AN FES-ASSISTED GAIT TRAINING SYSTEM FOR HEMIPLEGIC STROKE PATIENTS BASED ON INERTIAL SENSORS

N.-O. Negård [*,**] **T. Schauer** [**] **R. Kauert** [***] **J. Raisch** [**,*]

[*] *Max Planck Institute for Dynamics of Complex Technical Systems, Sandtorstr. 1, D-39106 Magdeburg, Germany*
E-mail: negaard@mpi-magdeburg.mpg.de

[**] *Technische Universität Berlin, Fachgebiet Regelungssysteme, D-10587 Berlin,Germany*

[***] *HASOMED GmbH, Paul-Ecke-Straße 1, 39114 Magdeburg, Germany*

Abstract: An inertial sensor mounted on the foot of the affected body side represents an alternative to traditional foot switches in Functional Electrical Stimulation (FES)-assisted gait rehabilitation systems. The inertial sensor consisting of 3 gyroscopes and 3 accelerometers can be utilised to detect gait phases which can be applied to synchronise the electrical stimulation with the gait. Additionally, the sensor can be applied to estimate orientation and 3 dimensional movement of the foot. Based on the estimated orientation and linear position several movement parameters can be defined. The most important are the foot clearance, which is defined as maximal distance between foot and ground, and the sagittal angle of the foot in relation to the ground at the time as the heel hits the ground. In this paper we describe a practical system for FES-assisted gait training based on inertial sensors where the electrical stimulation is triggered by the gait phase detection and the stimulation intensity is automatically tuned by feedback of movement parameters.

Keywords: Electrical stimulation, Inertial sensing, Biomedical systems.

1. INTRODUCTION

The impact of stroke on the life of an individual can be dramatic both mentally and physically. Physically the motor control of one body side may be detoriated. Such detoriated motor functions can be improved by training. Electrical current pulses can be used to excite intact peripheral nerves, and then cause muscles to contract. This muscle contraction will generate muscle forces and a corresponding joint torque leading to a body movement. Normally, these activations are controlled by the brain or the spinal cord. In cases of dysfunction, like in individuals with stroke or in some other types of upper motor neuron lesion, this normal activation is not possible. In such cases, the activation can be generated artificially as described. Use of electrical stimulation with the intention to restore useful body movements is called Functional Electrical Stimulation (FES). Since Liberson *et al.* (1961) for the first time applied electrical stimulation to elicit the withdrawal reflex during the swing phase, many systems for FES-assisted gait training and Drop

Foot Stimulator (DFS) systems have been designed. In order to trigger the stimulation gait phases must be detected, either as a simple detection of heel-off or a more refined detection of several phases.

Gait Phase Detection (GPD) systems have already been developed where the gait cycle is divided into several phases. The number of these phases can vary as well as the definitions of these. There is no standard terminology in the literature but the definition by Perry (1992) is the most used. By her definition the gait is divided in eight phases for each leg. These eight phases are divided according to functional tasks of the gait and consist of initial contact, loading response, mid stance, terminal stance, pre-swing, initial swing, mid-swing and terminal swing. In gait phase detection systems the sensors used force a limitation of the phases which are possible to detect, and normally a less refined detection with typically four phases is applied: stance, pre-swing, swing and loading response.

In methodology, there are mainly two different approaches to gain the gait phases, the first one is a rule based approach where the gait phases are defined as states of a finite state machine and the transitions between states are logic functions of the sensory input (Papas *et al.*, 2001; Dai *et al.*, 1996; Willemsen *et al.*, 1990; Sabatini *et al.*, 2005; Kotiadis *et al.*, 2004). The second approach is from the methodology completely different, instead of clearly defined rules for transitions, the detection system is represented as a black box where the gait phase is the output and sensory information is the input (Williamson and Andrews, 2000; Ng and Chizeck, 1997). Such black-box systems are usually related to machine learning technics, fuzzy logic systems or neural networks. The advantage of such systems is that they can be trained to accurately and robustly detect gait phases for one subject. The disadvantage is that the system has to be trained for each subject separately with some sort of reference detection system, possibly by foot switches or manually by a hand switch. This is a time-consuming procedure that has to be redone as the gait rehabilitation progresses and consequently the gait changes.

