An EMG-based Assistive Orthosis for Upper Limb Rehabilitation

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Abstract: In this work an upper limb active orthosis for assistive rehabilitation is presented. The design and torque control scheme of the orthosis that take into account important aspects of human rehabilitation, are described. Furthermore, first results of successful muscle activity detection and processing for the operation of the orthosis in two movement directions are presented. The proposed system is the first step towards an adaptive support of patients with respect to the strength of their muscle activity. To allow an adaptive support, different methods for EMG analysis have to be applied which allow to correlate muscle activity strength with the recorded signal and thus enable to adapt the support of the orthosis to the needs of the patient and state of therapy.

1 INTRODUCTION

The ability to move is one of the most important characteristic of life and is determined by the functions of our muscles, which are required in tasks like breathing, blood circulation, eating and locomotion. In other words, in any kind of activity. For human locomotion and activities a complex and fine-tuned musculoskeletal system has evolved. However, if functions of the muscular system are impaired or disabled it can not only have far-reaching consequences on the personal professional and social life, but also on psyche of the affected person and their families.

In Henze (Henze, 2007) it is stated that loosing the ability to move (even of one single extremity) is in many cases associated with a drop in independence and therefore reduces the quality of life of the person. Generally, motor restrictions are often the result of neurological disorders.

These can be caused by illness, accidents, or birth defects. In this context stroke plays a major role, since it is one of the most common causes of neuromotor disorders and permanent disabilities in western civilization (Deaton et al., 2011).

The inability to move the affected arm or even to use it in a coordinated way is a very common and serious consequence of a cerebral stroke. About 40 % of the affected people suffer from a non-functioning upper extremity. Therefore, it is not surprising that in the recent years, several studies and findings on stroke rehabilitation were published (Albert and Kesselring, 2012; Platz and Roschka, 2009).

In this field, rehabilitation robotics has also made great progress and is currently subject of many research projects (Loureiro et al., 2011). The aim of robotic systems in the context of neuromotor rehabilitation is to optimize the rehabilitation process and to support the therapist in labor-intensive therapies.

Further, there is the ambition to re-enable patients to execute self paced movements of the paretic extremity using their movement intention. This is thought to increase the patients motivation and to support processes that are important for neuronal plasticity [1] (Brewer et al., 2007).

A central, but time consuming part of stroke rehabilitation is the process of re-learning directed hand and arm movements according to the patients needs. Hence, we want to introduce a rehabilitation device for the upper extremity, which supports stroke patients and therapists in their daily rehabilitation routine. This device is an active elbow orthosis (see Figure 1) with one degree of freedom (see Section 3). A patient-specific control can be realized by processing

[1]Neuronal plasticity is the ability of brain to reorganize itself by forming new neural connections. This form of adjustment allows the brain to compensate injury and disease and to adjust activities in response to new situations or to changes in the environment (Johnston, 2009).

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the patients muscle activity measured with the electromyogram (EMG) (see Section 5). In the long term, this device could be used for the entire rehabilitation process. For this, three different functionalities can be provided by the active orthosis, namely passive, active-assisted, and active-resisted modes of operation (Gomez-Rodriguez et al., 2011). In addition, the progress of therapy can be evaluated by monitoring and analyzing the muscle activity via EMG.

Figure 1: The active orthosis is designed to be easily worn by the user. The carrying system distributes the weight, increasing the comfort.

2 APPLICATION ENVIRONMENT

There are several approaches for EMG-controlled active elbow orthotics. For example the *mPower 1000* system (Myomo® Inc, Cambridge, USA) is a commercially available EMG-controlled orthosis for the upper limb. A german research group is developing an active orthosis for paraplegic persons (Schmitz et al., 2011). In contrast to these approaches, the proposed device offers a unique combination of sensors and planned functions.

The therapeutic (long-term) goal of the orthosis is recovery of lost motor functions of the upper extremity after neurological diseases. As mentioned before, the human brain is able to compensate functional impairment. This requires intensive and early training after, e.g., stroke. Therefore, it is important to design a system, which motivates the patient for a constant training.

The device can enable patients to perform the following exercise modalities:

- **Early and Intensive Practice.** Start of the arm rehabilitation, e.g, few days after acute stroke with a high intensity, when indicated.
- **Repetitive Practice.** Repetitive target movements across various sequences.
- **Task-oriented Training.** Exercise oriented on everyday life situations, e.g., in an exercise-kitchen.
- **Independent Training.** Therapeutic treatments with intermittent supervision by the therapist.

