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Design and Control of an Experimental Active Elbow Support for Adult Duchenne Muscular Dystrophy Patients

Joan Lobo-Prat[1], Peter N. Kooren[2], Arvid Q.L. Keemink[1], Micha I. Paalman[2], Edsko E.G. Hekman[1], Peter H. Veltink[3], Arno H.A. Stienen[1,4] and Bart F.J.M. Koopman[1]

Abstract— **Currently, a considerable group of adult Duchenne Muscular Dystrophy patients lives with severe physical impairments and strong dependency on care. Active arm supports can improve their quality of life by augmenting their arm's residual motor capabilities. This paper presents the design and control of an experimental active elbow support specially made to investigate different control interfaces with adult DMD patients. The system can be controlled either with EMG or force signals which are used as inputs for an admittance-based controller. A preliminary test with a 22-year-old DMD patient with no arm function left, shows that the system is capable of successfully supporting the elbow flexion-extension movements using the low-amplitude EMG and force signals that still remained measurable.**

I. INTRODUCTION

Duchenne Muscular Dystrophy (DMD) is the most common muscular dystrophy affecting 1 in 3500-6000 living male births [1]. DMD is caused by the absence or defect of the dystrophin protein [1]. Defective mutations in the dystrophin gene result in progressive degeneration of skeletal, respiratory and cardiac muscles leading to loss of independent ambulation in the early teens, followed by the development of scoliosis and loss of upper extremity function. The life expectancy of boys with DMD used to be no more than 20 years [2]. Long-term survival has improved substantially in the last five decades due to improvements in care, drugs and the introduction of home care technology such as artificial ventilators. As a result, currently there is a considerable group of adult DMD patients living with severe physical impairments and a strong dependency on care [3].

A special characteristic of DMD is that patients lose the ability to move their arms due to the weakening of proximal muscles, while distal muscles, such as hand and finger muscles, remain less affected [4]. Therefore, DMD patients can benefit from devices that support the arm movement taking advantage of the user's residual hand function and proprioception. Commercially available arm assistive devices

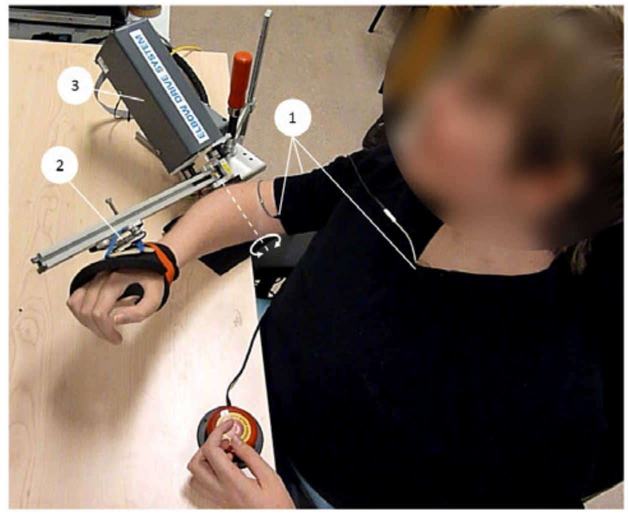

Fig. 1. An adult DMD patient testing the active elbow support system. 1) EMG electrodes. 2) Force sensor. 3) DC Motor with gearbox, encoder and torque limiter. Note that the DMD patient has severe shoulder contractures and therefore the alignemnt of the upper-arm with the device is not optimal.

support arm function by compensating its weight using mainly elastic elements such as metal springs or rubber bands [5]. These devices, however, become insufficient at the last stages of the disease, when patients can barely produce any force with their muscles [6], [7]. Therefore, adult DMD patients can potentially benefit more from active arm supports, which are able to provide the (extra) assistance that adult patients need for the performance of basic activities of daily living.