Different sensors have been used to detect gait phases. Foot switches based on Force Sensitive Resistors (FSR) have traditionally been applied for triggering stimulation. Usually, the foot switch is attached under the heel and triggers the stimulation as the heel lifts of the ground. Later, combinations of several foot switches attached to different positions underneath the foot have been applied (Papas *et al.*, 2001) in order to improve the robustness of the detection. Because of the short life span and lack of mechanical robustness of foot switches, other sensors have been investigated as replacement. In Willemsen *et al.* (1990) four accelerometers were used to measure the radial and tangential acceleration of the shank segment. A rule based algorithm was developed to detect four distinct gait phases with the emphasis on detecting heel off as this is essential in a peroneal nerve stimulator. This algorithm worked fine for three out of four patients, but for the fourth patient the heel-strike was constantly detected too early due to disturbances. Other alternatives for triggering stimulation like goniometers measuring hip-, knee- and ankle-joint angles (Ng and Chizeck, 1997) have also been proposed. Recently, the interest in using a combination of gyroscopes and accelerometers has grown for the purpose of detecting gait phases. Kotiadis *et al.* (2004) was using an inertial sensor. Although measuring with a complete inertial sensor, only the 2 accelerometers in the sagittal plane and a gyroscope measuring angular velocity in the sagittal plane were considered in that work. Contrary to our work, the sensor was fixated on the shank segment just below the knee. A similar placement of tilt sensors in (Dai *et al.*, 1996) showed that this sensor configuration leads to a bad differentiation between gait movement and standing up/sitting down. A similar study was done by Sabatini *et al.* (2005) where additionally spatial gait parameters were estimated. To the authors' knowledge no GPD system utilising all gyroscopes and accelerometers in an inertial sensor unit has been developed until now.

Although two accelerometers and one gyroscope are collecting the most valuable information, the remaining sensors might improve the robustness and accuracy. Furthermore, an inertial sensor unit provides more information than a reduced sensor leading to a more accurate estimation of the foot velocity and position. These derived signals can also be used in a gait phases detection system, possibly making it more robust. When using a full inertial sensor, another advantage arises; how the sensor is attached to the foot does not influence the performance of the gait phase detection system anymore, as the sensor is able to find its own orientation.

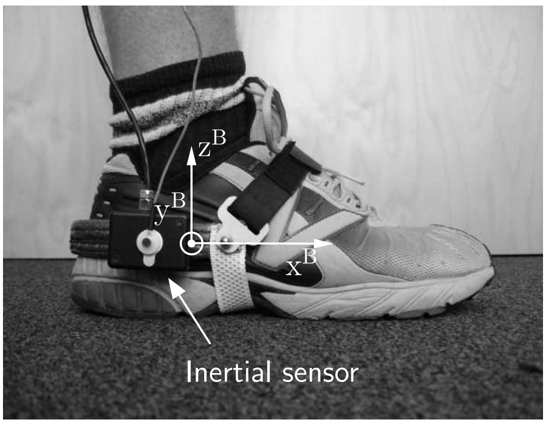

Figure 1. Experimental setup: Inertial sensor attached to the foot segment by a bracket.

In this paper, a practical FES-assisted gait training system is described. Algorithms for estimating foot movement and detecting gait phases as well as strategies for electrical stimulation are presented.

2. METHODS

A prototype FES-assisted gait training system has been developed. An inertial sensor system is used as sensory input, namely the RehaWatch system consisting of two miniature Inertial Measurement Units (IMU) and a Digital Signal Processing (DSP) unit. An IMU consists of three accelerometers and three gyroscopes measuring angular rate and acceleration about three orthogonal axes. The sensor system was developed by the Fraunhofer Institute for Factory Operation and Automation (IFF), Magdeburg (Germany), and the company HASOMED GmbH. Sensor signals are sampled with a frequency of 500 Hz. The algorithms described in this section do only assume one sensor on the disabled body side.

