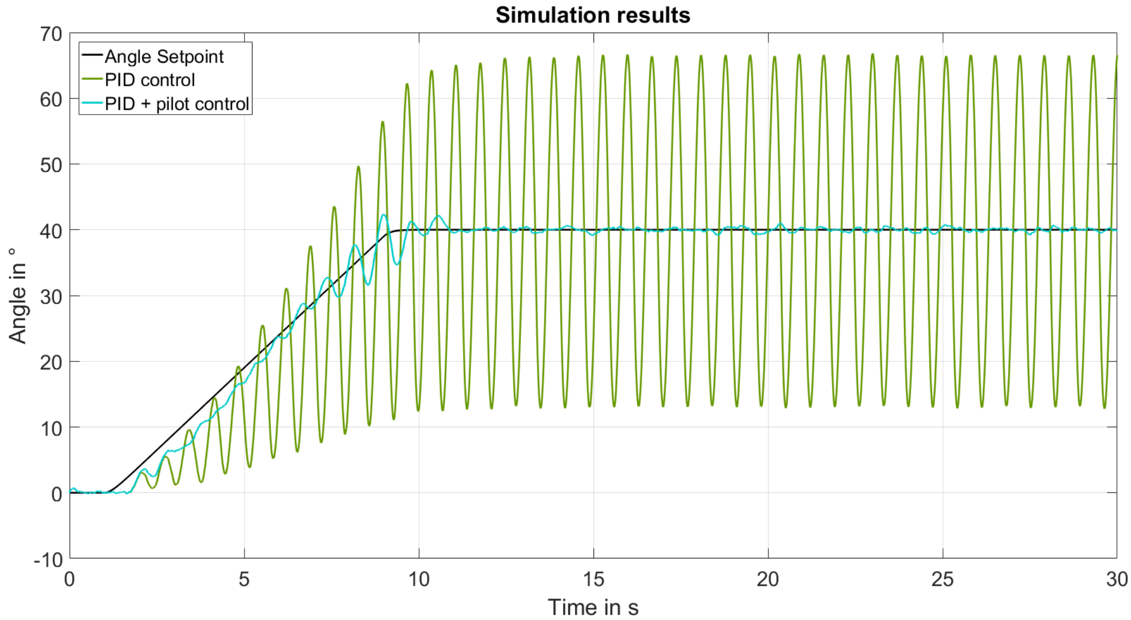

Fig. 9 - Matlab Simulink Simulation results

4.2 Experimental Results

The IMU sensor noise caused the D-Part of the controller to generate small but painful peaks. Therefore, the D-Part was removed for the experimental results, leaving a PI control and a PI + pilot control strategy. Several trials with and without pilot control were carried out. Both controllers performed rather similar. Figure 10 shows the standard deviation of the angle error for each trial. An angle below 15° was not achievable.

