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AN ADAPTIVE IMPEDANCE CONTROL BASED ON EMG – DRIVEN MODEL FOR ROBOT – AIDED REHABILITATION

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Abstract. *This paper deals with adaptive impedance control of rehabilitation robotic systems based on an EMG-driven patient's activity model. The patient's torque and stiffness are estimated using a simplified and optimized model of muscle forces based on electromyographic (EMG) signals. The proposed adaptive control strategy considers these patient's activity measures to compute the stiffness parameter of the robot's impedance control, such that the patient participation on the therapy is always encouraged. Experimental results of trials with healthy subjects are presented, considering the EMG signals of selected muscles acting at the joint knee during flexion and extension movements using a variable impedance knee orthosis.*

Keywords: *rehabilitation robotics; impedance control; electromyography; adaptive control*

1. INTRODUCTION

Since the 1970s, exoskeletons (robotic devices coupled to the human body) have been used to give support in gait and execute forces beyond the user's capacities. Potential applications for these devices range from military units to increase physical capacity of soldiers, support factory workers while carry heavy loads, rehabilitations devices, up to support devices for people with physical disabilities or elderly people (Dollar and Herr, 2008).

In the medical field, there are two categories of exoskeletons for lower limbs: assistive devices and rehabilitation devices. The main goal of assistive exoskeletons is to give total support to patients who lost the possibility to recover movement. On the other hand, rehabilitation exoskeletons are designed for patients who suffer some type of motor impairment and seek to promote motor cortical reorganization to improve pattern movements.

The initiative to introduce robotic devices in rehabilitation of people with motor impairment arises as a technological solution aimed at improving traditional recovery therapies, focusing to quantify the functional recovery of patients, and improve therapist efficiency.

The most successful robot-administered therapy to aid neuro-recovery is based on several principles of learning. The use of robotic devices aimed at rehabilitation technics (Lotze, Braun et al. 2003, Kaelin-Lang, Sawaki et al. 2005) show that voluntary motion execution by the patient increase motion memory codification. As well, motion assistance to achieve a target (which was not possible to achieve without assistance) stimulates peripheral cortical areas by activating natural mechanisms of motor learning, providing safety in motion and generating a positive psychological encouragement when patient reach the target (Harkema, 2001). Hogan (2014) adds to this concept the use of visual aids indicating targets to be achieved with the aim to stimulate the intention of motion, and highlights that the use of exoskeletons for rehabilitations increases the number of repetitions to be performed by the patient at each therapy session. After all, different types of sensors coupled to the robotic device can be used to monitor performance and physiological parameters as well as therapy advances.

Consequently, the robotic device in constant contact with the user's body must have intelligence and flexibility to develop the tasks coming from the user, providing the necessary strength and resistance. Therefore, while user has motor skills, cooperation between them is designed in such way that the user is in control of movements. Thus, the control system of the robotic device is designed to be driven by biological signals coming from the nervous system of the patient, captured in the muscles, named electromyography (EMG) signals.

Considering this fact and based on the knowledge that muscle's mechanical impedance plays an important role in control of posture and motion, even under external disturbances, this paper takes into account the impedance control developed by Hogan (1984). The impedance control allows that dynamic interaction between two physical systems can be modulated, regulated and controlled, allowing a compliant behavior.

In this paper it is proposed an adaptive impedance control of rehabilitation robotic systems based on an EMG-driven patient's activity model. The patient's torque and stiffness are estimated using a simplified and optimized model of muscle forces based on electromyographic (EMG) signals. The proposed adaptive control strategy considers these patient's activity measures to compute the stiffness parameter of the robot's impedance control, such that the patient participation on the therapy is always encouraged. Experimental results of trials with healthy subjects are presented, considering the EMG signals of selected muscles acting at the joint knee during flexion and extension movements using a variable impedance knee orthosis.

The paper is structured as follows: Section 2 presents the active knee orthosis, the EMG-based torque and stiffness estimation, and the proposed adaptive control strategy; Section 3 presents the experimental results of the evaluation of the estimation approach and the adaptive strategy; and Section 4 presents the conclusions.

2. METHODS

This section presents the exoskeleton (an active knee orthosis) used to evaluate the EMG-based adaptive control strategy as well as the mathematical formulation of the proposed approach.