In order to operate active arm supports the user needs to communicate his motion intention to the device through a control interface. The selection of the control interface in response to specific user needs and capabilities—which in the specific case of DMD change significantly over time—is a crucial determinant of the usability of the assistive device. We think that two promising strategies to achieve a natural and intuitive control of the active arm support are EMG- and force-based interfaces.

The large majority of active orthoses and prosthesis existing today, including commercially available devices (e.g. DynamicArm, Otto Bock HealthCare GmbH., Duderstadt, Germany; mPower 1000, Myomo Inc., Cambridge, USA), are controlled using surface myoelectric signals [8], [9], [10]. The most common strategy for controlling active orthoses/exoskeletons is the estimation of muscular joint

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[1]Department Biomechanical Engineering, University of Twente, Enschede, The Netherlands. [2]Department of Physics and Medical Technology, VU University Medical Center, Amsterdam, The Netherlands. [3]Department of Biomedical Signals and Systems, University of Twente, Enschede, The Netherlands. [4]Department of Physical Therapy and Human Movement Sciences, Northwestern University, Chicago (IL), USA. Corresponding author: Joan Lobo-Prat, email: j.loboprat@utwente.nl

torques from the EMG signals of the muscles that mainly contribute to the supported motion. Several methods have been proposed for the estimation of muscle joint torques, including neural-networks[11], neuro-fuzzy classifiers [9] and Hill-type models [8], [12]. Lenzi et al. [10] designed a simpler EMG-based controller that provided an assistive torque proportional to the envelope of the EMG signal. The strategy of the authors was to exploit the high adaptability of the human controller to compensate for the torque estimation errors caused by the simplification of the EMG-torque relationship.

Force-based interfaces have been used in assistive-powered wheelchairs [13], in which the wheelchair detects and amplifies the force applied by the user. Recent studies implemented force-torque sensors [7], [14], or simple force sensor resistors for the control of active upper-extremity orthoses [15] and prosthesis [16]. These kind of interfaces generally implement admittance control strategies where the output acceleration, velocity or position is related to the input force [17]. Haptic force-based control interfaces are very often implemented in rehabilitation robots where patients need training to regain motor control, mobility and strength [18]. The advantage of implementing haptic interfaces, such as admittance or impedance control, in assistive robots is that the apparent dynamics of the robot can be modified to enhance the interaction experience [19].

Compared to the large number of active arm prosthetic devices available for amputees [20], very few active devices for supporting upper extremity function of people with severe muscular weakness are being developed. An example is the active version of the WREX (JAECO Orthopedic, USA), which can actively support vertical shoulder and elbow movements with two series-elastic actuators controlled by the user's residual force [14]. Another example is the work of Baklouti et al. [21], [22], who developed a 4 degree of freedom (DOF) active arm support that can be controlled using the residual arm forces or facial expressions.

In the Flextension A-Gear project [23] we have the goal of developing an inconspicuous five DOF active arm support that adapts to the time-varying needs of DMD patients. The selection of the most suitable control interface for the A-Gear arm support requires a better understanding of the limitations and capabilities of different control strategies, through objective and quantitative evaluations during functional tasks. This paper presents the design and control of an experimental active elbow support for adult DMD patients that was specifically built to investigate the performance of EMG- and force-based control interfaces (Fig. 1). We show that the system is capable of successfully supporting the elbow flexion-extension movements during a screen-based discrete tracking task using the low-amplitude EMG and force signals that still remain measurable in a 22-year-old DMD patient. Note that the system presented in this paper does not represent an early version of the A-Gear arm support.

II. DESIGN

A. Requirements

The active elbow support is designed to investigate different control interfaces while performing a simple movement close to a basic activity of daily living such as eating, drinking or face scratching. Elbow flexion-extension movements against gravity were chosen considering that individuals with muscular weakness particularly need support in the vertical direction. From this concept the following requirements were derived.

