Human-Robot-Interaction Control for Orthoses with Pneumatic Soft-Actuators – Concept and Initial Trails

David Baiden and Oleg Ivlev
FWBI Friedrich-Wilhelm-Bessel-Institute Research and
IAT Institute of Automation,
University of Bremen, Germany
baiden@fwbi-bremen.de

***Abstract*—The purpose of this paper is to present a concept of human-robot-interaction control for robots with compliant pneumatic soft-actuators which are directly attached to the human body. Backdrivability of this type of actuators is beneficial for comfort and safety and they are well suitable to design rehabilitation robots for training of activities of daily living (ADL). The concept is verified with an application example of sit-to-stand tasks taking conventional treatment in neurology as reference. The focus is on stroke patients with a target group suffering from hemiplegia and paralysis in one half of the body. A 2 DOF exoskeleton robot was used as testbed to implement the control concept for supporting rising based on a master-slave position control such that movements from the fit leg are transferred to the affected leg. Furthermore the wearer of the robot has the possibility to adjust support for stabilizing the knee joint manually. Preliminary results are presented.**

I. INTRODUCTION

HUMAN-ROBOT-INTERACTION (HRI) is an important issue using robotic systems for training of activities of daily living (ADL) or movement therapy in neurorehabilitation after stroke and similar ailments. Especially when this kind of robots are attached to human limbs in order to transfer motions to paralyzed arms or legs, robot behavior is crucial with respect to comfort as well as safety and dependability.

There are several possibilities to improve safety and increase human involvement as defined by not bounding the person to a fixed reference trajectory. Two general approaches are to add more compliance to the device or to adapt the reference movement i.e. the trajectory to the individual movements of a person. The extent of interaction depends on the purpose of rehabilitation. In the immediate post stroke phase more guidance is necessary, while after a certain progress of therapy when the person is able to generate own effort, less guidance and more freedom i.e. minimized interaction forces are desirable. Different methods can also be combined and until now some sophisticated rehabilitation robots have been developed which provide robot assisted therapy according to the approach of “assist-as-needed”, “patient-cooperative“ or “subject-driven” [1], [2]. The idea behind this concepts is that assistance is sufficient to guide and complete desired physiological movements while challenge demands patients to provide maximal own effort [3], [4]. In that way neuroplasticity is stimulated and motor learning is regained.

Impedance control in combination with electrical drives allows deviations from a reference trajectory [5]. Riener *et al.* implemented this approach for the gait trainer Lokomat and achieved an improvement of HRI [2]. The authors remarked that this control strategy seems to be suitable for hemiplegic patients because the emphasis is more on guidance than on full release. For another gait trainer LOPES Virtual Model Control was developed where the reference is adapted to the patient’s behavior [6], with Complementary Limb Motion Estimation desired motions for the affected leg are estimated depending on the physiological inter-joint couplings in the unaffected leg [7]. For the robotic suit HAL different control strategies have been presented which are selected depending on the treatment purpose and related to the capabilities of the wearer [8]. The Cybernic Voluntary Control adjusts the robots joint torques for assistance depending on the measured muscle activities by myoelectricity. The Cybernic Autonomous Control algorithm estimates motion intention based on acceleration and ground contact forces for reproducing a stored movement pattern including gravity influences and balance control. Chugo *et al.* presented a mobile robotic walker system to assist elderly persons in sit-to-stand transfer, walking and seating [9]. The device is equipped with a lifting mechanism and a support pad as arm rest. Depending on the body posture a position controller moves the person from sitting to standing condition by adjusting the height of the support pad. Another approach of an assistive sit-to-stand device for paraplegic patients is presented by Jović *et al.* in [10]. Voluntary trunk motion is detected by acceleration sensors in order to trigger a pre-programmed pattern for assisting the sit-to-stand movement by functional electrical stimulation (FES).

Most of these robotic systems focus on gait training, operate with complex control schemes or require a large number of sensors. In this study a lower extremities exoskeleton robot is used as a testbed to analyze HRI control strategies. Initial results of the first concept for supporting the sit-to-stand transfer with a simple controller are presented. Further approaches are being developed.