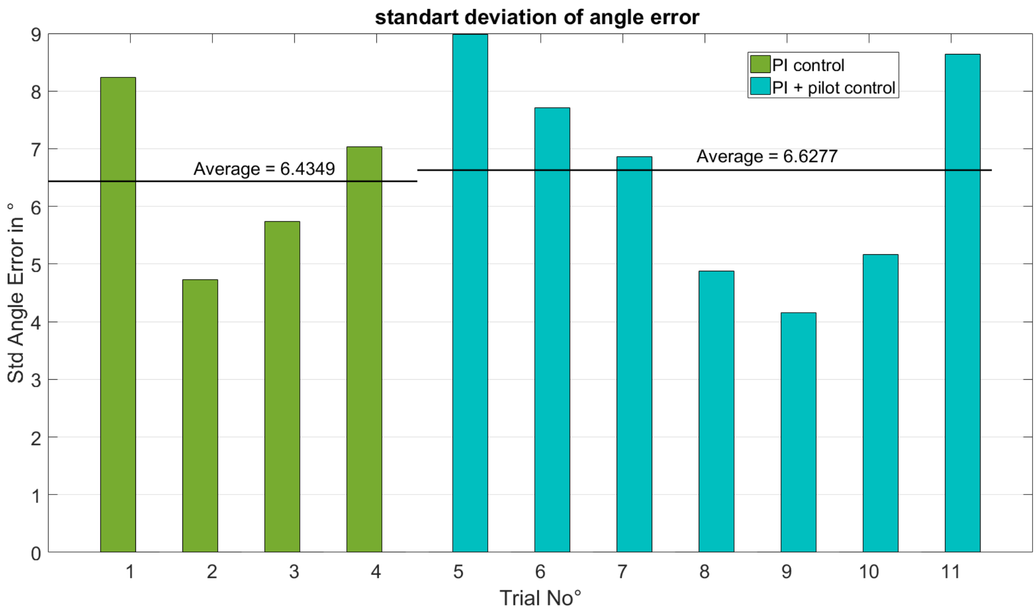

Fig. 10 - Standard angle deviation

As of the best result, Figure 11 shows the trend of set point and angle over the 45 second trial.

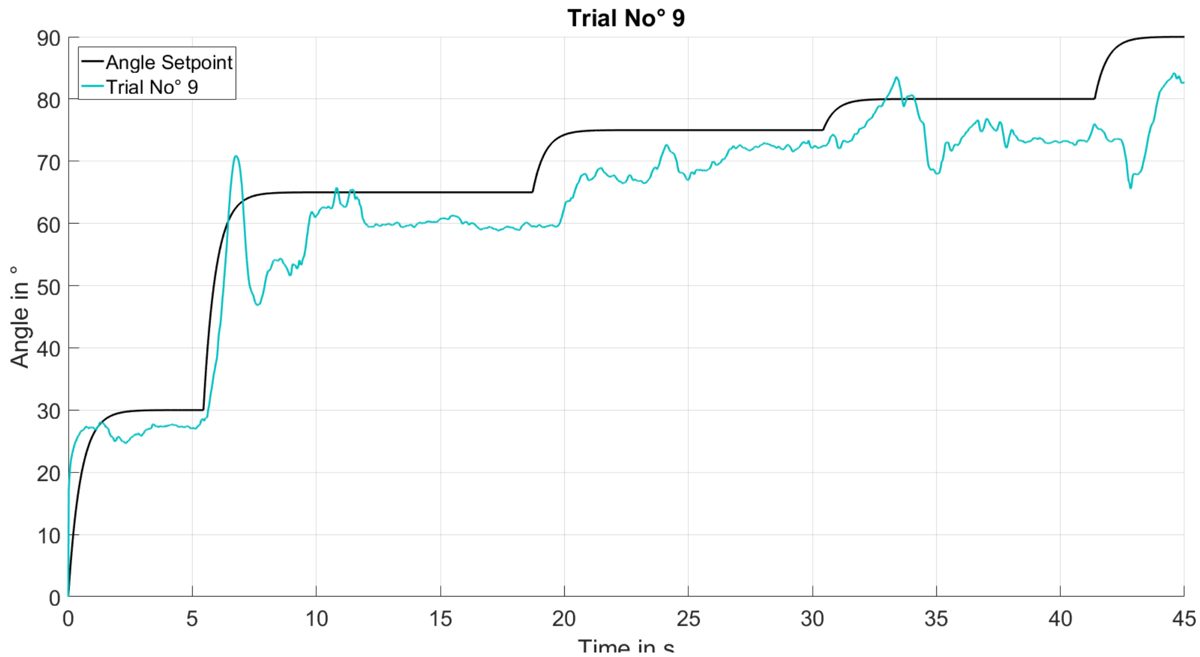

Fig. 11 - Best trial

Another finding is an overshoot of the armmovement, ones the desired control angle reaches about 90°. (Figure 12)