The system has to actively support elbow flexion and extension movements in a (maximum) range of 45 to 135 degrees. To achieve a natural feeding movement, an adjustable DOF that allows shoulder internal-external rotation is also required. Additionally, the system should allow the freedom to be installed over a table surface, so that it can be adjusted to the sitting position of user. The system needs to be comfortable and ensure proper alignment with the elbow joint. All the areas that have contact with the user need to be soft due to the high skin sensitivity of adult DMD patients.

The required elbow angular velocity and torque were determined by measuring the movement of a healthy subject during a feeding task. The kinematic data was used as input for a simple dynamic model of an inverted pendulum which represented the forearm of the user (see Table I). A mass of 1.5 kg was chosen as the maximum endpoint payload.

In order to obtain a satisfactory control performance of haptic control strategies, the system must have low inertia, low friction and low backlash. Finally, it is required that safety is always guaranteed for the user and the researcher or therapist operating the system.

TABLE I
KINEMATIC AND DYNAMIC REQUIREMENTS FOR EACH DOF

DOF	ROM (deg)	Velocity (deg/s)	Torque (Nm)
Elbow FE	45 - 135	90	7.5
Shoulder IE rot.	0 - 45	passive	passive

Note: ROM, range of motion; DOF, degree of freedom; FE, flexion-extension; IE rot., internal-external rotation.

B. Mechanics and Actuation

The elbow support (Fig. 1) has one active rotational DOF actuated by a brushed DC motor (A-max 32, Maxon Motor AG, Switzerland) connected to a gearbox with a reduction ratio of 111:1 (GP 32 C, Maxon Motor AG, Switzerland). The motor axis is aligned with the elbow joint of the user which rests over the table surface. Perpendicular to the motor axis an aluminum beam extends along the forearm in which the hand of the user is fixated with an ergonomic plastic hand cup and a Velcro strap. In order to ensure a proper alignment between the motor axis and the elbow joint, the hand cup is connected to the aluminum beam through a slider that allows linear displacement along the beam. Additionally, a soft foam pad with a spherical hole is placed under the elbow joint of the user to increase comfort and stability of the elbow joint.

The system is equipped with several safety features in order to avoid any harm to the users. The torque transmitted to the elbow is limited with a mechanical torque limiter at 9 Nm (ESL, R+W Antriebselemente GmbH, Germany). Since DMD patients have low strength in their hands, the system is provided with a highly sensitive emergency stop button which can interrupt the power line of the DC motor. Finally, since adult DMD patients usually present severe joint contractures, the range of motion (ROM) of the active elbow support can be customized using two adjustable mechanical end stops to prevent hyperextension-hyperflexion of the elbow joint. Note that these safety features are heavily influenced by the requirements of the The Medical Ethics Committee of the Radboud University Nijmegen Medical Centre (Nijmegen, The Netherlands).

C. Sensors

The angular position of the active elbow support is measured with an optical encoder (500 pulses per revolution; HEDL 5540, MicroMo Electronics Inc., USA) attached at the back of the motor. The interaction forces between the human and the device are measured with a one DOF load sensor (LSB200 - 5lb, FUTEK Advanced Sensor Technology Inc., USA) located between the plastic hand cup and the aluminum beam. The force signals are amplified by a strain gauge amplifier with an output voltage of ± 10 volt. The attachment of the force sensor was designed in such a way that mainly forces acting parallel to the circular motion of the aluminum beam are measured. The muscle activation signals are measured from the biceps and triceps branchii muscles, which are the muscles that mainly contribute to the elbow flexion-extension movements. Two single differential-surface EMG electrodes (Bagnoli DE-2.1., Delsys, USA) are placed parallel to the muscle fibers according to the SENIAM (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles) recommendations [24]. The signals are amplified by a Delsys Bagnoli-16 Main Amplifier and Conditioning Unit with a bandwidth of 20 to 450 Hz and a gain of 1000. The active elbow support is also equipped with a one DOF joystick with adjustable spring stiffness. Our motivation behind implementing a classic hand-joystick is based on the fact that this type of interface is commonly used by individuals with severe muscular weakness to control electric wheelchairs, domestic devices and external robotic arms. Therefore the first time that DMD patients use the active elbow support, they can get to know the system dynamics using a control interface that is very familiar to them.