2.1 Active knee orthosis

The robotic device used to evaluate the adaptive strategy proposed is an active knee orthosis, shown in Fig. 1, designed and developed as part of the Exo-Kanguera project of the Rehabilitation Robotics Laboratory of the Engineering School of São Carlos – USP (dos Santos and Siqueira, 2014). It is actuated by a rotary series elastic actuator (SEA).

The mechanical design was conceived in order to obtain a compact and lightweight architecture. All housing parts were made of aluminum for the purpose of reduced weight. The final assembly of the rotary SEA consists of Maxon Motor RE 40, graphite brushes, 150 Watt DC motor; worm gear set (M1-150 of HPC Gears International Ltd.) with reduction ratio of 150:1; customized torsion spring; angular contact bearings; magneto-resistant incremental encoder, and opto-electronic incremental encoder. The resulting mass is 2.53 kg, allowing direct mounting of the actuator on the frame of a knee orthosis.

a)

b)

Figure 1. Knee orthosis. a) Rotary series elastic actuator; b) rSEA attached to a knee orthosis

The rotary SEA control system consists of two EPOS *positioner controller* connected to a CAN board build – in PXIe 1071 platform through National Instruments PXI 8512 board. EPOS are used to read the actuator encoder output (EPOS 24/5) and motor control (EPOS 70/10). Communication interface is performed by a CANOpen communication protocol with a transmission rate set at 500Kbit/sec. The control system has two levels connected in cascade, PXI as master, operated at 200Hz and the two EPOS as slave system. Finally, impedance control strategy is implemented, where desired stiffness and damping values for the knee joint can be set (Eq. (9) in the next section represents the result of the impedance control).

2.2 Computing torque from EMG

It is assumed that muscle activation signals are generated from EMG signals with a dimensionless values ranging between zero and one. First, the raw EMG signal measuring the muscle's electrical activation must be processed. From raw EMG signal, $e(t)$, the moving average of one hundred samples, $\overline{e}(t)$, is subtracted to eliminate offsets and movement artifacts. Then, the corrected signal is fully rectified, low-pass filtered and normalized to obtain the neural activation, $u(t)$. The filter corresponds to a second order Butterworth filter with a cut-off frequency of 2Hz. According to Potvin et al. (1996) and Lloyd and Besier (2003) muscle activation variations with relation to neural activations show an exponential relationship:

$$a(u) = \frac{e^{AuR^{-1}} - 1}{e^{A} - 1}, \tag{1}$$

where R is the value of maximum contraction isometric force, and A is a non-linear shape factor defining the curvature of the function, bound to an interval of $-3 < A < 0$.

Once obtained the muscle activation, is necessary to compute the resulting force. To do this, a muscle model is used. Normally, the muscle behavior is represented by Hill –type muscle model (Hill, 1938). The muscle is modeled as a contractile element producing the active force through contraction and a parallel elastic element that produces a passive force when muscle is stretch:

$$f^{M} = f_0^{M}\left[\left(af^{L}\left(\tilde{l}^{M}\right)f^{V}\left(\tilde{v}^{M}\right) + f^{PE}\left(\tilde{l}^{M}\right)\right)\right]\cos(\phi), \tag{2}$$

where, f_0^M is the maximum force at optimal fiber length, with $f^{L}\left(\tilde{l}^{M}\right)$, $f^{V}\left(\tilde{v}^{M}\right)$, $f^{PE}\left(\tilde{l}^{M}\right)$ corresponding to normalized active force–length, normalized force–velocity and normalized passive force–length curves (Millard et al., 2013) . Lastly, ϕ is the angle between the line force of the tendon and the muscle fibers, called as pennation angle, given as

$$\phi = \sin^{-1}\left(\frac{l_0^M \sin\phi_0}{l^M}\right), \tag{3}$$

where l^M is the muscle fiber length and ϕ_0 the pennation angle at optimal fiber length l_0^M .

To compute the torque contribution of the muscle, the moment arm, r , as a function of the joint angle, θ , is calculated according to the tendon displacement method proposed by An et al. (1984) :

$$r_i = \frac{\partial l^{MT}}{\partial \theta} . \tag{4}$$

Consequently, once calculated the forces and moment arms for all muscles acting above the joint, the torque is expressed by:

$$\tau = \left|\sum_{i=1}^{n} r_i f_i^M\right|_{agonist} - \left|\sum_{j=1}^{m} r_j f_j^M\right|_{antagonist} . \tag{5}$$

Note that agonist and antagonist muscles act in opposite way to generate the join torque. The stiffness above the joint is given by the action of all acting muscles forces. When forces increase proportionally, no movement is done, but the stiffness increase.

