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Active Elbow Orthosis

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Abstract This paper presents a novel approach to the design of a motorized rehabilitation device – active elbow orthosis (AEO) – inspired by the principles of robotic exoskeletons. The device is currently designed for the elbow joint, but can be easily modified for other joints as well. AEO determines the motion activity of the patient using a strain gauge and utilizes this measurement to control the actuator that drives the forearm part of the orthosis. Patient activity level is related to a free arm measurement obtained via a calibration procedure prior to the exercise. A high-level control module offers several types of exercises mimicking the physiotherapist. The device was successfully verified by tests on a number of patients, resulting in extended range of elbow-joint motion.

Keywords Rehabilitation Robotics, Active Orthosis, Upper Limb Rehabilitation

1. Introduction

For over 30 years, robotics has been experimentally used to assist severely disabled individuals. Recently a number of applications have also been proposed to deliver robot-assisted rehabilitation therapy. For an introduction to the field see [1]. Most research is focused on gait rehabilitation of patients with gait dysfunction [2], general lower-limb pathology [3] and stroke survivors, with emphasis equally distributed between lower-limb (see extended references review in [4]) and upper-limb rehabilitation [5-7]. Improvement of shoulder-joint integrity using continuous passive motion is illustrated in [8]. Application of robot-assisted arm training in patients with Parkinson's disease was recently reported in [9]. A comparison between EMG-driven and passive devices used on wrist rehabilitation with chronic stroke patients is presented in [10], proving interactive training to be superior to passive training.

However, the rehabilitation process is also essential during post-traumatic treatment after, e.g., inner-joint fractures [11], or following hand transplantation [12]. The main purpose of rehabilitation in these cases is to improve the post-traumatic stiffness that occurs very rapidly after both the injury and surgical intervention [13]. The stiffness can be partially released through additional surgical intervention [14-15], but rehabilitation can achieve motion-range improvement with smaller strain on the patient. At the moment, only passive motorized devices are commonly used with therapy led by trained personnel. When using passive motorized devices, inactivity of the patient degrades the rehabilitation process. With a physiotherapist in place,

the whole process depends on subjective evaluation of the patient's activity. Also, in both cases the patient has to visit the medical facility, as currently used passive devices are of large dimensions and weight, and the physiotherapist rarely visits the patient at home.

To avoid these limitations, active orthosis can be used, preferably in a portable form. It is a smart device that uses a sensor system to measure the activity of the patient, an actuator to perform the desired motion of the driven part of the orthosis, and a control system that controls the actuator based on the patient's activity, current position of the system, type of exercise, and other variables.

Keeping the above-mentioned statements in mind, the requirements for such a device can be formulated as follows:

1. Patient activity detection: the device has to recognize the activity of the patient to perform the motion and react accordingly.
2. Actuation: the device has to be able to move the actuated part of the orthosis with force/torque sufficient to overcome the post-traumatic stiffness of the joint.
3. Bilateral arm usability: the device should be usable for both left and right arm without hardware modifications.
4. Portability: the device must be transportable to the patient's home; a wearable version is not recommended, as the posture of the patient should be kept steady during the exercise.
5. Easy handling: the patient should be able to install and use the orthosis by him or herself after initial training in the medical facility.
6. Exercise control: high-level control subsystem should be able to mimic the essential activities of the physiotherapist with a range of exercise types, voice commands, recommendations and encouragement.
7. Safety: the device must not harm the patient under any circumstances.

This paper describes the design of a rehabilitation device that meets the above-listed requirements. The paper is organized as follows: a design overview is presented in chapter 2 describing mechanical, sensory and actuator subsystems. A high-level control module that handles the exercise itself is shown in chapter 3. Chapter 4 describes the verification of the device as used on patients. Discussion and conclusions are given in chapters 5 and 6.

2. Orthosis design

2.1 Sensing patient activity

An active orthosis can be designed for various joints, but regardless of the type of joint the measurement of the patient's activity is the key input for the control system. A number of methods exist based on various principles; in neurorehabilitation, the following two methods are used the most: Electromyography (EMG) detects the electrical potential generated by muscle cells when these cells are activated [16-18], while mechanomyography (MMG) detects low-frequency vibrations that appear when a muscle is contracted using an accelerometer or microphone placed on the skin [19-21]. MMG offers higher signal-to-noise ratio compared to EMG.