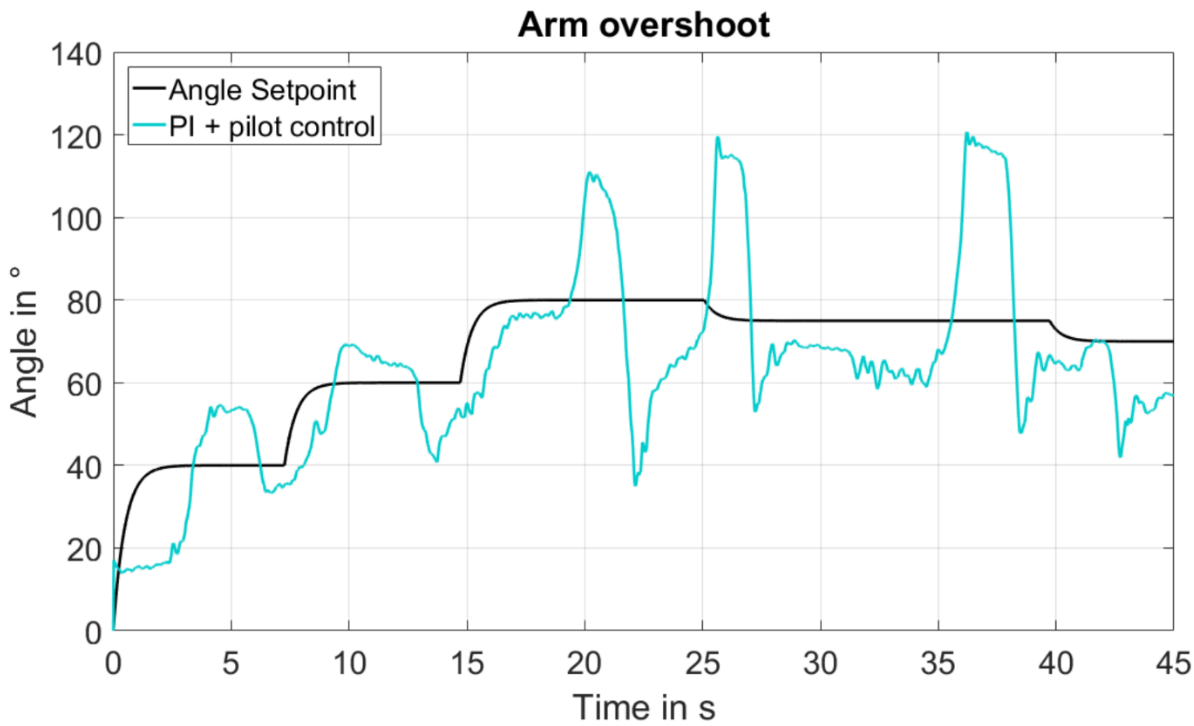

Fig. 12 - Angle overshoot

5 Conclusion

We could show, that control of an elbow joint angle by FES is possible in a range of 15-90 degrees with a standard deviation of the angle error lower than 5 degree.

Angles lower than 15 degrees were not archived, as the elbow in its relaxed hanging position is flexed to about 15 degrees. Decreasing the angle requires force from the flexor (triceps brachii).

The observation made considering the arm to overshoot when the control angle reaches more about 90 degrees, supposedly is due to the decreasing arm torque (caused by its mass) with angles increasing above 90° (see sin function in arm model). If the necessary torque for a static position of 90° is exceeded by the FES, the arm will accelerate towards higher angles, this will result in even smaller torques required for standstill. As the FES and the muscle model delay in time (see dynamic muscle behaviour in section 3.2), the current torque is too high at every point in time, causing the arm to overshoot.

This issue, might be solved by stimulation of the antagonistic muscle (triceps brachii), perhaps in a method of co-contraction.

Against our expectations, even though the simulation would suggest the pilot control to increase the systems stability, the standard angle deviation remained quite the same as with PI control only. This may be the result of damping in the system (i.e. joint friction and especially the co-contraction caused by the FES – see ANSYMB I project).

6 Literature

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