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II. Concept of the Sit-to-Stand Trainer

The target group are hemiplegia patients who are paralyzed in one side of the body but still are able to move their limbs regularly on the unaffected body part and are not restricted in hip movements. An active orthosis to stabilize and move the leg and a support frame for balance are the main elements of the sit-to-stand trainer (see Fig. 1). The control algorithm was developed by using a wearable 2 degrees of freedom (DOF) lower extremities exoskeleton robot as testbed.

Since compliant actuators are beneficial for comfort and safety [11], recently developed soft-actuators are applied. They are of direct rotary type and belong to the class of antagonistically arranged pneumatic muscles. The similar operation principle of a previous generation of actuators with rotary elastic chambers (REC-actuators) has already been explained in [12]. One important characteristic property of the actuators is high power/weight ratio compared to electrical drives which makes them well suitable for HRI, a successful integration in an assistive acting movement therapy device for knee rehabilitation is presented in [13]. According to the conceptual overview (see Fig. 2) the motion controller calculates the desired trajectory related to human-robot-interaction and specifies necessary torques $\boldsymbol{\tau}_d$ for the subsequent soft-actuator subsystem. Generated torques $\boldsymbol{\tau}_a$ are induced by the pneumatic soft-actuators to the human who weares an active orthosis on the affected leg. On the fit leg a passive orthosis is attached.

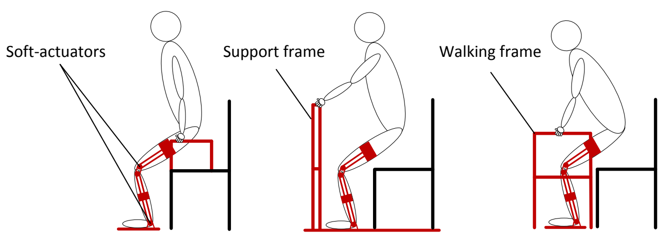

Fig. 1. Concepts of the sit-to-stand trainer. An active orthosis with pneumatic soft-actuators is attached to the impaired leg of a hemiplegia patient who grasps a support frame for balance while performing repetitive rising tasks. The application of different handles and arm rests need to be analyzed.

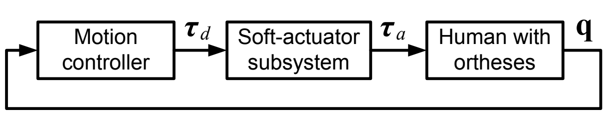

Fig. 2. Simplified control concept containing trajectory calculation by the motion controller, torque/pressure control in the soft-actuator subsystem and the human wearing one active and one passive orthosis.

III. Control Concept

Obviously conventional stiff position controlled robotic devices which reproduce pre-programmed trajectories are not suitable for true HRI, because any effort is countered by the robot. Adding compliance to the actuators only by means of elastic elements or an impedance controller with constant parameters would neither lead to the desired objective. High compliance results in a phase shift and reduced amplitudes so that the desired range of motion (ROM) is not reached, while low compliance again leads to poor interaction potential. However an impedance controller can often be used as fundament combined with adjustments in position or torque by higher level controllers. Due to natural compliancy based on air compressibility and elastic chambers the position controlled soft-actuators have passive compliance and behave similar to an impedance controller with stiff actuators which is considered as active compliance. The three key factors of the proposed concept are individual reference trajectory generation, general compliance and involvement of the human by offering to adjust support manually resulting in an increased extent of HRI without limitation of a fixed reference.

A. Conventional Therapy

For conventional treatment in neurology one therapist supports the patient for balance while another one stabilizes the knee joint. Only persons with severe paralysis need stabilization in hip joint also. Repetitive sit-to-stand training is performed starting from sitting on the edge of the bed with individually adjusted height. Advanced persons are able to walk a few steps afterwards using a regular walker. According to medical partners the hemiplegia patients of the target group are able to perform sit-to-stand tasks, when they grasp a handle to support balance and the affected leg is stabilized.

