Implementation of EMG- and Force-Based Control Interfaces in Active Elbow Supports for Men With Duchenne Muscular Dystrophy: A Feasibility Study

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***Abstract*—While there is an extensive number of studies on the development and evaluation of electromyography (EMG)- and force-based control interfaces for assistive devices, no studies have focused on testing these control strategies for the specific case of adults with Duchenne muscular dystrophy (DMD). This paper presents a feasibility study on the use of EMG and force as control interfaces for the operation of active arm supports for men with DMD. We have built an experimental active elbow support, with a threefold objective: 1) to investigate whether adult men with DMD could use EMG- and force-based control interfaces; 2) to evaluate their performance during a discrete position-tracking task; and 3) to examine users' acceptance of the control methods. The system was tested in three adults with DMD (21–22 years). Although none of the three participants had performed any voluntary movements with their arms for the past 3–5 years, all of them were 100% successful in performing the series of tracking tasks using both control interfaces (mean task completion time EMG: 6.8 ± 4.8 s, force: 5.1 ± 1.8 s). While movements with the force-based control were considerably smoother in Subject 3 and faster in Subject 1, EMG based-control was perceived as less fatiguing by all three subjects. Both EMG- and force-based interfaces are feasible solutions for the control of active elbow supports in adults with DMD and should be considered for further investigations on multi-DOF control.**

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***Index Terms*—Arm support, assistive devices, Duchenne, electromyography (EMG) control, force control, human–robot interaction.**

I. Introduction

DUCHENNE muscular dystrophy (DMD) is the most common form of muscular dystrophy in children, with an incidence of 1 in 5000 male live births [1]. Defective mutations in the dystrophin gene result in progressive degeneration of skeletal, respiratory and cardiac muscles leading to loss of independent ambulation by the age of ten years, followed by the development of scoliosis and loss of upper extremity function [2], [3]. The mean life expectancy of boys with DMD used to be no more than 20 years [4], but in the last five decades, long-term survival has improved substantially due to improvements in drugs and the introduction of home care technology, such as artificial ventilators. As a result, currently, there is a considerable group of adults with DMD living with severe physical impairments and a strong dependency on care up to their 30's [5]. Commercially available arm supports, which mainly provide passive gravity compensation [6], become insufficient to support the arm function of adults with DMD [7]. Active arm supports, which can provide extra assistance, could enable adults with DMD to continue performing basic activities of daily living and participate in social activities.

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, is a crucial determinant of the usability of the assistive device. We consider that surface elcectromyography (EMG) and force-based interfaces are two promising strategies to achieve a natural control of active arm supports as they have been widely implemented in prostheses and orthoses/exoskeletons [8], [9].

The clinical standard EMG-based control strategy implemented in upper limb prosthetics is a simple amplitude-based dual site control approach also known as direct control [10]. This method measures EMG from two independent residual muscles, or by distinguishing different activation levels of one residual muscle. Switching techniques such as muscle co-contraction are commonly implemented for enabling the sequential operation of different degrees of freedom (DOF). More advanced EMG-based control strategies for operating active

TABLE I
SUMMARY OF PARTICIPANTS' CHARACTERISTICS

Subject's Code	Age (years)	Brooke Scale	Preferred Arm	ROM (deg)	Pas. Sup. ROM (deg)	E_{mvic_b} (mV)	E_{mvic_t} (mV)	F_{mvic_f} (N)	F_{mvic_e} (N)
1	22	5	Right	58.5	15	0.09	0.02	0.94	2.7
2	21	5	Right	60.5	5	0.004	0.004	0.80	1.47
3	22	5	Right	48.0	5	0.002	0.001	0.55	0.82

Note: ROM, passive range of motion reached when the forearm is moved by a therapist; Pas. Sup. ROM, range of motion that the subject can reach when using passive support for gravity compensation of the forearm;E_{mvic_b}, EMG amplitude during maximal voluntary isometric contraction of the biceps; E_{mvic_t}, EMG amplitude during maximal voluntary isometric contraction of the triceps; F_{mvic_f}, force amplitude during maximal isometric voluntary contraction of the biceps; F_{mvic_e}, force amplitude during maximal isometric voluntary contraction of the triceps. Note that the values of E_{mvic_b} and E_{mvic_t} do not indicated the maximum EMG electrical potential measured but the maximum signal value after the envelop detection. The EMG evelopes during rest (i.e. noise level) presented maximum values of 0.0007 mV in all subjects.

prostheses and orthoses/exoskeletons are based on estimating joint angles or torques from the EMG signals of the muscles that mainly contribute to the supported motion. Common estimation methods include pattern-recognition-based algorithms [11] and regression-based algorithms [12]–[14].

