Design and Control of the Lower Limb Exoskeleton KIT-EXO-1

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Abstract— **We present a new lower limb exoskeleton with series elastic actuators for augmentation of human performance and for rehabilitation of the musculoskeletal system. The KIT-EXO-1 consists of 3 DOF, linear series elastic actuators with progressive helical springs which change the spring stiffness over spring deflection and a force-based interface to the human body. We describe the design actuator and its sensor system. In addition, we describe a force sensor system for capturing interaction forces between the exoskeleton and the human body. We present a force-based control approach, which allows the generation of motion pattern based on interaction force pattern between the exoskeleton and the human body. Our first experimental results demonstrate the performance of the new linear series elastic actuator and provide a proof of concept for the force-based control approach.**

I. INTRODUCTION

In recent years, extensive research efforts have been dedicated to the area of exoskeletons for augmenting human performance in daily and working environments, assisting the motion of people with physical limitations by providing robotic-centered technologies and methodologies in the area of rehabilitation. Stationary systems like the Lokomat [1], the LOPES [2] or the ALEX [3] are used in hospitals or rehabilitation centers to assist the rehabilitation training. They provide highly repetitive training and adjust the assistance to the patients capabilities while walking on a treadmill. However these systems are not portable and can not be used at home or for activities of daily living. Systems like the EKSO [4], the Indego [5] or the Rewalk [6] are commercially available exoskeletons which should enable SCI Patients to walk again. All of them have two actuated Degrees of Freedom (DOF) in the hip and knee joints and require the patient to use crutches. Other robotic devices like the HAL ([7], [8]) can be used for medical applications as well as for private use to augment single joints or the whole leg. The Bleex [9] and IHMC Mobility Assisit [10] exoskeletons focus on augmentation of the users movement. The mindwalker exoskelton [11] has five DOF per leg, where four of them are active, and every joint is equipped with elastic elements [12] realized by double disc shaped springs. EKSO, Indego and ReWalk use predefined trajectories to control the device, which can be adapted by the trainer while HAL uses surface electromyograms (sEMG) signals as input information for the controller and relates the signals to joint torques. In Mindwalker control approaches based on Brain Neural Computer Interface (BNCI) as well as EMG are investigated and implemented.

Fig. 1. The KIT exoskeleton consists of two active DOFs in the knee and the ankle. CAD design (left) and first prototype (right)

The interaction forces between the exoskeleton and the human body are of utmost importance for the realization of physical Human Robot Interface (pHRI) of exoskeletons. Such forces build the basis for the control approaches of the Muscle Stiffness Sensor [13] and the Walking Power Assist leg (WPAL) ([14], [15]). Another important design parameters of exoskeletons is the the Range of Motions (ROMs) of the individual degrees of freedom.

In this paper, we present a new exoskeleton which can be used for augmentation of human performance as well as for rehabilitation of musculoskeletal system, the KIT-EXO-1 (see Fig. 1). The exoskeleton is driven by Series Elastic Actuator (SEA) with progressive helical springs and is described in Section II. In Section III we describe our force-based control approach which only make use of interaction forces between the exoskeleton and the human body. The evaluation of the new actuator and exoskeleton are presented in Section IV. Section V concludes the paper and discuss our future work.

II. DESIGN OF THE EXOSKELETON

The first exoskeleton design serves as proof of concept for our approach in designing linear elastic actuators with progress springs and using only interaction forces for control. The goal was a lower limb exoskeleton which allows a healthy subject to walk and perform squat motions. In the following we describe the design of exoskeleton's frame and the actuator, the mechanical coupling of the exoskeleton to the users leg and finally the hardware architecture.

A. Design of Frame and Structure

The objective in the leg construction was to build the exoskeleton variable enough to mount different actuators (elastic/nonelastic) on the knee or ankle joint. Three basic frame parts for thigh, shank and foot are connected by orthothic joints (Otto Bock, 17B47=20 / 17B57=20) and allow mounting the actuators (see Fig. 1). Choosing soft aluminum (EN-AW 5083) permits bending the frame by hand as well as using water cutting technology which results in low production costs of 50 EURO for all three frame parts. Although the aluminum is soft, the frame is rigid enough to carry a body weight of 75 kg while providing three DOF, namely knee flexion/extension, ankle plantarflexion/dorsiflexion and ankle pronation/supination. The pronation and supination motion in the ankle is realized by a passive revolute joint placed under the user's heel. Using plain bearings, the overall height of the heel structure is reduced to 28 mm and thus it is possible to embed the joint in the heel of a sports shoe, with the remaining shoe sole damping the rotation in pronation/supination direction. With the actual frame an angle of 7° is possible in both directions. Mechanical stops of the orthotic joints limit the dorsiflexion/plantarflexion to 25° / 45° in the ankle and to 0 ° knee extension. Limited by the length and mounting position of the actuator posterior of the leg the maximum knee flexion is 65° which should allow walking with moderate speed. The assembled frame has a weight of 1.1kg including all straps, cables, motor controllers and the sports shoe.