However, both methods suffer from the necessity of installing and calibrating expensive measuring devices. This is in contrast to the requirement for the device to be portable and easily handled by the patient. Therefore, a different way to measure the patient's activity was selected, using a strain gauge sensor.

A strain gauge is a device that measures strain of an object, at present mostly employing a thin metallic-foil meander-like pattern attached to the object. As the object deforms, the foil is deformed as well, resulting in a measureable change of its electrical resistance. Strain gauges can be combined to form a sensing device that measures only certain components of the force applied to the device. Such a measuring device is used as an integral part of the mechanical design of the orthosis, creating a structural element between the forearm and the orthosis drive gear. This way the force applied by the forearm part can be measured, composed of the weight of the forearm assembly, weight of the patient's forearm, and – the point of our interest – the patient activity. Separation of the patient's activity component, crucial for the application, is described in section 3.1.

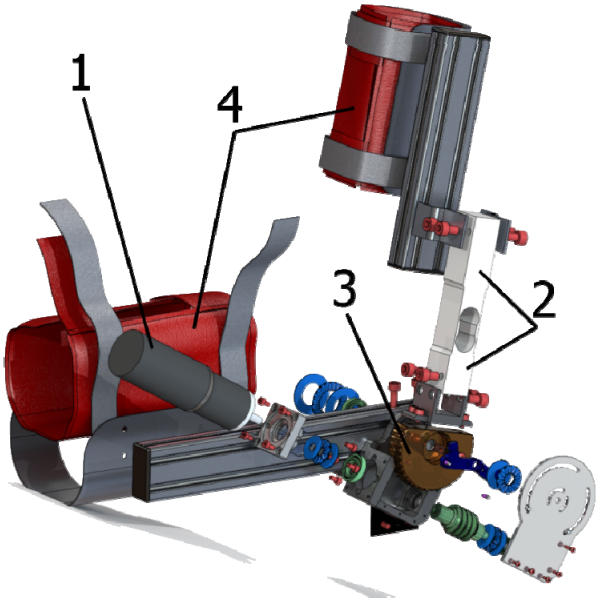

Figure 1. Exploded view of the orthosis. 1 – actuator, 2 – strain gauge sensor, 3 – worm gear, 4 – vacuum pads

Strain-gauge-based sensing is very robust; its temperature dependency is well compensated to make the operational temperature range sufficiently broad. The ability to measure a single component of the force enables simple mechanical design of the orthosis, as the measuring element can be placed parallel to the forearm. The sensor integration is illustrated in Figure 1. The strain-gauge-based single-point load-cell sensor PW6KRC3 with a maximum load of 400 N, produced by HBM, was selected.

2.2 Mechanics and actuation

The elbow joint is physiologically complex – it consists of three separate joints. However, from a kinematics point of view, respecting the needs of rehabilitation, it can be considered a single-degree-of-freedom mechanism, allowing flexion/extension only, while limiting pronation and supination. Therefore, the orthosis is designed as a hinge-joint mechanism.

The frame of the orthosis consists of the arm and forearm parts, both made from aluminium profiles. These are connected through a worm gear providing reciprocal rotary movement invoked by the actuator. The self-locking character of the worm gear provides stiffness to the orthosis, which is essential for proper functionality, especially in a static regime.

The worm gear transmission is driven by a direct-current (DC) motor Maxon RE36 with nominal power of 70 W. Furthermore, the motor is equipped with the planetary pre-gearbox GP32A and a quadrature incremental encoder HEDL 5540. The worm is housed in both axial and radial bearings. The worm wheel is housed on a tenon over one radial ball-bearing and two axial needle bearings. The transmission is dimensioned to the maximum load of 94 Nm and the operational range is -5° to 150°. Details of the gear box are shown in Figure 2.