These therapy modalities are based on established and evidence-based rehabilitation methods (Platz and Roschka, 2009).

The goal is to achieve a therapy session comparable to a guided session by a therapist, without having him at site.

In the early stage of treatment the device can be used to passively move the patients arm. With therapy in advanced stages the residual muscle activity will be measurable again. This low residual activity may not be sufficient for moving the arm, but result in myoelectric signals. By measuring these signals with EMG, they can be used to detect the patients movement intent.

Further these signals can be used to move the patients arm in a self motivated way. This kind of treatment can support processes that underlay neuronal plasticity. In later stages of treatment the patient should regain more and more muscle strength. Therefore, it is planned to adjust the assistance level of the device via the measured muscle activity, in a way that higher muscle activity leads to a lower level of assistance.

3 DESIGN AND MECHANICAL STRUCTURE OF ORTHOSIS

In this section the mechanical design of the active orthosis is presented (see Figure 2).

The orthosis is designed with five degrees of freedom, four passive joints are required to compensate misalignments and one actuated joint to support the flexion/extension movement of the elbow joint. The active joint is driven by a 24 *V Maxon A-max 22* DC-Motor with a *333:1 Maxon* planetary gear and a 4:1 worm wheel gear. For a natural force interaction, safety reasons and to measure the applied force

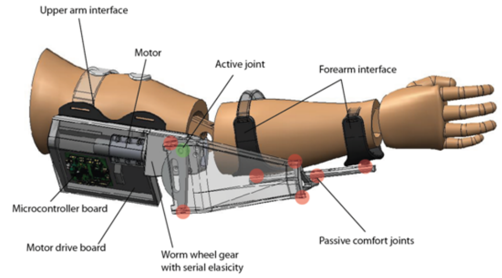

Figure 2: Mechanical design of the active Orthosis. The red dots represent the positions of the passive degrees of freedom.

interaction, the actuated joint is compliant. This compliance is generated via serial elasticity in the worm wheel gear set-up. The worm is axial moveable and centred in the gear via disc springs. In case a load is applied, the worm is pushed to one side and thus, the spring is compressed on this side. The position of the worm wheel is measured with a *Bahluff* inductive sensor. In this way the applied load can be calculated (see Section 4). The position of the joint is measured with an *IC-Haus-MH* position encoder.

Furthermore, the used electronics consist of a *STM32F103VE* microcontroller, offering several data acquisition (GPIO) and communication (USART, CAN-bus) ports, and a *BD6232* PWM H-Bridge driver. The current used DC-Drive can generate a torque of about 16 *Nm*.

To avoid any danger for the user, various safety aspects are considered. Therefore, the natural working range of the human elbow is limited by mechanical stops. Furthermore, at too high forces the forearm interface will release from the orthosis (similar to the principle of ski binding).

The active range of motion of the elbow orthosis corresponds to the anatomic workspace of the human joint and is individually adjustable to each subject.

Since an additional and unilateral load can represent a major influence on, e.g., neurological patients, the orthosis' weight with respect to the user must be kept as low as possible. Therefore, the orthosis' materials is a combination of carbon reinforced plastics and polyamid *PA6*, for a lightweight, robust and stiff design. Additionally, a carrying system was developed, which distributes the weight of the device on both shoulders.

4 CONTROL SCHEME OF THE ORTHOSIS

Several research groups have described robotic devices for upper limb rehabilitation and their strategies to control them in a user-oriented way. In (Rosen et al., 2001) the torque applied to the elbow joint of an upper extremity exoskeleton is measured via a load cell, while the set torque is calculated via muscle models. In a second step, the authors derive four performance indices, in order to calculate the magnitude of support by the exoskeleton from EMG data. In (Andreasen et al., 2005) an impedance control scheme is implemented. Two load cells in series estimate the joint torque which is fed into a dynamic impedance function.

In the following the torque control system of the proposed active orthosis will be presented. This can be visualized in the simplified block diagram in Figure 3.

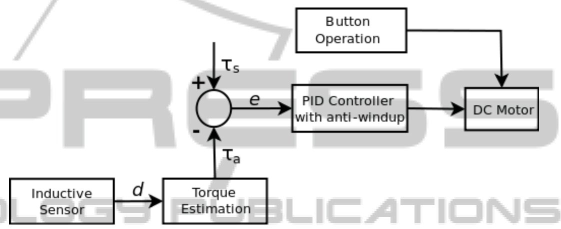

Figure 3: Block diagram of the torque control loop.