A joint stiffness could be estimated considering a stiffness coefficient:

$$CR = \left|\sum_{i=1}^{n} \tau_i\right|_{agonist} + \left|\sum_{j=1}^{m} \tau_j\right|_{antagonist} . \tag{6}$$

Finally, the joint stiffness can be estimated as (Karavas et al., 2013):

$$k = \alpha CR + \beta \tag{7}$$

where α (*rad*$^{-1}$) and β (*Nm/rad*) are constant parameters.

2.3 Adpative control strategy

In this section, the proposed adaptive strategy to defined the robot's stiffness is presented. First, it is assumed that exoskeleton and user are trying to perform the same task. In this case, the dynamical equation that defines the system is given by:

$$\tau_{user} + \tau_{exo} = M(\theta)\ddot{\theta} + C(\theta,\dot{\theta}) + G(\theta), \tag{8}$$

where $M(\theta)$ is the inertial matrix, $C(\theta,\dot{\theta})$ is the vector of Coriolis and centripetal forces and $G(\theta)$ the vector of gravitational forces corresponding to the coupled dynamic system. We also assume that the cooperation task between exoskeleton and the user needs a known torque to be executed, as presented in (Ibarra et al., 2014). This torque will be the sum of the torque contributions of each participant in the task:

$$\tau_{nec} = \tau_{user} + \tau_{exo} \ . \tag{9}$$

As mentioned in Section 2.1, the robot is controlled by an impedance control law. It is given by:

$$\tau_{exo} = k_{exo}\theta_e - B_{exo}\dot{\theta} \tag{10}$$

where $\theta_e = \theta_d - \theta$ is position error given by the difference between desired and measured position, $\dot{\theta}$ the angular velocity, k_{exo} and B_{exo} are the stiffness and damping defined by the designer. Following the same structure, we assume the user perform a similar impedance control strategy with:

$$\tau_{user} = k_{user}\theta_e - B_{user}\dot{\theta}, \tag{11}$$

where and k_{user} and B_{user} are the user's stiffness and damping, respectively.

Replacing (10) and (11) in (9), one obtains:

$$\begin{aligned} \tau_{nec} &= \left(k_{user}\theta_e - B_{user}\dot{\theta}\right) + \left(k_{exo}\theta_e - B_{exo}\dot{\theta}\right) \\ \tau_{nec} &= \left(k_{user} + k_{exo}\right)\theta_e - \left(B_{user} + B_{exo}\right)\dot{\theta} \end{aligned} \tag{12}$$

Considering steady state conditions, the term of damping, directly affected by velocity will be null, then:

$$\tau_{nec} = \left(k_{user} + k_{exo}\right)\theta_e \ . \tag{13}$$

Consequently, robot stiffness could be calculated as:

$$k_{exo} = \frac{\tau_{nec}}{\theta_e} - k_{user} \ . \tag{14}$$

Since stiffness is by definition the ratio between force and position:

$$k_{exo} = k_{nec} - k_{user} \ . \tag{15}$$

From eq.(15) it is possible to observe that when user does not participate of the task, i.e. $k_{user} \approx 0$, the robot assume the highest stiffness to assist the whole movement. To avoid abrupt changes in robot stiffness, it is included an adaptation factor $0 < f < 1$, to limit the effective stiffness of the exoskeleton, then, for a given tine instant *i*:

$$k_{exo}(i) = f k_{exo}\left(i-1\right) + \left(1-f\right)\left[k_{nec} - k_{user}\left(i\right)\right] \ . \tag{16}$$

Thus, for small values of *f* , variation among previous and actual stiffness is considerable.

3. RESULTS

In this section, results obtained from two experimental setups are presented. The experiments were conducted by a healthy male subject (33 year old, height: 187 cm, and weight: 73kg, without known neurological or biomechanical knee disorder) wearing the active knee orthosis and performing knee flexion and extension movements. The subject gave prior written consent for voluntary participation. This study was approved by the Ethics Committee of the Federal University of São Carlos.