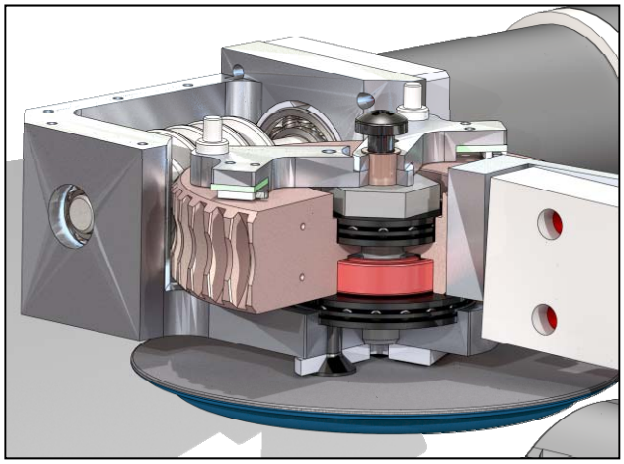

Figure 2. Worm-gear transmission housing detail

The gearbox aluminium body is open and consists of the drive flange and the main body. It houses the worm gear wheels and is rigidly connected to the arm holder. Furthermore, the body includes a mechanism that defines the angular range of the orthosis motion. There are two aluminium stops with end switches; the positions of the stops can be changed as they move on a circular trajectory in a groove – see Figure 3. The angular range's lowest limits that can be set by the stops are 90° for extension and 120° for flexion.

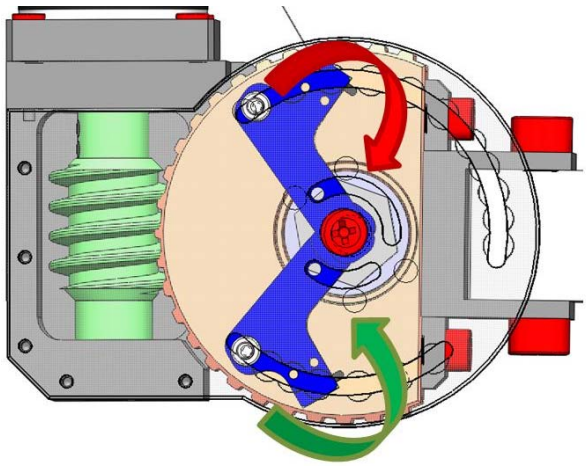

Figure 3. Setting the angular-range safety switches

The strain-gauge-based single point sensor creates a link between the forearm frame and the worm-gear wheel, allowing measurement of the tension generated by a patient's effort to move the forearm. There are holders attached to both the forearm and arm frames, allowing the patient to attach their limb to the orthosis. The holders are adjustable in length, so different patients can be fitted. The soft-tissue-padded holders with Velcro straps that were used initially provided either comfortable or rigid fixation, but rarely both. Therefore, custom-made vacuum pads were designed as a replacement, with vacuum applied after the initial fit to the patient. This type of holder is sufficiently rigid and comfortable at the same time.

To ensure the most comfortable and repeatable seating posture of the patient during the exercise, the device can be connected to a table, with adjustable height. The orthosis can be used on both arms and can be attached to the arm by the patient himself. A photograph of the orthosis prototype is shown in Figure 4.

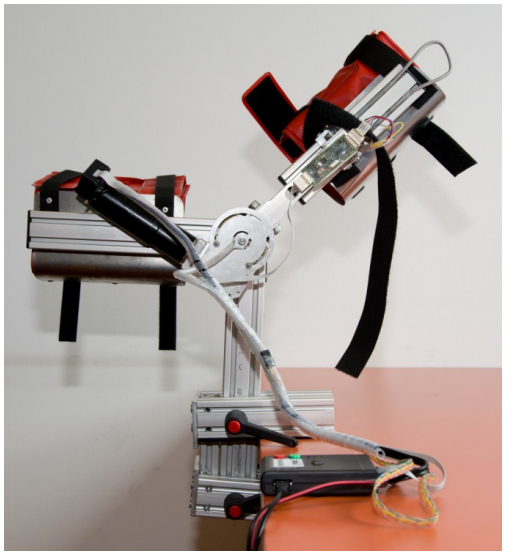

Figure 4. AEO prototype mounted to a table in working position

2.3 Low-level control

The control of the orthosis is divided into two layers. The low-level (LL) layer processes raw data from the strain-gauge sensor and from the incremental rotary sensor on the actuator, and controls the motion of the actuator. It also receives commands from and provides measurement data to the high-level (HL) layer, which is responsible for the overall behaviour of the orthosis. A block diagram with the signal flow is shown in Figure 5.

The main role of the low-level control unit is to provide closed-loop position control of the actuator with speed and acceleration limiting. To ensure safe usage of the AEO it also implements several hard-wired protection mechanisms that stop the motion in case of a hardware failure or an out-of-range orthosis position.