Force-based interfaces have been used in assistive-powered wheelchairs [15], in which the wheelchair detects and amplifies the force applied by the user. Other studies implemented force-torque sensors, or simple force sensor resistors for the control of active upper-extremity orthoses [7], [16], [17] and prostheses [18]. Force-based interfaces generally implement control strategies where the output motion is proportional to the input force. Additionally, haptic force-based control interfaces are often implemented in rehabilitation robots where patients need training to regain motor control, mobility and strength [9]. The advantage of implementing haptic interfaces, such as admittance or impedance control, in human-interactive robots is that the apparent dynamics of the robot can be modified to enhance the interaction.

While there is an extensive number of studies on the development and evaluation of EMG- and force-based control interfaces for prostheses and orthoses/exoskeletons [19]–[23], no studies have focused on testing these control strategies for the specific case of men with DMD. The selection of the most suitable control interfaces to operate an active arm support for adults with DMD required a better understanding of the limitations and capabilities of EMG- and force-based control interfaces through objective and quantitative evaluations during functional tasks.

This paper presents a feasibility study on the use of EMG and force as control interfaces to operate active arm supports for adults with DMD. We built an experimental active elbow orthosis, the Flextension Elbow Drive (Fig. 1), with a threefold objective: 1) to investigate whether adult men with DMD with very limited arm function could use EMG- and force-based control interfaces; 2) to evaluate their performance during a discrete position-tracking task; and 3) to examine users' acceptance of the control methods. The system presented in this paper is a research platform for the evaluation of EMG- and force-based control interfaces and does not represent an early prototype of an actual arm support for daily use.

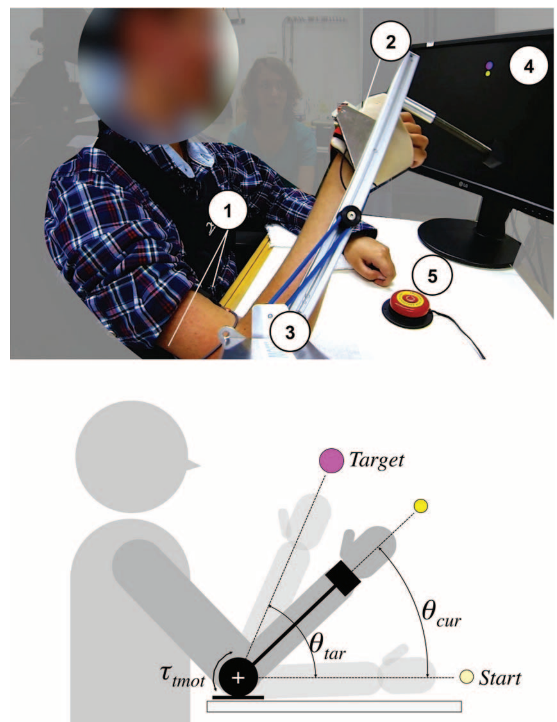

Fig. 1. (Top) An adult man with DMD using the Flextension Elbow Drive system during the control interface evaluation. 1) EMG electrodes. 2) Force sensor. 3) Motor housing. 4) Display of the experimental task. 5) Emergency stop. (Bottom) Schematic drawing of a subject using the Elbow Drive system and performing the discrete position-tracking task. The Elbow Drive generates an assistive torque (τ_{mot}) that helps the subject to reach from the start position the target angle (θ_{tar}) with his forearm (θ_{cur}). Note that in reality the axis of the motor is tilted vertically to align the movement of the elbow to a comfortable/natural orientation.

II. METHODS

A. Participants

Three adults with DMD participated in this study. Participants were carefully selected considering that they should have been users of passive arm supports and have experienced severe difficulties using them due to muscular weakness and increased joint stiffness. All subjects were not able to perform any voluntary movements with their arms for the past 3–5 years. Subject's demographic information is shown in Table I.