Figure 2 shows the experimental setup, including the EMG Trigno system (Delsys Inc.), the NI PXIe 1071 platform (National Instruments), the EPOS Positioning System (Maxon Motors), and the active knee orthosis. The Trigno system, with wireless sensors, is used to acquire EMG data at a sample rate of 2kHz. Initially, it is measured the Maximal Voluntary Contraction (MVC) of each muscle. For this, subject was asked to activate their muscles to the maximum level and maintain it for 5 seconds, while keeping his leg constrained. Measurements were acquired 3 times with a rest time of 30 s between them for flexion and extension muscles, separately.

The OpenSim software (Delp et al., 2007) was used to scale a generic musculoskeletal geometry model to match the subject's anthropometry. Once the model is scaled the following parameter are extracted to estimate the patient torque: Muscle – tendon length; Tendon length;Normalized muscle length;Pennation angle; Moment arm; Active and passive force curves.

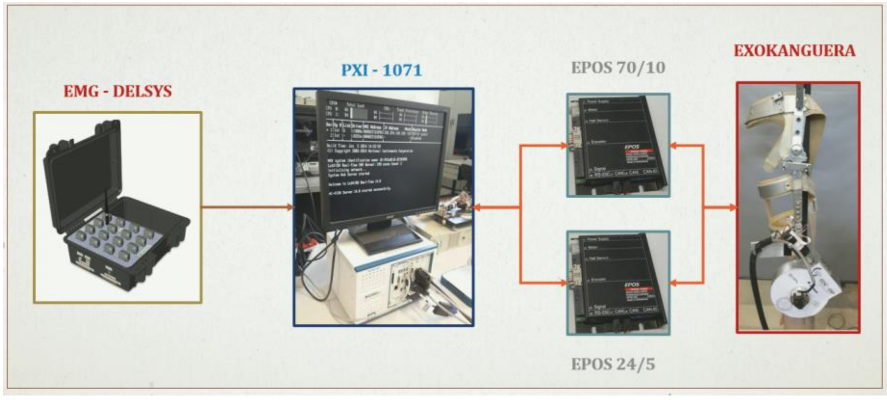


Figure 2. Experimental setup.

3.1 Stiffness estimation

In the first experiment, the subject is asked to perform knee flexion and extension movements while seated. EMG signals of three extensor muscles (*rectus femoris* (RF), *vastus medialis* (VM), *vastus lateralis* (VL)) and two flexor muscles (*biceps femoris* (BF), *semitendinosus* (ST)) are used to estimate the human torque and the joint stiffness. The active knee orthosis is set to follow a desired sinusoidal trajectory (0.2 Hz) with three different degrees of stiffness (k_{exo} = 0 Nm/rad, 30 Nm/rad, and 60 Nm/rad, where k_{exo} represents the robot's stiffness).

First, the subject is asked to actively follow the desired trajectory (a metronome is used) for the three different degrees of robot's stiffness. Then, the subject is asked to relax the leg and let the robot does the whole work with stiffness set to 30 Nm/rad and 60 Nm/rad (with k_{exo} = 0 Nm/rad and subject in passive condition there is no movement).

Figures 3 to 5 show the joint trajectories (θ) for twelve flexion and extension movements (here, total extension occurs at θ = 90º), the subject's torque (τ_{user}) and stiffness (k_{user}) estimates, the robot's torques (τ_{exo}), and the torque estimates for each muscle for the five conditions regarding robot's stiffness and subject participation (active or passive). Table 1 shows the RMS (root mean square) values for all signals. It is noteworthy that the subject's stiffness decreases with the increasing of the robot's stiffness, keeping the same amount of applied torque.