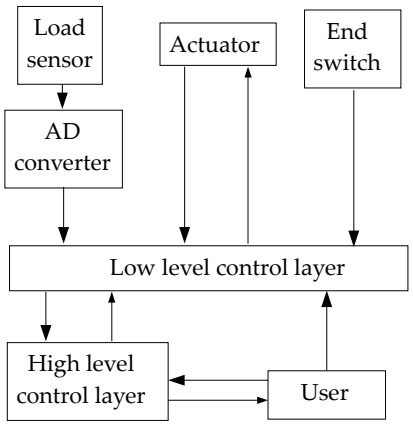

Figure 5. Orthosis block diagram and signal flow

The control unit features an H-bridge power-output stage capable of driving the orthosis actuator; DC motors up to 28V/5A are supported. The output is protected from short-circuiting and high-temperature conditions.
The position of the orthosis is determined by summing the signal from the actuator-mounted quadrature encoder. This position is, however, relative to the orthosis power-on position. Therefore, a homing procedure that allows setting of the reference position (zero) is implemented.

To achieve smooth motion of the orthosis, the position feedback control loop builds upon a speed control loop with defined acceleration coefficients (linear acceleration ramp). The speed governor loop consists of a software-implemented controller with only the proportional and summation terms (PS) derived using the conventional Optimum Module method; its process runs at the frequency of 100 Hz, ensuring flawless regulation.

As implied above, the load cell PW6KRC3 is the primary feedback component of the AEO. It features the common strain-gauge bridge that, with voltage applied to its excitation leads, outputs bipolar-difference voltage signal proportional to the mechanical load (the bridge sensitivity is 2 mV/V). This low-amplitude signal is conditioned by a bridge amplifier before being sampled by a 24-bit A/D converter; the sampling frequency is 66 Hz. The amplifier, A/D converter and a managing microcontroller form an independent measurement unit that is connected to the low-level controller over an I²C bus. This configuration allows connecting of additional feedback devices when needed.

The communication with the high-level control layer is maintained through a USB virtual serial port. To facilitate easy development of the communication messages and prevent malformed data, the LCM library is utilized both for command (from HL) and data (to HL) messages. It provides language-independent definition of the messages and automatic generation of the handler code. This greatly reduces the time needed for the implementation and ensures correct interpretation of the received byte stream.

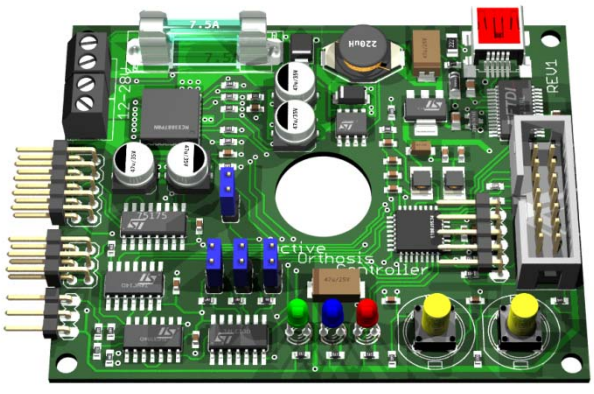

Figure 6. Model of the orthosis controller board

3. High-level control

The high-level control layer (HL) is implemented as an independent application running on a PC under Windows OS. It receives the strain-gauge measurements together with orthosis position and drive power, and provides the LL control unit with motion commands. HL runs in two basic modes: calibration and exercise.

3.1 Calibration

The calibration procedure serves as a basic tool for determining the zero activity of the patient from raw measured data. The aim of the calibration process is to find a representation of zero activity for all angular positions of the orthosis and both motion directions for a particular patient. Not only the weight of the patient's forearm and stiffness of their elbow joint, but also the way the orthosis is attached to the patient play a role. This force-angle dependency is based on sensor readings obtained in free arm motion. The calibration consists of the following steps:

1. Determination of the angular range for a given patient. To do so, the patient moves the orthosis actuated forearm part by pressing buttons on the low-level controller, or moves the orthosis using the HL control application.
2. Free arm motion in the complete range of mobility. The patient leaves the arm free; they neither help the orthosis nor resist the motion. The free arm motion is performed solely by the orthosis actuator with a constant velocity for both directions with a short hold in between. The patient has to be instructed prior to this step.
3. Storing measured data in force-angle form. During the free arm motion, the sensor data are logged and stored for further processing.
4. Building the representation of the patient's zero activity. Logged data are used to calculate a parametric curve for both directions that represents the angle-force function of the free arm motion. This curve is then used during the exercise as the activity baseline for exercise controllers. Patient activity during the exercise is determined as the difference between the measured values for a given angle and the corresponding value of the parametric curve.