B. Trajectory Generation

Often it is criticized, that pre-recorded movements from a fit person are used as reference movements for therapy robots which are attached to an impaired patient. Another common used method is to calculate mean values from a group of different persons who should represent the average of the population. Even references obtained from the patient by a teach-in procedure with manual therapist support might not be convenient, because movements change over time depending on the treatment progress. All this reference signals often result in non-fitting and not comfortable movements for the impaired patient who has an individual movement pattern. Therefore the movements are adapted individually by the trajectory generator inside the motion controller (see Fig. 2).

C. Soft-Actuator Subsystem

The soft-actuator subsystem consists of an open loop torque controller and a model-based pressure controller [14]. The pneumatic chambers generate a torque that is a non-linear function of chamber pressure and rotation angle

$$\tau = f\left(p, q\right). \tag{1}$$

All actuators are assembled with an antagonistically arranged pair of two identical chambers and are represented by the same model. The total torque τ_j with joint index $j = 1, 2$ calculates as difference

$$\tau_j = \tau_{jp}\left(p_{jp}, q_{jp}\right) - \tau_{jn}\left(p_{jn}, q_{jn}\right) \tag{2}$$

with the single chamber torques τ_{jp} for positive and τ_{jn} for negative turning direction, the relating chamber pressures p_{jp} and p_{jn} and the chamber rotation angles q_{jp} and q_{jn}. Using inverted actuator models from (1), each desired pressure is a function of desired joint torque and current joint angle $\mathbf{p}_{dj} = \mathbf{f^{-1}}\left(\boldsymbol{\tau}_{dj}, \mathbf{q}_j\right)$. For simplification the single chamber rotation angles are substituted by $\mathbf{q}_j$. Initial stiffness on pressure level (not on spring-like torque level) is adjusted by a torque offset τ_{j0} which is converted to initial pressure so that both chambers exert the same torque against each other resulting in zero-torque. The arising increased norm pressure is advantageous for stability of the pressure control loop.

D. Master-Slave Position Control

This HRI control concept is based on a master-slave (MS) structure with compliant position control. In this approach the reference movement is sensed from the sound leg and transfered to the impaired leg so that the wearer generates the reference with his individual movement pattern by himself. The input for the joint position controllers is the difference

$$\boldsymbol{\Delta}\mathbf{q} = \mathbf{q}_{ma} - \mathbf{q}_{sl}$$

between posture of master $\mathbf{q}_{ma}$ and slave $\mathbf{q}_{sl}$. According to Fig. 4 then the position controller output is the desired torque $\boldsymbol{\tau}_d\left(\boldsymbol{\Delta}\mathbf{q}\right)$ for the subsequent open loop torque/pressure control subsystem. All torques are estimated based on the actuator models and measurements of pressure and angle sensors only, which allows to exclude costly torque sensors. The controller will automatically generate enough torque to stabilize the affected leg without the need of any model of robot or human.

In another approach presented in [13], a model-based gravity compensation based on separated models of the robot and the human's upper extremity was applied. There the total desired torque is the sum of torques for compensating gravity and assistive torques for inducing movement.

Different from that for sit-to-stand movements both legs always move in parallel with small differences only and the error depending resulting desired torque $\boldsymbol{\tau}_d\left(\boldsymbol{\Delta}\mathbf{q}\right)$ therefore remains within a small adjustment range that is advantageous for dynamic response. Thus the effect of the MS-concept here is similar to the influence of a feed-forward model which linearizes the plant. The control structure comprises of a PID controller for the ankle joint to provide more guidance and to keep track of the reference, while for the knee joint higher compliance is desired and thus a P controller is used (see Fig. 3). Since the torque generation is based on the position error, so far the resulting effect is that the slave will follow the master with a certain phase shift depending on the selected stiffness parameters i.e. the generated torque will reduce for $\boldsymbol{\Delta}\mathbf{q} \to \mathbf{0}$. However this MS-concept has the benefit that the controller will always generate enough torque to stabilize the affected leg without the need of any model of robot or human and represents the fundament for next studies.