B. Series Elastic Actuator

As shown in [16] the energy consumption of a series elastic actuator (SEA) driving an exoskeleton joint depends on the spring stiffness used in the actuator. A higher body weight which leads to higher joint torques demands for a stiffer spring, while higher velocities demand softer springs. Taking this into account we designed a linear SEA consisting of a satellite roller screw linked to driving unit with brushless DC-motor (Parker K044, max. 400W) and an optional elastic element. An incremental encoder is keeping the commutation of the 6 pole rotor. Fig. 2 shows the structure and different elements of the linear SEA.

The chosen roller screw (Rollvis RV 7x1) has a lead of 1 mm and can be loaded with a nominal force of 10.9 kN. Efficiency of the transmission system is 0.86. A cross-shaped roller bearing (THK 2008) simultaneously captures the thrust loads of the roller screw and keeps the motor radially in position. If more force is applied on the SEA the springs get stiffer. The elastic unit positioned at the end of the driving unit incorporates therefore two progressive helical springs which are connected to the driving unit via a flange positioned in the middle of them. The position of the flange

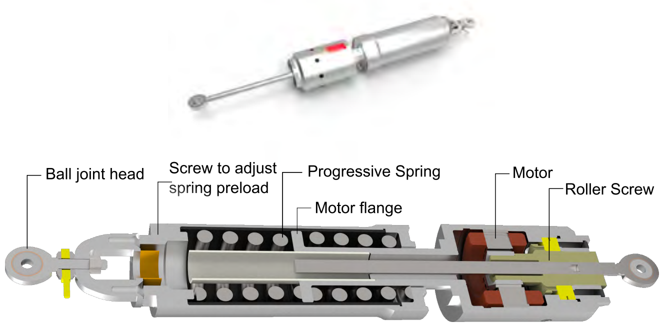

Fig. 2. The new linear series elastic actuator. The linear SEA is used in the knee joint while a non-elatisc version is used in the ankle joint of the first exoskeleton prototype.

allows the cushioning of traction and compression forces. Pre-loading the springs to half deflection during assembly prohibits the springs of loosing contact to the flange while loaded. This adjustment can be done offline (exoskeleton is shut down) by tighten a screw at the end of the elastic unit. The maximum deflection of each spring is 21 mm. Due to the spring arrangement, the total spring stiffness changes from 375 N/mm to 435 N/mm when deflecting the springs as shown in Fig. 3. Motor and springs were chosen to support full body weight of a 100kg (120 Nm maximum joint torque) person while walking at a speed of 1 m/s.

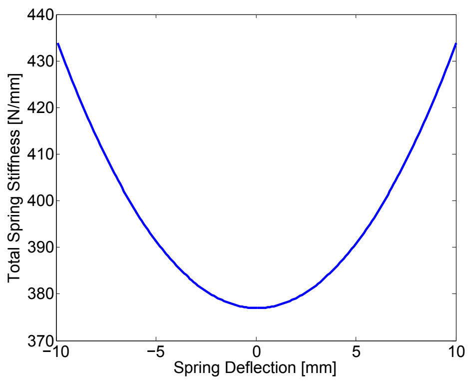

Fig. 3. Spring stiffness of the SEA (theoretical values)

Ball joints connect the driving unit to the frame so that it is applicable for various human joints. Based on this design, we also developed a second version of the actuators without the elastic unit. This version is used in the ankle joint of the first exoskeleton prototype. The weight of the SEA is 1.38 kg in which the weight of the elastic unit is 800 g.

Both actuators are controlled by ELMO Whistle motor controllers (ELMO Motion Control GmbH) which are equipped with in-house developed boards, that allow the connection of motion and force sensors and the communication with a master PC. Angular position of the knee and ankle pitch joint is measured with a resolution of 4096 counts/rev and the link sided motor position with 5760 counts/rev. Comparing link and motor sided positions allows to measure spring forces with a resolution of 18 N.