Before sampling, the signals are filtered through an analogue Butterworth filter with a cut-off frequency of 100 Hz. The measurement range of the accelerometers is ±4 g and the range of operation of the gyroscopes is ±700 [deg /s]. The measurements are transfered onto a laptop through an USB-interface.

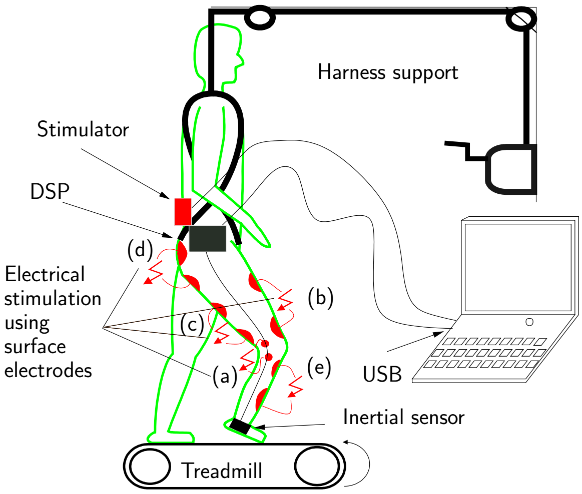

Figure 2. The muscles and nerves stimulated during FES-assisted gait training: (a) peroneal nerve, (b) quadriceps, (c) hamstrings, (d) gluteus maximus and (e) tibialis anterior.

A 8-channel stimulator[1] is connected through an USB-interface and is controlled by a special protocol called ScienceMode[2].

For straightforward testing of new stimulation strategies a MATLAB/SIMULINK user interface was written where new stimulation patterns can be easily realised. The algorithms described in the following sections have been implemented in C++ and are running on a laptop with Linux as operating system.

1 http://www.rehastim.de/
2 http://sciencestim.sourceforge.net/

2.1 Orientation estimation

Position and orientation of the foot can be estimated from an IMU when attached to the foot (cf. Fig. 1). The rotation of the sensor/foot can be found by integration of the angular velocity measured with the gyroscopes. It is important to have an accurate estimate of the orientation because this is the basis for calculation of the foot movement. As the pure integration will unavoidably drift off after a short time, a Kalman filer has to be applied to avoid drift (Negård *et al.*, 2005*a*).

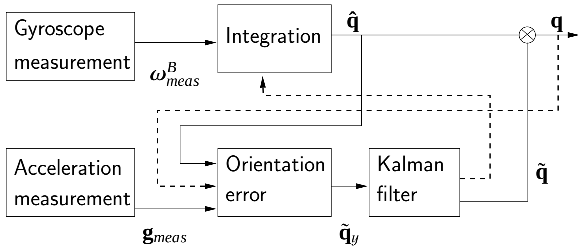

Figure 3. Block structure of the indirect Kalman filter

The structure of the Kalman filter is shown in Fig. 3. The filter is an indirect filter where the states are the error in orientation and biases of the angular velocity measurement. The accelerometers measure the gravity vector as long as the sensor is not accelerated which is nearly the case in the stance phase. The gravity can be used to estimate the orientation and this is later compared with the orientation estimate from the angular velocity integration giving the measurement input $\tilde{\mathbf{q}}_y$ to the Kalman filter. In the Kalman filter the bias of the gyroscopes and the orientation error $\tilde{\mathbf{q}}$ are calculated. This estimate is used to correct the orientation $\hat{\mathbf{q}}$ obtained by simple integration. From the orientation $\mathbf{q}$ of the sensor, angles between the foot and the ground in the sagittal and the coronal plane can be easily extracted.