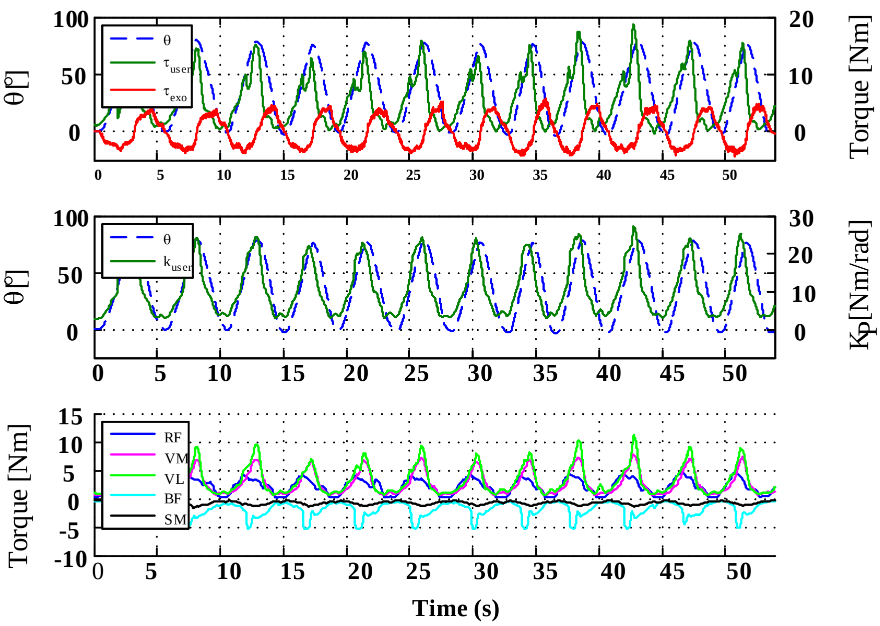
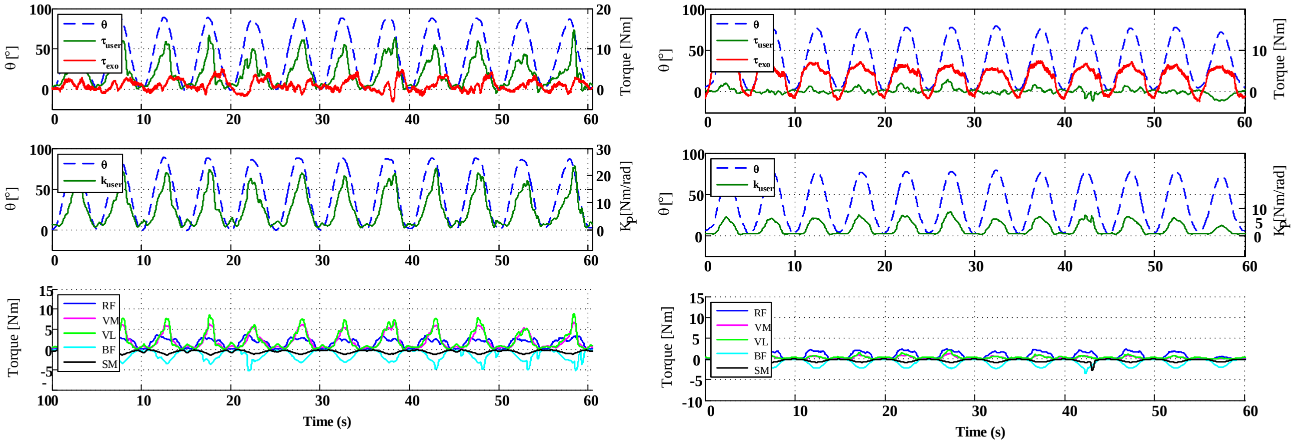
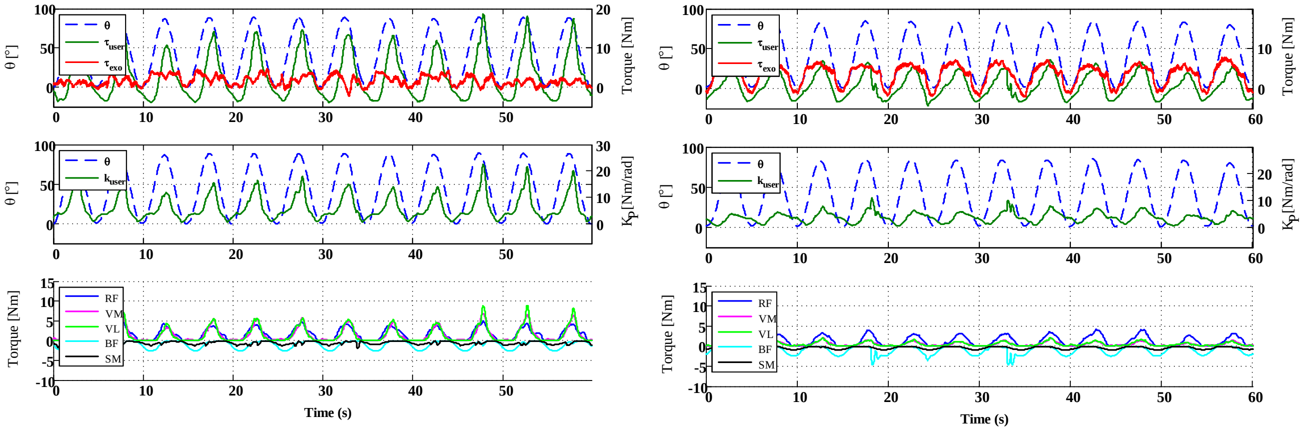