Examples of the free arm measurements for an empty orthosis, a healthy person, and a patient are shown in Figure 7.

Polynomial curve fitting was selected as the parametric representation of measured data, using common-least-squares fitting to calculate the coefficients of polynomials. Due to the limited resolution in angle and high data-acquisition frequency, multiple values of force for the same angle commonly appear in the data. The order of the curve is set by the operator; lower orders (2-4) are usually sufficient. Polynomial of third order is used as default. Lower-order polynomials are recommended to prevent unrealistic values outside the measured range, as the angular range may increase during the exercise.

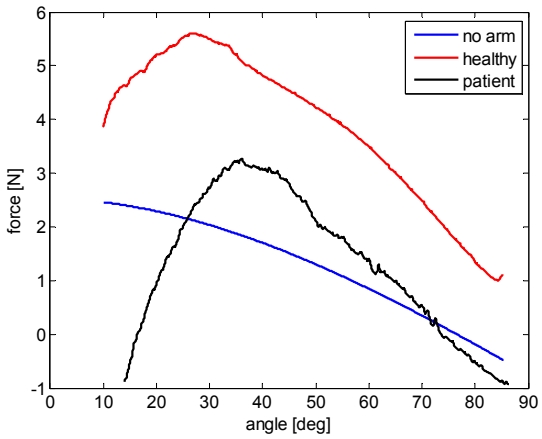

Figure 7. Calibration data measurement. A – orthosis without patient, B – healthy person, C – patient

Examples of raw data and corresponding zero-activity curves, together with the approximated error values, are shown in Figure 8.

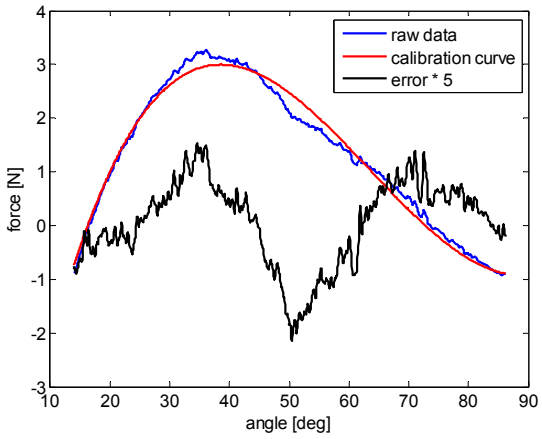

Figure 8. Calibration data-fitting examples

3.2 Exercise

The exercise procedure is the main purpose of the orthosis; it actually performs the exercise with the patient. There are several types of exercise, called exercise modes:

1. Passive mode. In passive mode, the range of motion is given by limiting angles in the up and down positions; the only parameter is the speed, which is constant during the whole exercise. In limit positions, there is an optional hold for a preset period of time. The hold is implemented in all other modes as well.

2. Active mode. Active mode requires the patient to move their forearm in the correct direction to determine the speed of the actuator. The exercise is performed within given angular limits. Each cycle is ended once the limit is reached, or no effort of the patient is detected for a given period of time.
3. Resistance mode. In resistance mode, the orthosis performs the motion within defined angular limits and the patient's goal is to resist the motion, i.e., to push against it. The speed of the actuated forearm part depends on the determined patient's activity; more effort results in a lower speed. The patient's effort is recorded, as during all other modes, and an evaluation is given at the end of each exercise cycle.
4. Override mode. Override mode is the most important mode, as it mimics the most common exercise led by a physiotherapist. It partially behaves as the active mode, but the orthosis does not move within angular limits. Instead, the orthosis moves until the patient's effort drops below a certain value; once this state is reached, the forearm continues to move further for a preset angle to overcome the joint stiffness, while the resistance against the motion is continually measured to prevent injury.

To provide the above-described functionality, a state-machine-based control algorithm is implemented with the key state variables listed in Table 1. The immediate delta force value is determined from the difference between the current reading and the corresponding value of the calibration curve. There are two delta force variables used, the current one and the long-term one. Both variables are calculated as the weighted mean of the immediate delta-force-values history. The current delta force is employed to determine orthosis velocity; it uses triangular shape weights with emphasis on the last value. The length of the buffer is set to cover 500 msec. The long-term delta force uses moving average and is used to determine whether the patient stopped their activity for the current cycle. The length is set to two seconds.