C. Sensor System of the Exoskeleton

To measure interaction forces between exoskeleton and user as well as to determine muscular contraction, force sensing resistors (FSR) are used. These sensors have a flat surface and small size which allow their integration in narrow spaces. Taking account of the cost efficient design these sensors are available for about 10 to 20 EUR. As recommended by the sensor manufacturer (IEE S.A., Luxembourg) an interface was developed providing a flat and rigid mounting surface made by a rapid prototyping technique. For a reliable coupling of forces onto the active sensing area of the FSR, a thin elastomeric layer is used. This layer fits on the active area of the sensor and avoids inhomogeneities by distributing and smoothing the pressure to the overall surface though ensuring a correct force measurement.

To achieve a fast response of the exoskeleton to the user's movement, the muscular contraction of specific thigh muscles is observed. By measuring the level of muscular activity it is possible to implement a force augmentation. As shown in [17], a value of muscular contraction (of thigh muscles) can be obtained by measuring a force signal perpendicular to the muscle. This signal is produced 0.3 s prior the movement referred to the muscular activity. Further, [18] shows that the musculus rectus femoris (and the musculus quadriceps in its entirety) is heavily involved in pulling the limb up against gravity and generally acts as knee extensor. According to [19], [20] the hamstrings muscles (among these the musculus biceps femoris) at the thigh and the musculus tripes surae at the shank are important muscles for the knee flexion. Placing sensors at shank muscles is difficult regarding the exoskeletons kinematics. Misalignments in the knee joint between human and exoskeleton can result in deviating knee angles and the loss of contact at the calf.

The sensor setup is illustrated in Fig.6. Placing the FSRs on top of the muscle bellies of the m. rectus femoris (S1) and m. biceps femoris (S2) emphasized as sufficient to identifiy the user's movement. Figure 4 shows the sensor signals of FSRs mounted on a cuff interface aligned to the desired muscle bellies.

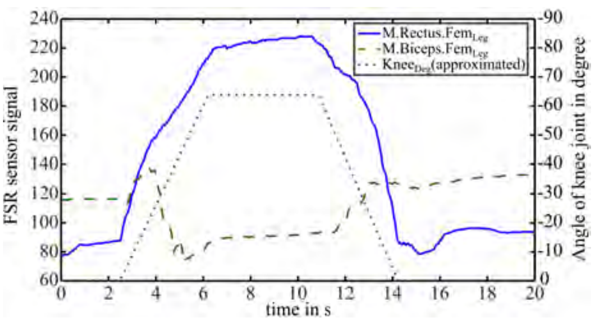

Fig. 4. Signal sequence of two FSRs mounted on a cuff interface and aligned to the bellies of m. rectus femoris and m. bieps femoris. This measurement is done without the exoskeleton and thus contains no quota of interaction force.

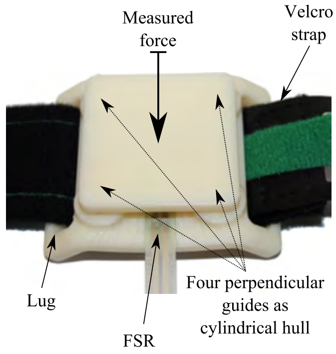

Fig. 5. Design of the muscle sensor interface providing lugs for attachment to the Velcro straps.

During free swing motions the muscle activation is low.Therefore FSRs are mounted on the top end of the exoskeleton frame (S3) and at the achilles tendon (S4). The signals represent mainly the interaction force and facilitates free movements of the user in cooperation with the exoskeleton. Last two sensors (S5 and S6) at the bottom of the shoe underneath the heel detect contact events with the ground.

To avoid shear movements to the sensors and to amplify the force acting on the sensor, appropriate interfaces were developed. These interfaces transmit only forces perpendicular to the active area of the sensors by guiding this movement. The interfaces used for the sensors at the thigh muscles provide additional lugs to attach them easily to the Velcro straps as shown in Fig. 5. The 40×40 mm^2 sized contact surface is reduced on the rear side to get only in contact with the active area of the FSR.