1) Subject 1 (S1): presented no arm function and limited hand function (Brooke scale: 5 [24]) that allowed him to do some writing and drive an electric wheelchair using a hand joystick. Subject S1 was not able to actively flex or extend his wrist. Using the Elbow Drive with only passive weight compensation Subject S1 was still able to move his elbow over a range of motion of 15 degrees. Subject S1 suffered from sever scoliosis which was surgically corrected at the age of 12.

2) Subject 2 (S2): presented no arm function and limited hand function (Brooke scale: 5) that only allowed him to use a computer mouse (using a passive arm support) and drive an

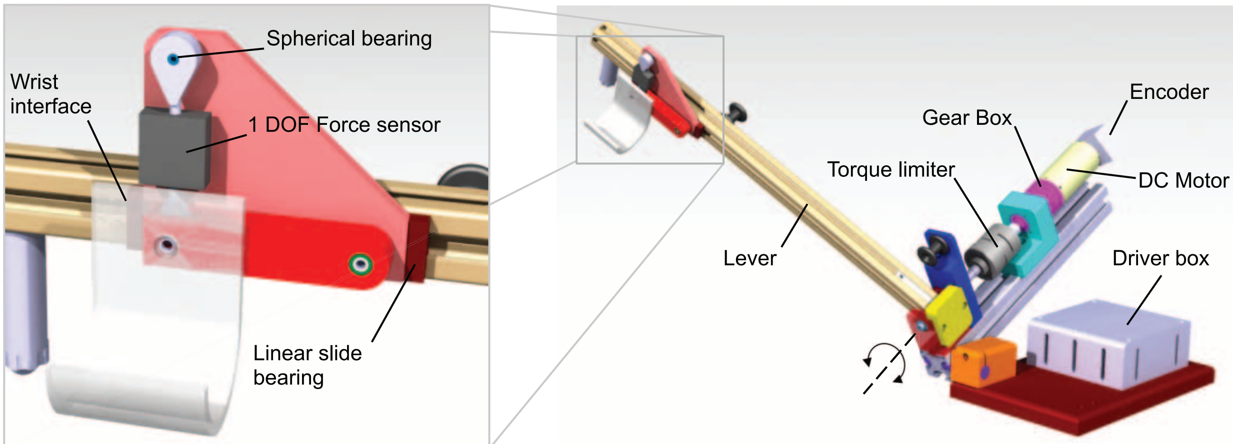

Fig. 2. Components of the Elbow Drive system. Elbow Drive system was build to investigate if subjects with DMD could use EMG- and force-based interfaces to operate an active arm support. A dc motor drives the lever through a mechanical torque limiter that prevents high torques from being transmitted to the elbow joint. The wrist of the participant is attached to the Elbow Drive system using an anatomically shaped plastic interface (in the figure the shape of the interface is simplified). The interaction forces between the user and the Elbow Drive system are measured with a one DOF force sensor mounted between the wrist cup and the lever arm. We used a linear slide bearing to prevent discomfort due to misalignment between the elbow joint and the motor axis.

electric wheelchair using a hand joystick. However, Subject S2 was not able to write or actively flex or extend his wrist. Subject S2 suffered from sever scoliosis which was surgically corrected at the age of 14.

3) Subject 3 (S3): presented no arm function and limited hand function (Brooke scale: 5) which did not enable him to write or use a computer mouse. Subject S3 was still able to control an electric wheelchair using a hand joystick. Subject S3 used an artificial ventilator and did not suffer from severe scoliosis.

The Medical Ethics Committee of the Radboud University Nijmegen Medical Center approved the study design, protocols and procedures, and informed consent was obtained from each subject.

B. Setup: Elbow Drive System

The Elbow Drive was designed to investigate different control interfaces while performing a challenging yet simple and functional movement. Elbow flexion-extension movements against gravity were chosen considering that it resembles a functional movement needed when eating, drinking or face scratching, and that individuals with muscular weakness specially need support in upward movements. The Elbow Drive (Figs. 1 and 2) has one active rotational DOF aligned with the elbow joint of the user which rests on 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 hand interface made of thermoplastic.

The interaction forces between the human and the device were measured with a one DOF force sensor (LSB200 – 5lb, FUTEK Advanced Sensor Technology Inc., USA) located between the plastic hand cup and the aluminum lever. The attachment of the force sensor was designed in such a way that only forces acting perpendicular to the aluminum lever are measured (see close-up image in Fig. 2). The muscle activation signals were 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 dry-electrodes (Bagnoli DE-2.1., Delsys, USA) were placed parallel to the muscle fibers according to the SENIAM (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles) recommendations [25] and manual muscle exploration. The EMG signals were 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 Elbow Drive was also equipped with a one DOF hand-joystick with the only purpose of letting the participants familiarize themselves with the system dynamics and the tracking task.