The general control structure is designed to be cascaded, while the main- and inner loop of the control architecture is a torque control loop. As shown in Section 3, the DC-drive of the device is provided with two disc springs performing the serial elasticity of the drive. These springs deflect when load is applied to the joint. One is used for movements that are upwards directed and one for movements that are downwards directed. The inductive sensor detects this deflection d. With these measurements it is possible to obtain a nearly linear function between spring deflection and the actual torque applied to the joint, τ_a. The curve, shown in Figure 4, was empirically determined applying several load (torque) values to the joint, and matching this load values with the resulting spring deflection.

The set (desired) joint torque τ_s at this point is fed externally, from a computer via *USART* port. The difference between these two torques is the control error e, which is propagated into an anti-windup PID controller. For more information about anti-windup controllers and methods, please refer to, e.g., (Bohn and Atherton, 1995). The controller computes the voltage for the motor needed to reach the desired torque. The performance of the control system was verified with weight discs in order to simulate values for τ_a, and giving the corresponding τ_s to the system, resulting in an acurate balancing of the weights. Furthermore, the resulting measured torque was compared with the deflection-torque curve depicted above. Figure 5 il-

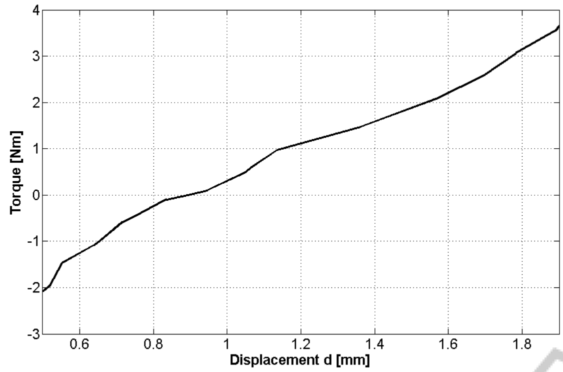

Figure 4: The calculated joint torque - deflection curve.

Figure 5: Setup for control experiments.

lustrates the experimental setup.

The torque control loop already allows to use the upper limb orthosis in a free-running mode (with $\tau_s = 0\,\mathrm{Nm}$), or in other words, to use the device as a completely passive one.

Alternatively, the orthosis can be manually operated via two buttons at any time, supplying a constant voltage of $\pm 15\,\mathrm{V}$. This allows corrections and re-positioning of the joint if needed.

5 DETECTION OF MOVEMENT INTENTION BY EMG ANALYSIS

This section describes how muscle activity measured by the EMG is used as a control signal for the active orthosis. EMG signals are measured at two muscles on the upper right limb (same arm at which the orthosis is used) named M. biceps brachii and M. triceps brachii. *Ag/AgCl* electrodes are used in a bipolar

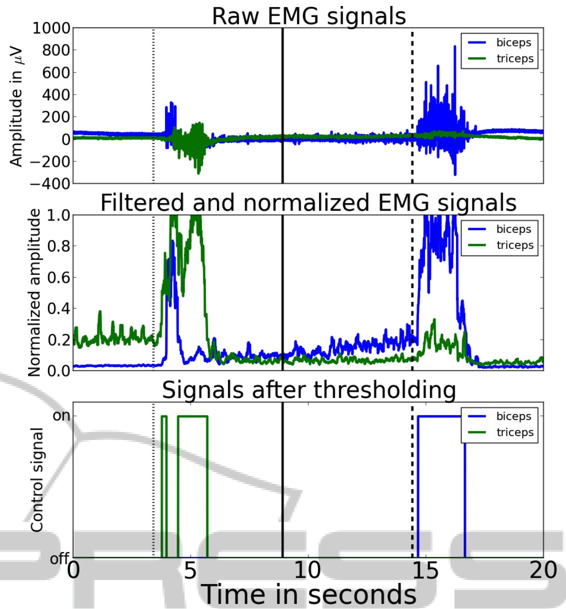

Figure 6: Resulting signals after preprocessing and thresholding: (top) Raw EMG signals from biceps (blue) and triceps (green), (middle) filtered and normalized signals, and (bottom) results of the thresholding algorithm.

setup. The signals are amplified and digitalized by a BrainExG MR (Brain Products GmbH, Germany) amplifier and transfered to a computer with a sampling frequency of 5000 Hz where they are saved or directly processed.

The acquired signals are filtered. For this, a filter that is based on the standard deviation (blind review Ownpaper, 2013) is used. For this filter a sliding window of 100 ms is passed along the signal. The standard deviation of this window is assigned to the last sample. This is done consecutively for all samples of the streamed EMG-signals.