Variable name	*Variable values*
Cycle number	Integer
Motion direction	(up, down)
Exercise state	(active, override, hold)
Delta force current	Float
Delta force long term	Float

Table 1. Key state variables of orthosis high-level control for override mode

The whole implementation runs asynchronously in several independent threads. The lowest-level thread reads the measurements from the sensors and fills up circular buffers for data processing. The main control thread uses the processed data and implements the state machine controlling the behaviour of the orthosis. It runs with a 10 Hz frequency. To illustrate the control mechanism, see the pseudo-code for 'Override' mode listed in Figure 9.

The velocity of the orthosis motion in active mode is calculated as limited linear function of the current delta force variable value. Maximum velocity, number of exercise cycles, limiting long-term delta force and other constant parameters are set and stored to enable simple repetition of the exercise for a given patient. The key parameters are listed in Table 2.

```
repeat
   GetSensorData
   UpdateBuffers
   CalculateDeltaForce(state)
   switch state
      case active:
         if DeltaForceLongTerm < Preset
            state = override
         else if Angle > limits
            state = hold
         else
            SetVelocity(DeltaForce)
         endIf
      endCase
      case override:
         if (DeltaForce > limits) OR (Angle > limits)
            state = hold
         endIf
      endCase
      …
   endSwitch
until CycleNumber = MaxCyclesCount
```

Figure 9. Active mode control pseudocode

Parameter	*Range, default value*
Initial motion angular range	-5°-150°, default found during calibration
Exercise max. speed	10 – 60°/sec, 20°/sec
Max cycles count	1 – inf., 10
Hold duration	0 – 1 min, 5 sec
Max negative delta force	1 – 40 N, 20 N
Long-term inactivity limit	0.1 – 40 N, 1 N

Table 2. Exercise key parameters; the names are self-explanatory

The high-level control is implemented as a stand-alone application coded using the C# programming language with the Microsoft Visual Studio development tool. The application contains a graphical component showing the

current position of the orthosis to enable visual checks of the zero-angle position prior to the calibration/exercise. Exercise parameters as well as the exercise logs are stored in XML files.

3.3 Voice output

During the exercise with a physiotherapist, the patient is instructed vocally. It is desirable to mimic this feature in the AEO as well; therefore, a voice output is implemented. Vocal instructions allow the patient to focus on the exercise in a more natural way compared to on-screen visual hints and commands only.

Based on our previous practical experience with interactive mobile robots communicating with people, the voice is pre-recorded and not synthesized. A natural human voice is still more pleasant to hear and patients are more inclined to follow the commands. To avoid monotonous and uniform repeated sound, which can be annoying for a longer exercise, each sound is recorded in several takes with slightly modified wording and pronunciation. The particular sound to be played is selected randomly.

There are five groups of voice output messages:

1. Instructions. Typically before the calibration or the exercise – a short description of what is going to happen and what the patient should expect.
2. Commands. Whenever the patient is expected to do something, e.g., start to pull the orthosis in the flexion stage of the cycle.
3. Announcements. Typically during a state change, e.g., when the orthosis changes its mode from active to override, or from override to hold. Optionally, the number of cycles performed or the number of cycles still to come can be announced.
4. Encouragements. Some encouragements are mixed with the announcements, e.g., when half or two thirds of the desired angular range is achieved. Other encouragements are used when the patient, e.g., starts to fade during the exercise and the angular range achieved in the current cycle is narrower than the previous one.
5. Rewards. When progress in the exercise is made, e.g., the patient's activity in the current cycle is higher compared to the previous one, or the angular range has been improved.

4. Verification

The AEO performance was preliminarily verified on a group of patients in cooperation with the University Hospital Olomouc. The group of five patients consisted of two males and three females, age range 21–38 years. The orthosis was applied to the right arm elbow for three patients and to the left arm elbow for two patients. The initial angular range prior to the exercise was typically around 25°–85°.

The patients were first briefed regarding the purpose of the exercise, the functionality of the AEO, and the procedure of orthosis calibration. Because the calibration is essential for proper functionality, the patients initially carried out the calibration sequence using their healthy arm. We found this to be the best way to introduce the patient to the orthosis. When comfortable, the calibration procedure was repeatedly performed on their injured arm. Results of the calibration were briefly analysed to find out whether the true free arm motion was reached and the calibration results were repeatable.