2.2 Movement parameters

By use of the obtained orientation, the acceleration can be transformed into a global coordinate system and the foot movement can be estimated through a double integration of the transformed acceleration. For every stride the integration is started at heel off and continued until the foot flat event. In order to improve the accuracy, constraints on the integration are introduced. The velocity of the sensor is assumed to be zero at the beginning and at the end of a step. Furthermore, the position in the vertical direction is zero before and after a step by the assumption that the subject is walking on a horizontal surface. These constraints can be imposed on the integration by the introduction of an artificial bias on the acceleration measurement and more accurate position estimates can be calculated. In Fig. 4, position and orientation trajectories calculated from an inertial sensor unit for a healthy subject are plotted against the gait cycle percentage for three steps

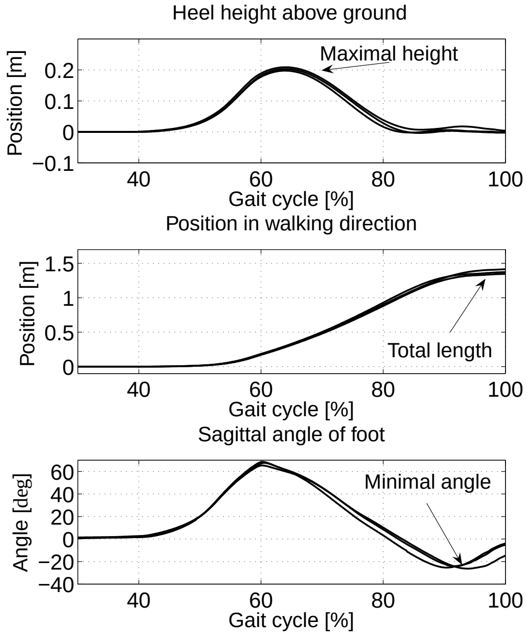

Figure 4. The upper graph shows position above the ground estimated from inertial sensor measurement. The middle graph shows the position in the walking direction, and the lower graph shows the estimated angle between the foot and the ground in the sagittal plane.

(during pre-swing and loading response phase). Movement parameters can be defined from the orientation and the position estimates as follows:

- *Foot clearance*: the maximal distance between heel and ground in the vertical direction during swing phase
- *Step length*: the total length of one step in the walking direction
- *Foot angle at heel-strike*: the angle between foot and ground in the sagittal plane at the moment as the heel hits the ground.

2.3 Gait phase detection

A gait phase detection algorithm has been developed where the gait cycle is divided into four distinct gait phases: stance, pre-swing, swing and loading response. These phases can be represented as a state machine with four states similar to the state machine described in (Papas *et al.*, 2001). The difference to that paper is the type of sensors applied, with the consequence that the transitions between states are different. The algorithm allows 6 transitions between the states (cf. Fig. 5).

Based on the angular velocity measurement a coarse detection is done whether the sensor is at rest or if it is moving. The same detection is also done for robustness purposes using the acceleration measurement. These binary variables are denotes as $x_{a,rest}$ for the accelerometers and $x_{g,rest}$ for the gyroscopes. The logic value one is indicating that the sensor is at rest

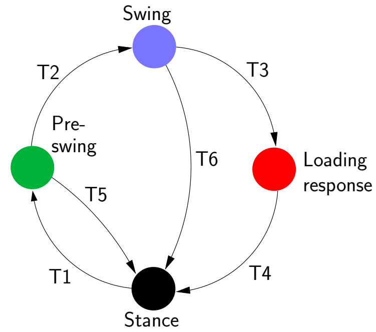

Figure 5. Gait phase detection system represented as a state machine. The gait phases are represented as 4 states where 6 transitions between the states are possible.

and zero is indicating that the sensor is moving. The transitions between the states have the following conditions:

T1: stance → pre-swing
In the stance phase, the only transition which can occur is to the pre-swing state. This is done when both $x_{a,rest}$ and $x_{g,rest}$ are indicating a movement:
$(\overline{x_{a,rest}}) \wedge (\overline{x_{g,rest}})$.

T2: pre-swing → swing
In the pre-swing state the algorithm anticipates the transition to the swing state. The condition for the transition to the swing phase is that at least one of the sensors is not indicating rest, and that the rotation of the foot around the y-axis changes from positive (in the pre-swing state) to negative direction:
$((\overline{x_{a,rest}}) \vee (\overline{x_{g,rest}})) \wedge (\dot{\omega}_y < 0)$.

T3: swing → loading response
In the swing phase the algorithm awaits the transition to the loading response phase which begins with the first contact of the foot with the ground. This transition is detected if the total acceleration of the foot in a global coordinate system is bigger than a certain threshold.