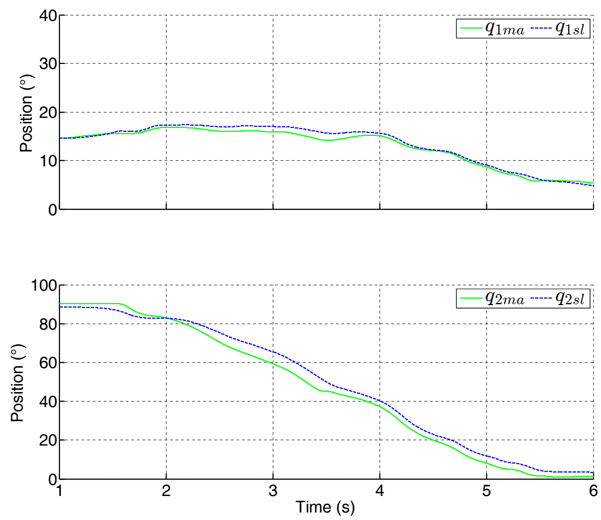

Fig. 3. Sit-to-stand transfer with MS-position control using a PID controller for the ankle joint (upper graph) and a P controller for the knee joint (lower graph). The affected leg q_{jsl} follows the fit leg q_{jma} with similar individual posture.

E. Involvement of Human

The position controller described in the previous section allows a person to perform sit-to-stand movements by guiding the weak leg equivalent to the posture of the unimpaired leg. But for severely affected persons this might not be enough support so that the knee torque needs to be augmented. The sit-to-stand process mainly requires a change of center of gravity and sufficient torque of the knee joint. According to medical partners the patients of the target group are not restricted in hip movements so that they are able to change their emphasis and pull themselves upwards using the support frame while additional torque for the affected knee joint is generated by the robot.

In this control concept the person is allowed to influence the control loop by adjusting stabilizing torques manually. Thus involvement of the human should be increased in order to facilitate the feeling of "being in control", away from the "robot-driven" experience of Continuous Passive Motion (CPM) devices. The human should be involved as much as possible in the process because this is assumed to increase motivation for voluntary participation. Only if support is manually requested by pulling a lever, which is mounted on the support frame and the affected leg did not yet reach fully extend posture, torque is generated. Stabilizing torque is generated by modifying the reference trajectory in order to create a virtual error with $\boldsymbol{\Delta}\tilde{\mathbf{q}} = \boldsymbol{\Delta}\mathbf{q} + \varepsilon_{usr}$ where ε_{usr} is the influence of the person (see Fig. 4). The trajectory generator increases the power of the knee joint gently without creating instability. The control law reads as follows:

$$\begin{aligned}\tilde{\boldsymbol{\tau}}_d\left(\boldsymbol{\Delta}\tilde{\mathbf{q}}\right) &= \begin{bmatrix} \tilde{\tau}_{d1} & \tilde{\tau}_{d2} \end{bmatrix}^T \\ &= \begin{bmatrix} K_{p1} \cdot \left(\frac{1}{T_N}\int \triangle\tilde{q}_1 \, \mathsf{dt} + T_V \cdot \frac{\mathsf{d}}{\mathsf{dt}}\triangle\tilde{q}_1\right) \\ K_{p2} \cdot \triangle\tilde{q}_2 \end{bmatrix} \end{aligned} \quad (3)$$