Following the calibration, a set of exercises was performed, first in the active mode and then in the override mode. Initially, the exercises consisted of four cycles only, with the number of cycles gradually increasing. The final exercise had 10 cycles, which for most of the patients was the limit before they became tired and needed a break. A patient with the AEO attached is shown in Figure 10.

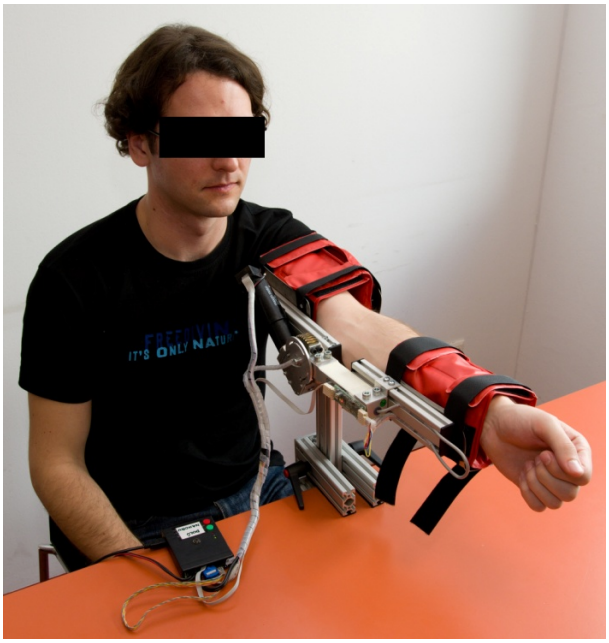

Figure 10. Patient with AEO attached, prior to the exercise

A course of delta force and the orthosis angle during the exercise for patient number one are shown in Figures 11 and 12. The patient's elbow showed stiffness in extended position only. The first five cycles of the override exercise are shown, with the lower safety limit set to -1°. It can be seen that during the initial two cycles the override mode was activated, while starting from the third cycle the patient had reached the final position by themselves already in the active mode regime.

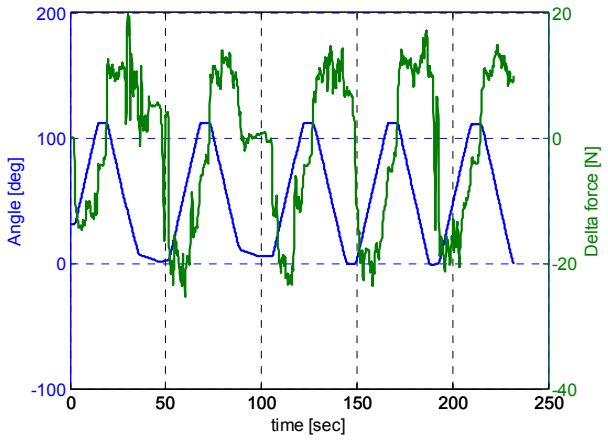

Figure 11. Time course of orthosis angle and delta force performed by patient #1 during the exercise

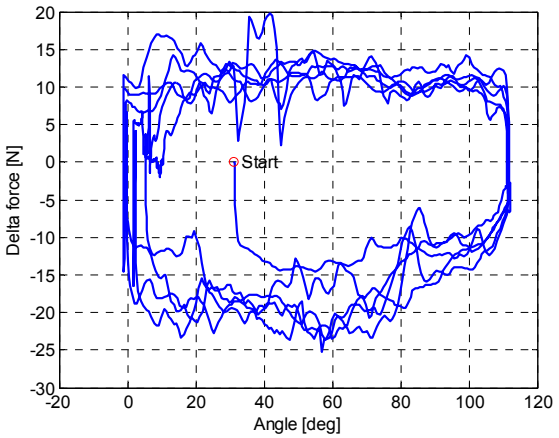

Figure 12. Delta force course depending on orthosis angle, patient #1

The same courses for another patient are shown in Figures 13 and 14. This patient had elbow stiffness in both extension and flexion positions. The graphs show the first five cycles in active mode (with no override).