All the signals from the sensors were 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 performed the analog-to-digital conversion with a sampling frequency of 1 KHz and 16-bit resolution. The controller was also running on the real-time computer and was sending the pulse width modulated (PWM) control signals to the motor driver (UK1122-L298 Dual H-Bridge 4A, Cana Kit Corp., Canada) through the same National Instruments card. Further details on the design requirements, mechanics, actuation, sensors and safety measures of the Elbow Drive system are described in [26].

C. Signal Processing

The envelope detection of the EMG signals was performed applying a full-wave rectification and 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 [27], [28] and pilot trials on our setup. The normalized EMG signals, $E_{\mathrm{nor},k}(i)$, and the resultant EMG control signal, $u_{\mathrm{emg}}(i)$, were obtained using (1) and (2) respectively

$$E_{\mathrm{nor},k}(i) = \frac{E_{\mathrm{env},k}(i) - E_{\mathrm{res},k}}{E_{\mathrm{mvic}_k}} \tag{1}$$

$$u_{\mathrm{emg}}(i) = E_{\mathrm{nor},b}(i) - E_{\mathrm{nor},t}(i) \tag{2}$$

where subscript k represents the abbreviations of the biceps (b) and triceps (t) muscles, $E_{\mathrm{env},k}(i)$ denotes the processed EMG signal at the ith time step, $E_{\mathrm{res},k}$ represents the average signal amplitude of the EMG envelopes during 3 s at rest, and E_{mvic_k} represents the mean maximum magnitude of $E_{\mathrm{env},k}(i)$ during 2 s of maximum voluntary isometric contraction (MVIC; see Table I).

To properly detect the movement intention of the user in force-based interfaces, it is critical to distinguish the voluntary forces from any other force, such as gravity or joint stiffness forces. Note that in Fig. 4 the measured force (F_{sen}) is the combination of the muscle torque (τ_{mus}) and the intrinsic/passive torques of the arm (τ_{pas}). In the Elbow Drive system, the estimated voluntary forces of the user, $\hat{F}_{\mathrm{vol}}(i,\theta)$, and the resultant force control signal, $u_{\mathrm{for}}(i,\theta)$, are obtained using (3) and (4) respectively

$$\hat{F}_{\mathrm{vol}}(i,\theta) = F_{\mathrm{sen}}(i) - \hat{F}_{\mathrm{com}}(\theta) \tag{3}$$

$$u_{\mathrm{for}}(i,\theta) = \begin{cases} \frac{\hat{F}_{\mathrm{vol}}(i,\theta)}{F_{\mathrm{mvic}_f}}, & \text{if } \hat{F}_{\mathrm{vol}}(i,\theta) > 0 \\ \frac{\hat{F}_{\mathrm{vol}}(i,\theta)}{F_{\mathrm{mvic}_e}}, & \text{if } \hat{F}_{\mathrm{vol}}(i,\theta) < 0 \end{cases} \tag{4}$$

where $F_{\mathrm{sen}}(i)$ denotes the measured force signal at the ith time step, $\hat{F}_{\mathrm{com}}(\theta)$ represents the estimated compensation force measured at angle θ, and F_{mvic_f} and F_{mvic_e} represents the mean maximum magnitude of $\hat{F}_{\mathrm{vol}}(i,\theta)$ during biceps (flexion) and triceps (extension) MVIC, respectively. The $\hat{F}_{\mathrm{com}}(\theta)$ was obtained by measuring the forces during a slow and constant descending movement (0.05 rad/s) from the upper limit to the lower limit of the elbow support with the arm of the subject relaxed and attached to the system (Fig. 3). A 10th order polynomial function was then fitted to the measured force to reduce any possible disturbances during the measurement. Pilot trials showed that this measurement-based method was able to accurately estimate gravity and joint stiffness forces along the range of motion of the elbow. A video showing the compensation force measurement can be found in [8] as additional file 4.