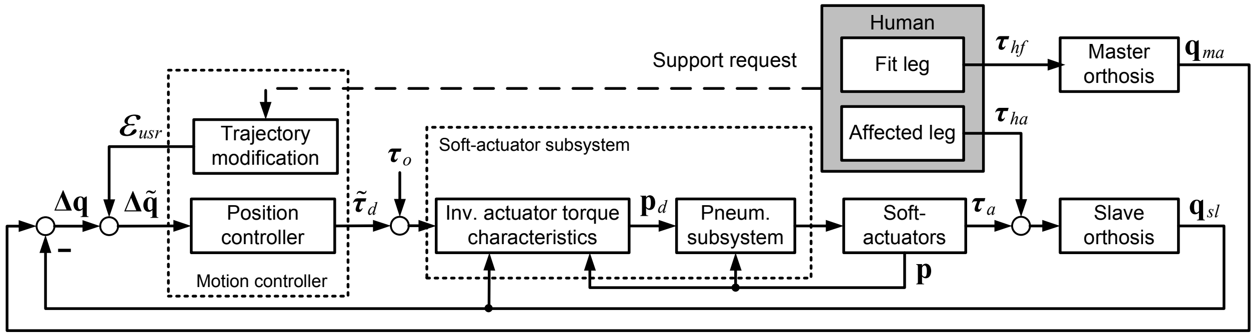

Fig. 4. Master-slave control structure with possibility of manual torque adjustment. During sit-to-stand movement the human is allowed to request further support resulting in increased torque for stabilizing the knee joint.

$$\Delta\tilde{\mathbf{q}} = \begin{cases} \Delta\mathbf{q} + \boldsymbol{\varepsilon}_{usr} & \text{for } q_{2sl} \geq q_{2limit} \wedge s_{2usr} > 0 \\ \Delta\mathbf{q} & \text{else} \end{cases} \tag{4}$$

$$\boldsymbol{\varepsilon}_{usr} = \begin{bmatrix} 0 & s_{2usr} & \cdot k_2 \end{bmatrix}^T$$

with s_{2usr} as signal given by the person due to lever activation and proportional gain k_2 for amplifying the support request. q_{2limit} is the upper limit for maximal desired knee extension which can be individually adapted. Further torque is applied only to the knee joint because the ankle joint does not have a wide contribution to lift the human body. The final concept aims for providing adaptive support as well as "human-in-the-loop" control for sit-to-stand and stand-to-sit training. Initially presented experiments were conducted with able-bodied subjects and should provide a proof of concept only. Further studies concerning application to hemiplegia patients with restricted standing abilities need to be conducted.

IV. Results

A. Experimental Setup

The experimental setup comprises of two leg orthoses which are attached to the lower extremities of a subject (Fig. 5). Each orthosis represents a 2 DOF planar robot which can be independently controlled and is equipped with soft-actuators in knee and ankle joints. The knee joint is constructed of a series connection of two actuators to achieve a full ROM from 0° (extension) to 90° (flexion). In the ankle joint a single actuator provides movements from 0° (extension) to 45° (flexion). Joint angles are measured with magnetic sensors (AMS PRAS 21). For safety reasons each joint is mechanically limited. The pneumatic system consists of a set of four highly dynamic proportional solenoid valves (Festo MPYE-5-1/8LF-010-B), whereof two valves are needed for each joint. The master orthosis remains passive and serves as sensing device for measuring the human joint angles and only the slave orthosis is pneumatically connected in order to transfer movements form the robot to the subject's other leg. Master and slave orthoses can be interchanged.

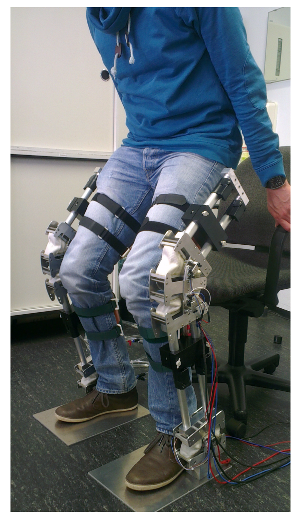

Fig. 5. Exoskeleton robot as testbed for analyzing human robot interaction strategies for lower extremities movements equipped with recently developed soft-actuators.