D. Signal Acquisition and Control Hardware

All the signals from the sensors are sent to a real time computer (xPC Target 5.1, MathWorks Inc., USA) by means of a National Instruments card (PCI-6229; National Instruments Corp., USA), which performs the analog-to-digital conversion with a sampling frequency of 1 KHz and 16-bits resolution. The controller also runs on the real-time computer and sends the control signals to the motor driver (UK1122-L298 Dual H-Bridge 4A, Cana Kit Corp.,Canada) through the same National Instruments card. The Matlab/Simulink graphical user interface (MATLAB 2012b, MathWorks Inc., USA) runs on a computer with Windows operating system (Microsoft Corporation, USA) and is connected to the real time computer by a local area network using TCP/IP protocol.

E. Signal Processing

In order to obtain the envelopes, which are known to resemble the muscle tension waveforms during dynamic changes of isometric forces [25], the EMG signals were full-wave rectified and smoothed using a second order low-pass Butterworth filter with a cutoff frequency of 3 Hz. The filter settings were chosen considering previous studies on EMG control [10], [8] and pilot trials on our setup.

The normalized EMG signals ($A_{nor,k}(i)$) and the resultant EMG control signal ($U_{emg}(i)$) are obtained using (1) and (2) respectively.

$$A_{nor,k}(i) = \frac{A_{env,k}(i) - A_{res,k}}{MVIC_k} \tag{1}$$

$$U_{emg}(i) = A_{nor,b}(i) - A_{nor,t}(i) \tag{2}$$

where subscript k represents the abbreviations of the biceps (b) and triceps (t) muscles, $A_{env,k}(i)$ denotes the processed EMG signal at the i^{th} time step, $A_{res,k}$ represents the average of the processed EMG signal during rest and $MVIC_k$ represents the mean maximum magnitude of $A_{env,k}(i)$ during two senconds of maximum voluntary isometric contraction (MVIC).

The gravity compensated force signals ($F_{vol}(i,\theta)$) and the resultant force control signal ($U_{for}(i,\theta)$) are obtained using (3) and (4) respectively.

$$F_{vol}(i,\theta) = F_{sen}(i,\theta) - F_{gra}(\theta) \tag{3}$$

$$U_{for}(i,\theta) = \begin{cases} \frac{F_{vol}(i,\theta)}{MVIF_u}, & \text{if } F_{vol}(i,\theta) > 0 \\ \frac{F_{vol}(i,\theta)}{MVIF_d}, & \text{if } F_{vol}(i,\theta) < 0 \end{cases} \tag{4}$$

where $F_{sen}(i,\theta)$ denotes the measured force signal at the i^{th} time step and at angle θ, $F_{gra}(\theta)$ represents the gravitational force measured at angle θ and $MVIF_u$ and $MVIF_d$ represents the mean maximum magnitude of $F_{sen}(i,\theta)$ during biceps and triceps MVIC respectively. $F_{gra}(\theta)$ is obtained measuring the forces during a slow descending movement from the upper limit to the lower limit of the elbow support with the arm of the subject relaxed and attached to the system. Pilot trials showed that that this method provided a better estimation of the gravity forces than using a simple dynamic model.