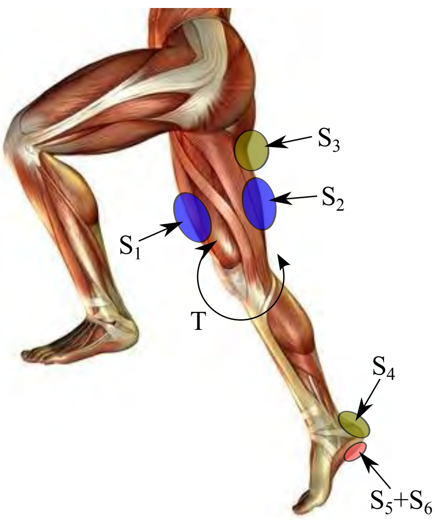

Fig. 6. Setup of the sensor system; S1-S6: FSR Sensors measuring the interaction forces between user and exoskeleton. Figure adopted from [21]

Sensor data is read by an Arduino Nano and send to the PC for further processing. The first prototype is equipped with a serial connection (RS 232) which allows a control frequency of 40-50 Hz. Such frequency does not allow a successful implementation of torque control. Thus the master PC sends velocity commands to the motor controller which generates the needed current (torque) commands internally with a frequency of 1000 Hz.

D. Attachment of the Exoskeleton to the Body

The attachment of the exoskeleton to the human body is central element of the physical human robot interface. Such interface must ensure the mechanical coupling between the human leg and the exoskeleton while taking into account the comfort of the user. To realize such interface velcro straps form a knee orthosis are used to connect the exoskeleton to the users leg on thigh and shank (see Fig. 5). The foot is secured by wearing the sports shoe, attached to the frame as shown in Fig. 1. Additionally padding with adaptable positions to the users preferences guarantees a comfortable fit in the aluminum frame.

III. Exoskeleton Control

The purpose of the exoskeleton is to assist the motion of a user in his/her daily activities in private and/or working environments or during rehabilitation process. At this point the exoskeleton provides two powered hinge joints in the sagittal plane, which represent the human joints at knee and ankle. These joints have to be moved via the above mentioned self-locking linear actuators and controlled over a master PC.

Instead of using movement information as a control scheme, the interaction forces acting between exoskeleton and user's body are used as input of the controller to make predictions of the user's motion. In general the prediction of the users knee motion is performed by measuring the summed forces contributing to the knee joints with the FSR sensors and control the position of the linear actuator in a way which result in reducing the interaction forces. The usage of force sensors for the exoskeletons is motivated by several reasons such as reliability, practicability, convenience to wear, etc. compared to EMG or EEG based sensor systems. These also come along with some disadvantages like a long calibration process and sensor noise in changing conditions as e.g. reported in [22] and [23]. FSRs disadvantage is the changing sensor signal for different pretensions of the straps demanding for offset measurement when donning the exoskeleton.

A. Control approach

The force signals processed by an Arduino are smoothed, multiplied with gains taking account the size of the sensor interface as well as the position of the sensor and summed up to define a criterion for the effective interaction force. According to [17], changing muscle contraction can be measured 0.3 s prior the intended movement. In combination with a filter it is additionally important to have a high sensing and control frequency to ensure an early movement recognition.

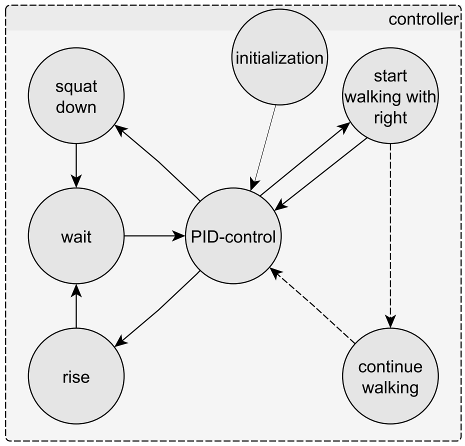

Fig. 7. Finite state machine to select the predicted user movement

Equation (1) describes the weighting of the sensor signals for the calculation of the control variable $F_{control}$. Weights were first determined by the size of the observed muscles and then adapted experimentally.

$$\begin{aligned} F_{control}(t) &= (S_1(t) - O_1) - 0,4 \cdot (S_2(t) - O_2) \\ &\quad -0,5 \cdot (S_3(t) - O_3) \\ &\quad -1 \cdot (S_4(t) - O_4) \end{aligned} \tag{1}$$

$F_{control}$ is composed of the interaction forces acting between the exoskeleton and the user's body which are measured by the FSR sensors. To consider the rise of the sensor signal occurring by tightening the Velcro straps, the measured signals are decreased by offsets $O_i, i = 1, \ldots, 4$. These four parameters are determined with an initial measurement (about 2 s) of the signals while standing in an upright resting position after each new donning of the exoskeleton or tightening up of the Velcro straps.