Afterwards the signals are normalized in a range from 0 to 1. Often the normalization is done using the isometric or isokinetic maximum voluntary contractions (Burden and Bartlett, 1999). Here a different approach is used. Basically the signal is normalized with the maximum value obtained after filtering so far. The maximum is initialized with 1.0. If a filtered sample has a higher value than the maximum, this value is assigned as the new maximum. The drawback of this method is, that big artifacts, e.g., produced due to resistance changes at the electrode site, artificially reduce the amplitude of the normalized signals. Therefore, a forgetting factor ε_1 with $\varepsilon_1 = 0.9999$ is used to reduce the maximal value in each time step. Further, a minimum is defined as one fourth of the maximal value. This minimum is again degraded with a forgetting factor ε_2 with $\varepsilon_2 = 0.99999$. The minimum is needed to keep the normalization factor in a value range, where a clear distinction between baseline noise and signal during muscle contraction can be achieved. Without the minimum the normalization factor could degrade to values lower than the baseline noise values. This would lead to a value of close to 1.0 after the normalization in ranges where no movement is performed.

The normalization needs to be calibrated, i.e., there has to be a contraction of each muscle to obtain meaningful values from the function. The control signals for the orthosis are obtained using a threshold algorithm similar to the on used in (DiCicco et al., 2004). Two thresholds are used for determining the on and off phases of each muscle. The on threshold was set to 0.4 and the off threshold to 0.3. This hysteresis is used to prevent continuous on and off switching if the normalized EMG signal has a value close to the threshold. For the off threshold an additional time constrain is used. The mode is only switch from on to off, if the signal is continuously below the off threshold for 20 ms. Further a resting time of 500 ms in between a direction switch has to be maintained. As a last step we had to decide which of the two measured muscles should be preferred, in case that both were active, we decide to set both signals to 0 and therefore to decide for none of the muscle or movement directions.

With this preprocessing in combination with the threshold algorithm, we are able to create three different control commands for the orthosis: (1) to flex the arm (M. biceps brachii is active), (2) to stretch the arm (M triceps brachii is active), and (3) to relax (none of both muscles is active). The control signals are derived on the computer acquiring the EMG data and send to the orthosis via *USART*. All described parameters were chosen by empirical testing on EMG data recorded with the orthosis attached to the subjects arm running in free-runing-mode see Section 4.

In Figure 6 the processing of EMG data, and the thresholding result for an arm flexion and extension are shown. In the top, the raw EMG signals from the biceps (blue) and triceps (green) are shown, the middle illustrates the filtered and normalized EMG signals, and finally in the bottom, the outcome from the thresholding is shown. The three vertical lines, dotted, dashed, and solid denote time points, where the subject was asked to stretch, flex, and relax the arm respectively. The stimuli for those three actions where presented on a monitor and marked in the EMG data. The obtained control signals can be used directly to operate the orthosis, e.g., support the users arm movement.

6 CONCLUSIONS AND OUTLOOK

In this work we presented an active orthosis, its possible application, design and mechanics, and control. To summarize, the system allows to support self initiated movement that are normally executed by both upper arm muscles M. biceps brachii or M. triceps brachii. Patients that are not able to effectively control both muscles can be supported for individual movements. This can be achieved by detecting EMG onset activity. If EMG activity onset is detected, the orthosis actively executes the directed movement corresponding to the active muscle.

The next developmental step is to adapt the strength of support with respect to the strength of muscle activity. To allow this, different methods for EMG analysis have to be applied. These methods (Ajiboye and Weir, 2005) allow to correlate muscle activity strength with the recorded signal and thus, enable to adapt the support of the orthosis to the needs of the patient and state of therapy.

Further, we established a collaboration with a clinical partner. Supported by the medical specialists, the design of the orthosis will be improved in order to follow further sanitary and medical guidelines and to define proper application scenarios. From the control side, it is important to find a way to measure the resistance of the human-robot interaction in order to recognize unvoluntary muscle activity due to spasticity. Further, the possibility of defining torque curves will be analysed. In specific, the development of the applied torque by the user depending on the angular position of the joint can be observed and defined as a mathematical function. Using the inverse of this curve may lead to a more natural movement.

Finally, user-friendliness elements have to be considered in the future. The possibility for therapist (and patients) to adapt important parameters online in an easy way, by means of some sort of intuitive user interface is the goal.

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