Figure 3. Estimated stiffness and torques. Active subject and k_{exo} = 0 Nm/rad



a) b)

Figure 4. Estimated stiffness and torques for k_{exo} = 30 Nm/rad. a) active subject; b) passive subject.



a) b)

Figure 5. Estimated stiffness and torques for k_{exo} = 60 Nm/rad: (a) active subject; (b) passive subject.

Table 1. RMS values.

Robot's stiffness (k_{exo}) [Nm/rad]	Subject's stiffness (k_{user}) [Nm/rad]		Subject's torque (τ_{user}) [Nm]		Robot's torque (τ_{exo}) [Nm]	
	Active	Passive	Active	Passive	Active	Passive
0	12.57	-	6.76	-	2.69	-
30	9.98	3.50	5.31	0.73	1.62	4.45
60	7.77	4.08	6.41	2.97	2.05	4.39

3.2 Adaptive Control Strategy

To evaluate the proposed adaptive strategy for robot's stiffness, in the second experiment the subject is asked to keep the knee flexed at θ = 90º (different from previous figures, total extension occurs at θ = 0º) while in stand posture. For simplicity, EMGs signals of only one flexor muscle, (*biceps femoris)* and one extensor muscle (*rectus femoris)* are measured to sense activity during knee flexion and extension movements. The set point for the active knee orthosis is set to 90°. The necessary stiffness (k_{nec}) is defined as the minimum stiffness to take the subject's leg (without collaboration of him) from 0° to 10% of the set point (81°). For this experiment, k_{nec} is set as 30 Nm/rad.

Figure 6 shows the set point position (θ^d), the measured position (θ), the necessary stiffness to maintain the set point position (k_{nec}) and the user's stiffness (k_{user}) estimated by (6), and the exoskeleton's stiffness (k_{exo}) calculated by (15) and (16). Note that the when the subject increases the stiffness, the exoskeleton and user system reaches the set point with a decreasing of the robot's stiffness as expected.

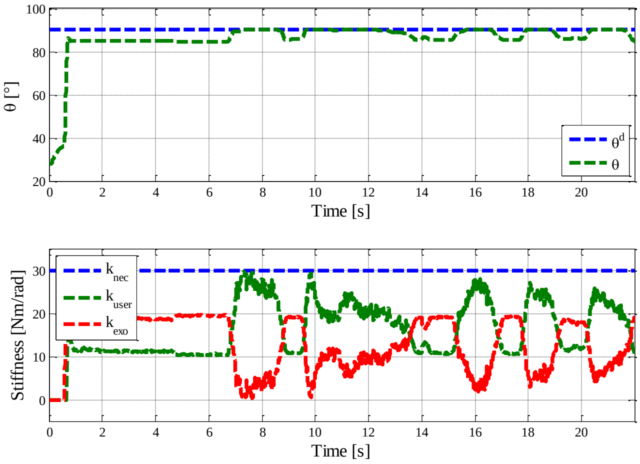


Figure 6 – Stiffness behavior to maintain the set point position during right leg flexion.

4. CONCLUSIONS

In this paper it is proposed an EMG-based adaptive control strategy for robot-aided. The user's torque and stiffness are estimated using a simplified model of muscle torques. The proposed adaptive control strategy considers these patient's activity measures to compute the stiffness parameter of the robot's impedance control, such that the patient participation on the therapy is always encouraged. Experimental results from a healthy subject wearing the exoskeleton and performing a set of flexion and extension knee movements show that the torque and stiffness estimation captures the intention of the user as well as the adaptive control strategy increases the level of participation of the user when performing the tasks.

5. ACKNOWLEDGEMENTS

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