Fig 8 illustrates the functioning of the control approach. The filtered and computed control variable is derivated once and filtered again to sustain a smoothed input for the derivative term of a PID controller. The FIR-filter used for the smoothing of the derivative values is a weighted moving average filter of 9^{th} order. According to [24] such low

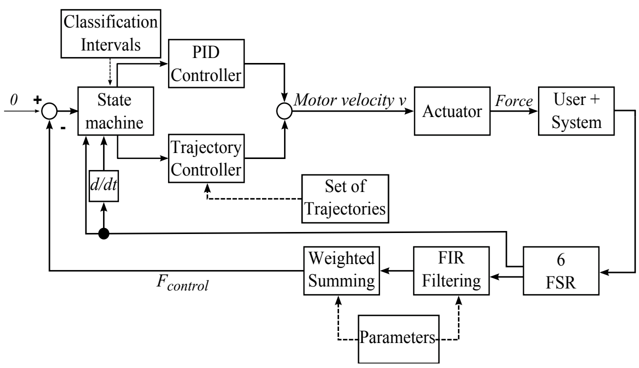

Fig. 8. Block diagram of the proposed control approach

pass-filtering helps avoiding unwanted high derivative values through high frequencies of the input. Beneath the derivative term the control variable is also integrated to receive the integral term of the PID controller. The proportional term is provided by the weighted interaction force itself.

Before reaching the controller a finite state machine (Fig.7) compares the control signal to trigger intervals to select the mode of operation. If no predefined motion is detected the PID controller is used and the exoskeleton shadows the leg movement of the user without exerting forces on it. Force signals corresponding to predefined motions cause the state machine to switch in a mode where the motor controller is commanded by predefined trajectories. In this mode the exoskeleton guides the user's leg. These trigger intervals are defined in advance for squat down, rise from squatted position and initialize walking with the right leg (the exoskeleton mounted on left leg) as shown in table I. $S_1, \cdots, S_6$ represent the sensors on the leg. The corresponding values in column 2 - 4 are raw sensor data derived by capturing one healthy subject being advised to initiate the desired movements. To differentiate from various free movements, the subject explicitly used a fast initiation movement. All conditions have to be satisfied to detect an event.

When triggering an action but the according state of the state machine (Fig.7) is selected and predefined trajectories provide input data for the actuator as mentioned before. When the movement is finished the state changes back to PID control (rise, squat down) or in case of walking transits to a subsequent state (continue walking). The trajectories are derived from the KIT Whole-Body Motion Database [25] and are captured with a Vicon motion capture system [26]. They could be adapted for any given case e.g. less knee angle while walking after knee surgery or new trajectories could be provided by motion capturing systems and integrated into the trigger evaluation. While the trajectory is running the PID controller signal is kept for continuous monitoring purposes.

The output of the PID controller and the trajectory generator are velocity values used for controlling the motion of the linear actuator. Reaching the angular limit is detected by the absolute encoder measurements.

IV. RESULTS

A. Actuators

To evaluate the capabilities of the actuators an existing test bed was adapted to measure the maximum force and the maximum velocity, as well as the maximum force at nominal speed. The actuator is capable to generate a maximum axial Force of 2900 N and a force of 900 N at the nominal speed of 100 mm/s which corresponds to the stroke length of the linear actuator. This maximum force results in a knee torque of 116 Nm which is sufficient to lift a user of 100 kg. Maximum speed of the actuator at no load is 260 mm/s. Force bandwidth was measured with both mounting points locked and is 17 Hz at 900 N and 5 Hz at 2600 N.

TABLE I

SIGNAL AND SIGNAL DERIVATIVES AS TRIGGER VALUES FOR MOVEMENT RECOGNITION.