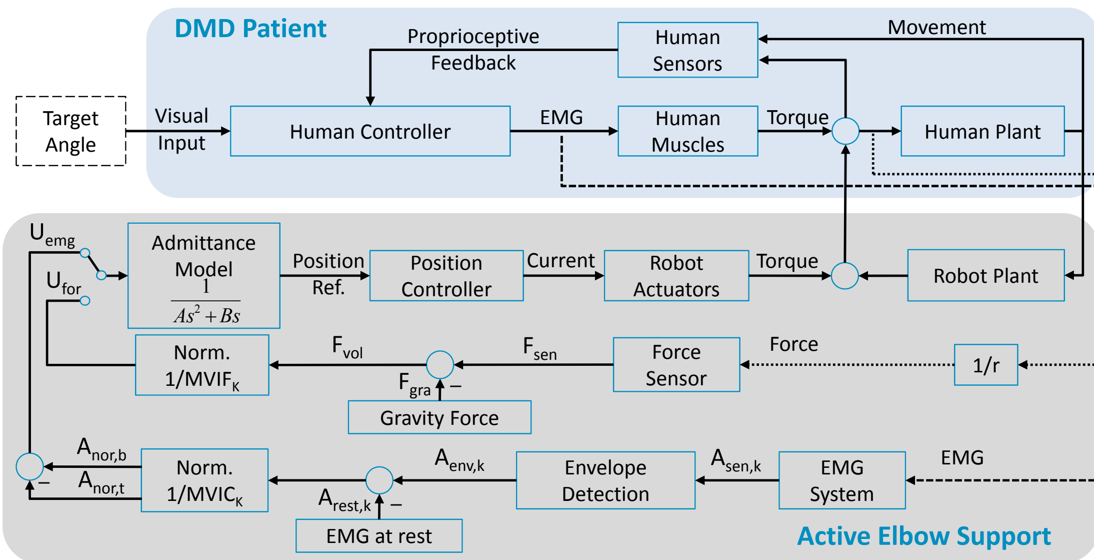

Fig. 2. Diagram of the control architecture implemented in the active elbow support.

F. Control

Fig. 2 shows the control architecture implemented in the active elbow support. An second order admittance model with a virtual inertia parameter (A) and a virtual damping parameter (B_v) was implemented as a high level controller. Both A and B parameters were set to 0.5. These values were chosen from pilot trials. The position reference obtained from the admittance model is controlled using a PD controller. The proportional and derivative gains of the PD controller were tuned using the common Ziegler-Nichols method. Fig. 3 shows the closed loop transfer function of the low level position controller (i.e. from the reference position to the measured position), which was estimated using a multisine signal as a reference position. The bandwidth of the position controller is around 3 Hz, which is high enough taking into account that our application has a target bandwidth of 1 Hz.

III. SYSTEM VALIDATION

A preliminary test to validate the usability of the active elbow support was carried out with a 22-year-old DMD patient. The Medical Ethics Committee of the Radboud University Nijmegen Medical Centre approved the study design, protocols and procedures. The participant was classified according to the Brooke upper extremity function scale [26] with a score of 6 (i.e. no arm function is left). The participant could still control his electric wheelchair and an external robotic arm with a two DOF joystick and several push buttons installed on the table of the electric wheelchair.

A one-dimensional discrete position-tracking task was presented to the participant on a computer screen by means of a C# (Microsoft Visual Studio, Microsoft Corporation, USA) audiovisual interface. The participant was asked to bring a circular cursor, which represented the end point of the elbow support, as close as possible to the center of a circular target and remain inside the target area (i.e. target angle ± 1 degree) for three seconds as predefined stabilization time. When the cursor was inside the target area a sound was played in order to inform the participant that he was in the right position. Three target angles were linearly distributed inside the ROM of the device. The participant performed the tracking task for 60 randomly ordered targets with each control interface. The first 20 targets were used as training trials. Fig. 4 shows the angular displacement of the elbow support along time resulting from the normalized EMG and force inputs during the last 40 targets. The bottom row of insets in Fig. 4 shows the force signals measured during EMG control and the EMG signals measured during force control. The participant presented a maximum force upwards and downwards of 1.5 N and 2 N respectively; and a maximum EMG voltage of the biceps and triceps muscles of 90 mV and 20 mV respectively. Note that the normalized force and EMG signals show for short intervals amplitudes larger than 1 because the MVICs or MVIFs were calculated taking the mean value during the two seconds that these lasted.