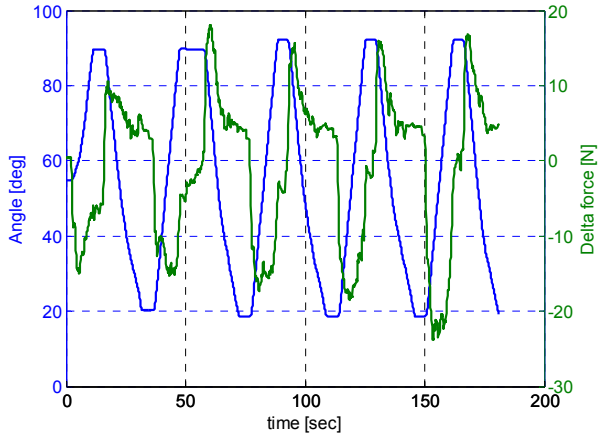

Figure 13. Time course of orthosis angle and delta force performed by patient #2 during the exercise

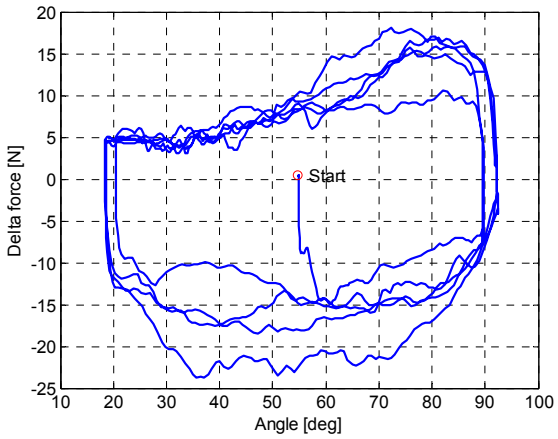

Figure 14. Delta force course depending on orthosis angle, patient #2

It can be seen that the angular range is slowly extending during the exercise. It is notable that the extension part of the cycle is much more difficult for the patient, as the delta forces for angles lower than 50° are quickly dropping to low values, compared to the flexion stage of the cycle.

One of the parameters that exhibit the performance of the orthosis and can be easily quantified is the change in angular range. With all the patients tested, the angular range has increased at the end of the exercise, when compared with the initial range. Particular changes are listed in Table 3.

Patient #	*Range change (RC) [°]*	*RC [%]*
1	(5, 111) → (-1, 111)	5.6
2	(25, 85) → (17, 96)	31.6
3	(16, 93) → (11, 97)	11.7
4	(23, 81) → (20, 89)	18.9
5	(19, 88) → (14, 94)	15.9
Average	11.2	16.7

Table 3. Angular range change during the exercise. Patient #1 had elbow stiffness in extension stage only; the upper limit is therefore constant. Patient #2 participated in two consecutive exercise sessions; the presented results are aggregate.

5. Discussion

The angular range change represents an objective yet single valued parameter expressing the improvement of the joint stiffness. The logs recorded during the exercise can be viewed and processed to give a more detailed view on how the patient performed. The course of delta force depending on the elbow angle tells us what portion of the motion is difficult to achieve. The time course of sum/average delta force in each cycle may indicate when the patient becomes tired. The advantage of the AEO lies within the objective measurement of the patient's effort, allowing the medical personnel to evaluate even the

progress between exercise sessions based on valid data.
The only issue that was not solved to our satisfaction was the necessity to repeat the calibration procedure during the exercise. The control of the AEO is solely based on measured patient activity; this activity is related to free arm calibration curves, while the curves are valid only in the angular range performed during the calibration. The angular range, however, extends during the exercise, and the control is then in limit positions related to extrapolated parts of the calibration curves beyond original angular motion limits of the patient's elbow. Therefore, it is necessary to repeat the calibration process when the angular range extension exceeds a certain value. We have found that with extension greater than 5° in the upper or lower angular position, the recalibration is necessary to keep the sensitivity of the control in a practical range.

6. Conclusions and Future Work

The presented device implements a novel approach to post-traumatic treatment rehabilitation. The requirements for the device listed in the introduction section were all met. Following a short training session patients can perform the rehabilitation by themselves, at home. The question whether this can replace the physiotherapist completely will have to be answered by a regular medical study on a larger number of patients.

Apart from further experiments with patients, future work will be focused on further post-processing of the data, aiming to develop a method to automatically detect a patient becoming tired as opposed to a patient with poor compliance. These two states are generally hard to distinguish, even for an experienced physiotherapist.

As the patient's arm is not fixed too rigidly to the AEO, the patients sometimes try to overcome the pain in limit angles by changing their posture, usually moving the shoulder. With the physiotherapist, the patients are often simply warned not to change the posture. Detection of posture change using a computer vision system is another topic to consider in future work.

7. Acknowledgements

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