Trigger	Start Walking	Squat Down	Rise
knee angle	$[-1, 7°]$	$< 7°$	$> 60°$
$\frac{dS_1}{dt}$	$[-300, -20]$	$[-190, -30]$	$[25, 225]$
$\frac{dS_2}{dt}$	$[-160, -5]$	$[-220, -20]$	$[20, 160]$
$\frac{dS_3}{dt}$	$[-60, 85]$	$[10, 380]$	$[-120, 360]$
$\frac{dS_4}{dt}$	$[20, 315]$		
$\frac{dS_5}{dt}$	$[-11., 135]$	$[-3.4, 19]$	$[-460, 300]$
$\frac{dS_6}{dt}$	$[-3.2, -100]$	$[-1.400, 6]$	$[-150, -360]$
S_5	> 2.5		> 0
S_6	> 800		> 0

B. Control

The control approach was tested with one healthy person on which the exoskeleton was calibrated (subject 1) and additionally three untrained healthy subjects that never used the exoskeleton before (subjects 2–4). The trigger values remained untouched for all subjects to examine their generality. For characterizing the amount of force augmentation neither sEMG measurements as in [17] nor trials with recorded and compared metabolic rates as in [27] are used yet. Hence a comparison of the sensor signals sequence measured at the bellies of *m. rectus femoris* and *m. biceps femoris* on the actuated and the unactuated leg is used to estimate the augmentation capabilities and is shown in Fig. 9. For squatting down the user has no need to apply muscular effort because of the systems damping due to the restricted motor velocity. During the rising movement the interaction force at the *m. rectus femoris* is increasing while the muscle contraction signal of the unactuated one is decreasing. This suggests that the muscular effort of the actuated leg is reduced in the early phases of rising. However additional muscular effort needs to be applied at small flexion angles due to the surpass of the actuated signal compared to the unactuated one.

Nevertheless while an active trajectory state is taken, the muscular effort of the user is reduced. Therefore, the reliability of the found trigger intervals are tested in the active controlled system on the four subjects. While the user on which the system was calibrated had a high success rate detecting squat down (95%) and rise (100%), success rate for rise was very low for the untrained subjects (max. 10%) and the recognition of the squad down motion succeeded in about half of the trials (max. 83%). Two of the three untrained subjects were able to start walking with a low success rate.

V. CONCLUSIONS AND FUTURE WORK

We presented the first prototype of the KIT-EXO-1, a lower limb exoskeleton with three DOF for augmentation and rehabilitation tasks, in which the knee joint is actuated by a new linear series elastic actuator. In addition, we present a force-based interface between the exoskeleton and the human

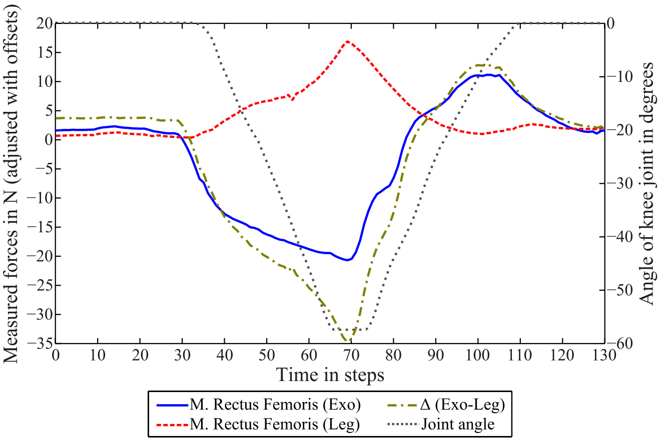

Fig. 9. Signals from the sensors aligned at the *m. rectus femoris* and the *m. biceps femoris* both of the exoskeleton and the free unactuated leg.

body and a first control schema combined with a finite state machine which allows the prediction of movement of the user.

Our first tests with the exoskeleton showed that the mechanical structure and the actuators are capable providing enough support for rehabilitation and augmentation tasks. Using a force-based control approach for the knee joint showed the potential in controlling an exoskeleton joint without angular information. In the future, we will investigate how the control approach can be applied to higher bus frequencies and how it could be combined with pattern recognition methods to improve the prediction of human motion. Future work will be devoted to the the transfer of the control approach to the ankle joint. To evaluate the actuator performance, extensive tests with different healthy subjects will be performed.

Moving the actuators for knee and ankle to a lateral position allows sitting and will be considered in the next prototype as well as the ability to align the exoskeleton to the users leg. An advanced control strategy for the SEA can reduce the energy consumption in the exoskeleton joints and should lead to a more fluent movement. Therefore, a simulation model of the leg and the SEA will be created and evaluated using reference human motion data captured with Vicon system.

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