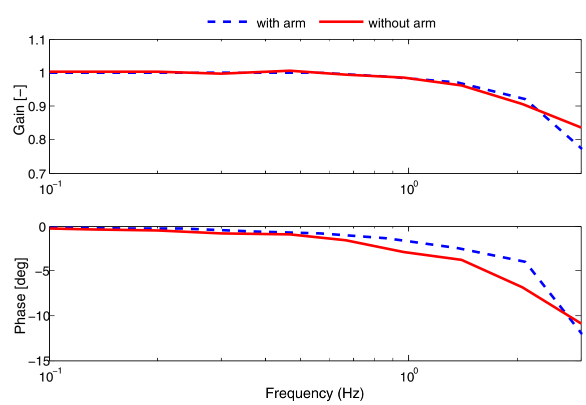

Fig. 3. Closed-loop transfer function of the low level position controller with (blue dashed linered soid line) and without (red solid line) the load of the forearm.

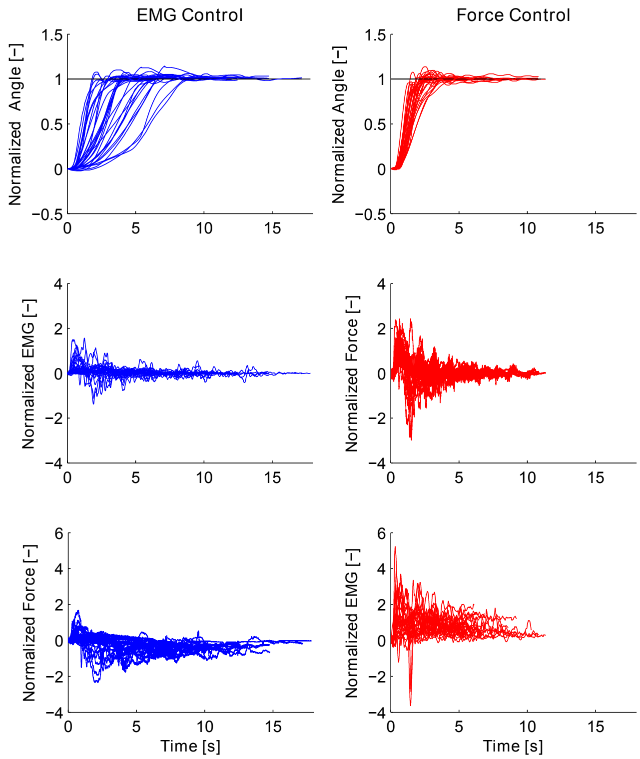

Fig. 4. Top) Normalized angular displacement of the elbow support along time for EMG (blue) and force control (red). Middle) Normalized EMG (blue) and force (red) inputs used for controlling the active elbow support system along time. Bottom) Normalized EMG (red) and force (blue) signals measured during the control task along time.