T4: loading response → stance
After the loading response the next phase is stance which begins when both front and rear part of the foot touch the ground. This event is detected when both the accelerometers and the gyroscopes are indicating rest. The transition condition becomes
$(x_{a,rest}) \wedge (x_{g,rest})$.

T5: pre-swing → stance
If the subject lifts the heel and then put it back on the ground, is this event detected as a transition from pre-swing back to stance. This transition is detected when both the accelerometer and the gyroscopes are indicating rest. The transition condition becomes
$(x_{a,rest}) \wedge (x_{g,rest})$.

T6: swing → stance
In certain gait patterns where the proband hits the ground very softly, the transition T3 does not occur. When this is the case, a direct transition from swing to

stance is useful. This event is detected when both the accelerometers and the gyroscopes are indicating rest. Further requirements are that the rotational velocity around the y- axis and its derivative are close to zero. The transition condition becomes
$((x_{a,rest}) \wedge (x_{g,rest})) \wedge (||\dot{\omega}_y|| < \delta_1 \wedge ||\omega_y|| < \delta_2)$.

2.4 Stimulation strategy

As the FES training system is implemented on a PC platform, different frequencies can easily be realised. Orientation estimation and gait phase detection are performed with the same sample time used in the inertial sensor unit (typically 500 Hz). After a completed step, detected by the gait phase detection system, the 3D movement trajectory of the foot is calculated following the algorithm described in Section 2.2 using buffered sensory data of the last step.

After a completed step, temporal information of the gait like cadence is calculated and then averaged over the last three steps and the result is used in the pattern generator. Different patterns can be programmed where transitions of phases are used to trigger the stimulation, and duration can be easily programmed to be a percentage of certain gait phases. The amplitude can either be chosen to be constant or to follow an arbitrary curve scaled to the desired duration. In Fig. 6, a typical stimulation pattern is shown. The peroneal nerve stimulation is triggered by the detection of preswing and lasts until the loading response. Hamstrings stimulation is also triggered by the detection of preswing phase but is turned off earlier as the peroneal nerve stimulation. Furthermore, quadriceps can either be stimulated in the swing phase in order to improve the knee extention or in the stance phase to improve the stability. The gluteus maximus can in some cases be stimulated during the stance phase in order to stabilise the hip.

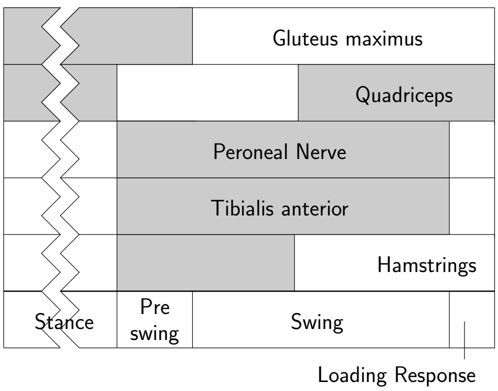

Figure 6. A typical stimulation pattern for FES-assisted gait training. Stimulation periods are indicated by grey bars.

Another possible stimulation configuration is the stimulation of the peroneal nerve in combination with the tibialis anterior whereas the latter is stimulated at the end of the swing phases. Peroneal nerve stimulation takes place with a frequency of 60Hz while muscles are stimulated with 20 Hz.

2.5 Feedback control

The movement parameters defined in Section 2.2 can be used to estimate the quality of the foot movement and to determine the required stimulation intensity. For normal gait these parameters vary between persons depending on height, gender and individual gait style, but remain fairly constant for one specific person. By assuming a relationship between a specific stimulation channel and one movement parameter, the possibility arises to control the gait movement by varying the stimulation intensity on a gait cycle basis. This is done by keeping the stimulation intensity constant during one step and updating it before the next step by evaluating the movement parameter. Foot clearance can be controlled by adjusting the stimulation intensity of the hamstrings, and the angle of the foot before touching ground is controlled by stimulation of tibialis anterior. A discrete-time PI-controller was designed for adjusting the pulse width for the above mentioned stimulation channels by using the movement parameters as feedback. In this scheme the duration, frequency and current are set to constant values, and only the pulse width is controlled by feedback. The relation between pulse width and the related movement parameter can be considered to be a linear relation