B. Sit-to-stand support

Initial experiments have been conducted with several able-bodied subjects who tried to perform sit-to-stand tasks by simulating one weak leg. The body weights of the subjects was 87.5 ± 12.5 kg with body heights between 184.5 ± 5.5 cm. In the first experiment the subject was asked to perform sit-to-stand movement followed by a stand-to-sit maneuver. Support was triggered right from the start in sitting position in order to facilitate the initial lift which is quite exhausting with one leg only. Fig. 6 illustrates the conducted experiment with four different sections of the task: sitting for $t \in [0 \ldots 1.1 \text{ s})$, sit-to-stand for $t \in [1.1 \ldots 3.7 \text{ s})$, standing for $t \in [3.7 \ldots 4.8 \text{ s})$ and stand-to-sit in the interval $t \in [4.8 \ldots 8 \text{ s}]$. In the topmost plot angles of the ankle joint are shown. Due to the PID controller the position error is kept small while the master q_{1ma} is always in lead before the slave q_{1sl}. Plot 6 b) shows the desired torque $\tilde{\tau}_{1d}$ for the ankle joint which has a relative small adjustment range due to nearly parallel movement of both legs. In Plot 6 c) the chamber pressures of the single

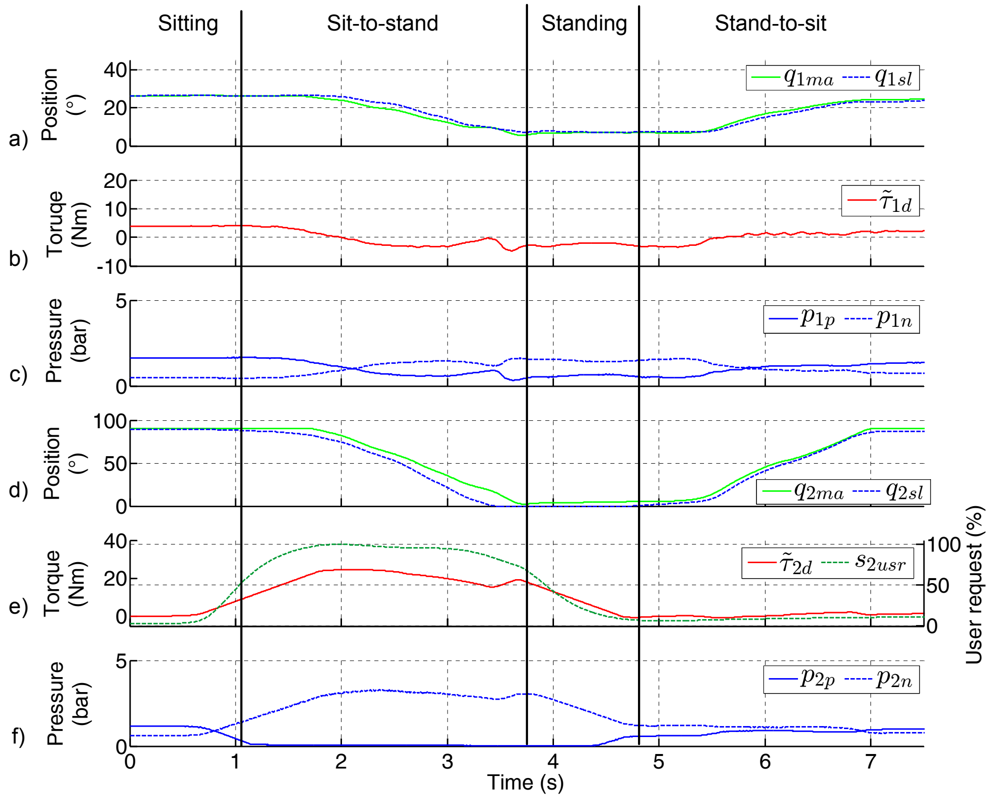

Fig. 6. Controller behavior during sit-to-stand and stand-to-sit task: the topmost plot illustrates the course of ankle angle from the master q_{1ma} (solid line) and the slave q_{1sl} (dashed line) using a PID controller with according desired torque $\tilde{\tau}_{1d}$ in the 2nd plot and chamber pressures p_{1p} and p_{2n} in the 3rd plot for the first joint. Following three plots c) - f) below represent the course of the knee joint using a P controller with related torque and pressures. The subject triggers support before sit-to-stand movement so that the knee joint is augmented right from the start and the subject is able to rise.