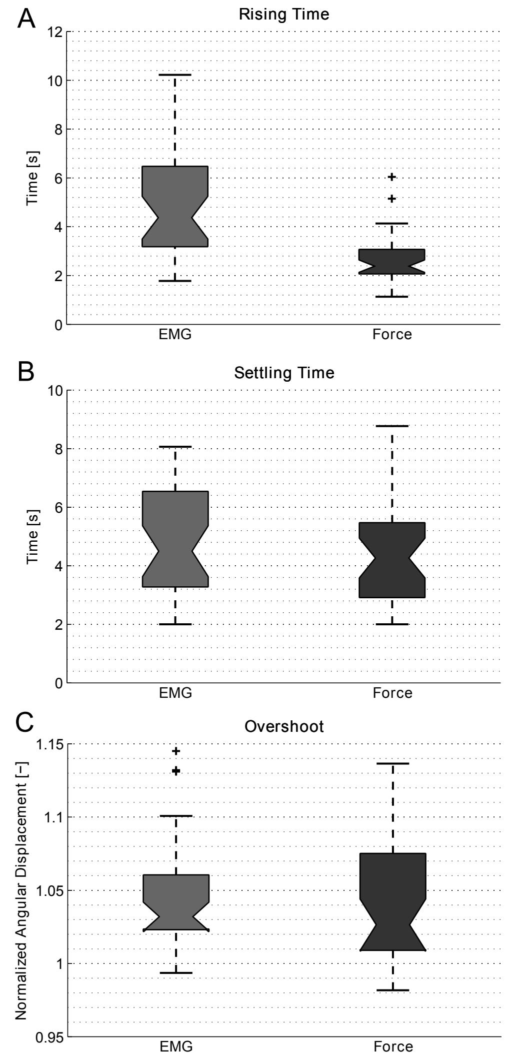

Fig. 5. A) Boxplots for the rising time parameter. B) Boxplots for the settling time parameter. C) Boxplots for the overshoot parameter. (+) indicates an outlier.

The performance of the EMG- and force-based control interfaces during the tracking task were evaluated in terms of rising time, settling time and overshoot. The results from the performance analysis show that the movements using EMG control present a remarkably longer rising time than when using force control (Fig. 5A). No large differences were found in terms of settling time (Fig. 5B) and overshoot (Fig. 5C). Another noticeable difference in Fig. 4 between EMG and force control is that the forces measured during EMG control are considerably lower than the forces measured during force control, suggesting that EMG control required less effort. Accordingly, the opinion of the participant was that he experienced force control more fatiguing than EMG control. Furthermore, the participant had a very positive experience, both in terms of comfort and performance, controlling the movement of his forearm with the EMG- and the force-based interfaces.

IV. DISCUSSION

While the results of this preliminary test are not conclusive to decide which control interface performs best due to the limited number of participants, we can foresee that performance differences may exist between EMG- and force- based control when used by adult DMD patients.

There is a fundamental difference on how EMG- and force-based control interfaces interact with the human plant. While a force-based interface is affected by the human plant (force loop closure), the EMG-based interface is 'detached' from it (see Fig. 2). Therefore, we can presume that an EMG-based interface measures signals that better represent the movement intention of the human controller since they are less disturbed by the human plant. Furthermore, a critical issue in force-based interfaces, that becomes even more critical in patients with severe muscular weakness, is the need for a highly accurate estimation of gravitational forces in order to identify the user's voluntary forces. The work of Ragonesi et al. [27] showed that voluntary forces for weak individuals were very hard to measure since gravitational forces were approximately ten times larger. Subject specific models were suggested as a strategy to optimize the identification of voluntary forces. Furthermore, adult DMD patients also present significantly higher joint and muscle stiffness [28] which adds more complexity to the identification of voluntary forces. In this respect, the use of EMG-based interfaces would also be advantageous since they are not disturbed by gravity. On the other hand, EMG-based interfaces present several practical issues, including the poor long term stability of the measurements, the high sensitivity to electrode location, the time required to place the electrodes and the uncomfortable feeling that the multiple electrodes may produce in contact with the skin for a long period of time.

V. CONCLUSIONS AND FUTURE WORK

This paper presents the design and control of an experimental active elbow support specially made to investigate EMG-and force-based control interfaces in adult DMD patients. We designed a system that actively assists elbow flexion and extension movements and that can be oriented such that movements of the hand from the table to the mouth/face are natural. We implemented both force- and EMG-based interfaces as inputs for an admittance model. A preliminary experimental validation of the system was carried out with a 22-year-old DMD during a screen-based position-tracking task. The results show that the system is capable of successfully supporting the elbow flexion-extension movements using the low-amplitude EMG and force signals that still remain measurable. From the results of the preliminary test we can foresee that performance differences may exist between EMG- and force-based control. Future work will involve a thorough evaluation of EMG- and force-based interfaces during the same screen-based tracking task with a larger group of participants. The results of the performance evaluation together with additional design criteria will be analyzed to decide which interface is the most suitable for the control of the wearable active arm support that is being developed in the Flextension A-Gear project [23].

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