$$y[k] = bu[k] \tag{1}$$

where $u[k]$ is the normalised non-saturated pulse width and $y[k]$ is the controlled movement parameter of the gait cycle k. A discrete PI-controller can be described like this

$$u[k] = u[k-1] + q_0 e[k] + q_1 e[k-1] \tag{2}$$

with

$$e[k] = r[k] - y[k],\ q_0 = K(1 + \frac{1}{2\tau_i}),\ q_1 = K(-1 + \frac{1}{2\tau_i}) \tag{3}$$

where $r[k]$ is the reference value of the movement parameter, K is the control gain and τ_i is the integration constant.

3. RESULTS

Gait phase detection and movement parameter estimation have been validated with intact subjects and stroke patients. The estimated movement parameters, e.g. foot clearance and step length, were compared with a reference measurement system, and a mean error of less than 5 % and standard deviation of the error less than 5 % were shown for 2 hemiplegic subjects walking on a treadmill. Furthermore, the gait phase detection system also showed a very good performance as all steps were successfully detected for both patients (Negård *et al.*, 2005*a*). It must be mentioned that these results are preliminary and more data must be collected in order to confirm them.

The proposed feedback strategy in Section 2.5 has been tested in simulations by using a mathematical model of the free swinging leg. To illustrate the performance of the controller the reference value for the foot clearance was changed after 20 cycles. The results from this simulation trial are shown in Fig. 7. A more detailed explanation of the model and the results can be found in (Negård *et al.*, 2005*b*). In the simulations,

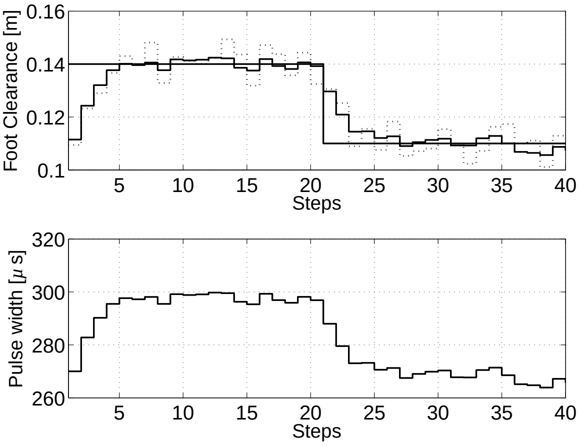

Figure 7. This figure shows the results of a simulation trial. The upper graph shows the foot clearance, the solid lines are the real foot clearance and its reference while the dashed line is the measured noisy foot clearance. The lower graph shows the stimulation pulse width.

the foot clearance could be controlled quite accurately by stimulation of the hamstring muscle group even in the presence of noisy measurements. When a step in the reference value was applied, it took about five cycles before the foot clearance was settled to the new reference value.

4. DISCUSSION AND CONCLUSIONS

Real-time control of intensity and automatic tuning of stimulation pattern have received little attention in the clinically applied FES-assisted gait rehabilitation. Any practical system must be easy to operate by physiotherapists and/or patients. Therefore, the choice of the sensors is important. Foot switches have been the de facto standard until now, but they do not provide any useful information for feedback control of stimulation intensity. Goniometers are more useful in that respect, but a time-consuming procedure has to take place to attach them and to calibrate the angular measurement before it can be used in any feedback loop. On the other hand, inertial sensors can replace foot switches for triggering purposes, and can be applied for feedback control. The system described in this paper does not require any extensive and time-consuming calibration of the sensors before taken into use. The algorithms are not depending on an exact initial orientation as long as the sensors are rigid mounted to the foot. The Kalman filter automatically detects the orientation of the sensor by use of the gravity measurement and will consequently deliver correct movement parameters. An extensive platform for testing feedback algorithms and stimulation strategies in FES-assisted gait training has been developed, but more experiments with stroke patients have to be performed in order to validate the algorithms described in this paper.

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