actuator for the ankle joint are shown. Since flexion is related to positive joint turning direction, the course of p_{1p} is similar to the desired torque $\tilde{\tau}_{1d}$ in the previous plot. Values for $\tilde{\tau}_{1d} \approx 0$ after $t = 6$ s result in similar pressures in both chambers. This are the pressures which result from the initial stiffness τ_{10} in the soft-actuator subsystem. The last three plots show the same physical quantities again for the knee joint with the series actuator. In Plot 6 e) the dashed signal is the support request triggered by the subject and scaled to the maximum value during this trial. Torque $\tilde{\tau}_{2d}$ is increased due to virtual trajectory modification until stable standing posture with for $q_{2sl} < q_{2limit}$ is achieved at $t = 3.7$ s. For the knee joint negative turning direction is related to knee extension so that the courses of $\tilde{\tau}_{2d}$ and pressure p_{2n} are similar. Note that due to additional stabilizing torque the slave i.e. the weak joint q_{2sl} in Fig. 6 d) is more guided and leads before the fit joint q_{2ma}.

C. Support in case of weakness

In a second experiment the subject was ask to rise and trigger support only in case of weakness. It can be seen from Fig. 7 that the subject gets stuck at $t = 2.3$ s and is reliant on help. Desired torque increases slowly, indicated by related pressure enhancement. At first the master q_{2ma} leads before the slave q_{2sl} until additional stabilizing torque augments the knee and brings the slave into lead and the subject is able to complete the movement.

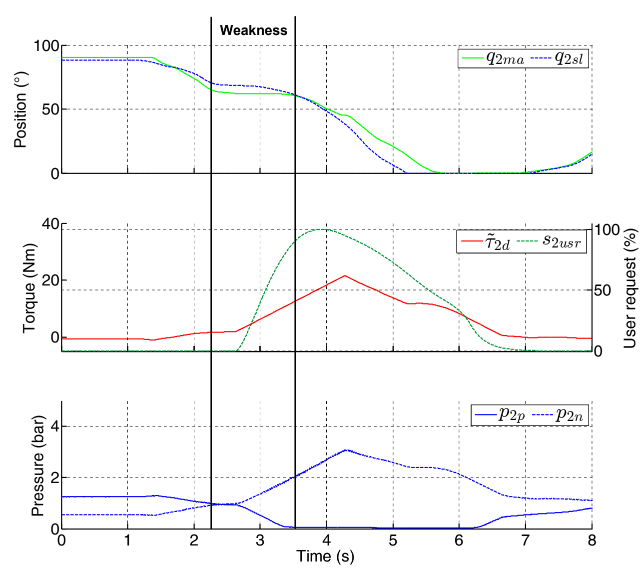

Fig. 7. Support in case of weakness. Subject got stuck at $t = 2.3s$ and triggers support to complete the sit-to-stand task.

V. Conclusion and Future Work

In this paper a concept for HRI control for an active lower extremities orthosis with pneumatic actuators is presented. The implemented controller is based on a master slave position control for ADL training using the example of sit-to-stand tasks. The target group are hemiplegia patients who are paralyzed on one half of the body but still are able to balance themselves using a support frame. The controller transfers the posture of the fit leg to the affected leg, while additional supportive torque can be triggered by the person in case of weakness. The advantage of this concept is that no fixed reference trajectories are necessary, individual reference is generated by the person himself and compliance is assured by utilizing soft-actuators. Furthermore no model of robot or human is necessary due to always similar posture of both legs resulting in small position errors and subsequent small controller adjustment ranges.

Experiments with able-bodied subjects show a proof of concept, subjects were able to interact with the robot and participated in the movement. In the next step voluntary effort should be detected or estimated depending on trunk movement in order to adjust support appropriately only in case of weakness in terms of load sharing. Further research should also include controllable stiffness for the pneumatic soft-actuators in order to achieve a spring-like behavior with independent stiffness-position or stiffnes-torque control.

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