D. Control

Fig. 4 shows the control diagram of the physiological (i.e., man with DMD) and the assistive system (i.e., Elbow Drive). The objective of the participant is to reach the target angle (θ_{tar}) with the arm. The arm muscles of the participant are too weak to generate the muscular torque (τ_{mus}) necessary to move the passive human arm dynamics and therefore an assistive device is used. The Elbow Drive can use either the interaction torque (τ_{int}) between the user and the Elbow Drive (measured with a force sensor, F_{sen}) or EMG signals (E_{sen}) to detect the motion intention of the user and support the movement towards the target angle. After processing the EMG and force signals as described in Section II-C, the resulting control signals (u_{for} or u_{emg}) are filtered by a second order transfer function ($H_{\mathrm{id}}(s)$) that represents the virtual dynamics of a mass-damper system

$$H_{\mathrm{id}}(s) = \frac{1}{0.5s^2 + 0.5s}. \tag{5}$$

The specific values of the parameters of the interface dynamics function were chosen from pilot trials with men with DMD. Note that the parameters of the interface dynamics are effectively scaled to the MVIC values by the normalization of the control signals u_{for} and u_{emg}. The angle reference signal resulting from the virtual dynamics (θ_{ref}) is sent to a low-level position feedback controller (i.e., PD controller) that operates the motor. The torque generated by the dc motor (τ_{mot}) together with the interaction torque (τ_{int}) move the passive dynamics of the Elbow Drive system, which assists the user to move his arm, making possible for the participant to perform elbow flexion/extension movements (θ_{cur}) and eventually reach the target angle (θ_{tar}).

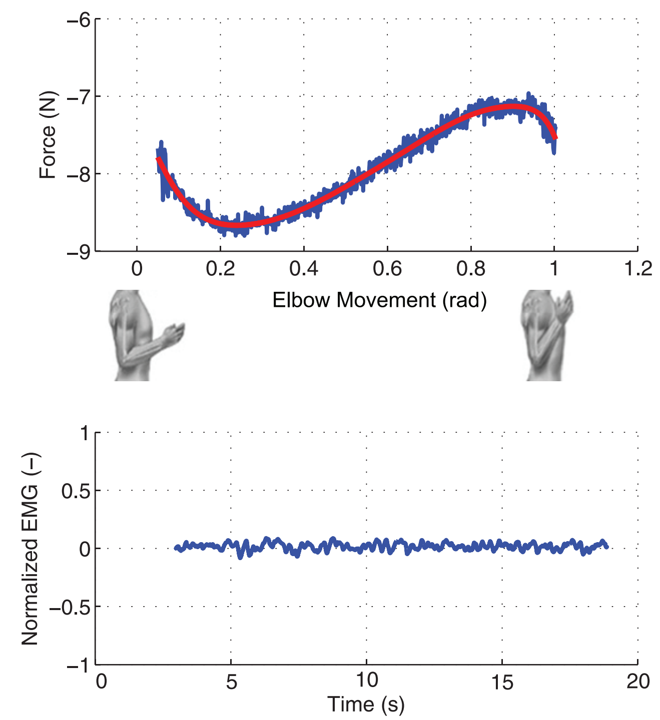

Fig. 3. (Top) Representative data of a compensation force measurement from one of the participants. For the measurement of the gravity and joint stiffness forces ($\hat{F}_{\mathrm{com}}$), the Elbow Drive moved with a low constant speed (0.05 rad/s) across the ROM of the elbow while the subject was relaxed. In blue the force measured over the elbow's ROM in one of the subjects. In red the curve fitted with a 10th order polynomial function. (Bottom) The u_{emg} signal of the same subject during the compensation force measurement. This signal was used to supervise if the subject was relaxed during this procedure.

E. Experimental Protocol

After a detailed explanation of the purpose and procedure of the experiment, the ROM of the setup, the elbow alignment and the amount of padding on the hand cup were adjusted to the convenience of the participant. The participants first used the one DOF joystick, operated with the contralateral hand, to control the Elbow Drive in order to familiarize themselves with the movement of the arm and the tracking task.

A one-dimensional discrete position-tracking task was presented to the participant on a computer screen by means of a custom-made $C^{\#}$ (Microsoft Visual Studio, Microsoft Corporation, USA) audiovisual interface (Fig. 1). The participant was asked to bring a yellow circular cursor (θ_{cur}), which represented the angle of the elbow, as close as possible to the center of a purple circular target (θ_{tar}) and remain inside the target zone (i.e., $\pm 2^{\circ}$) for two seconds as predefined dwell time. When the cursor was inside the target zone, a beeping sound was played in

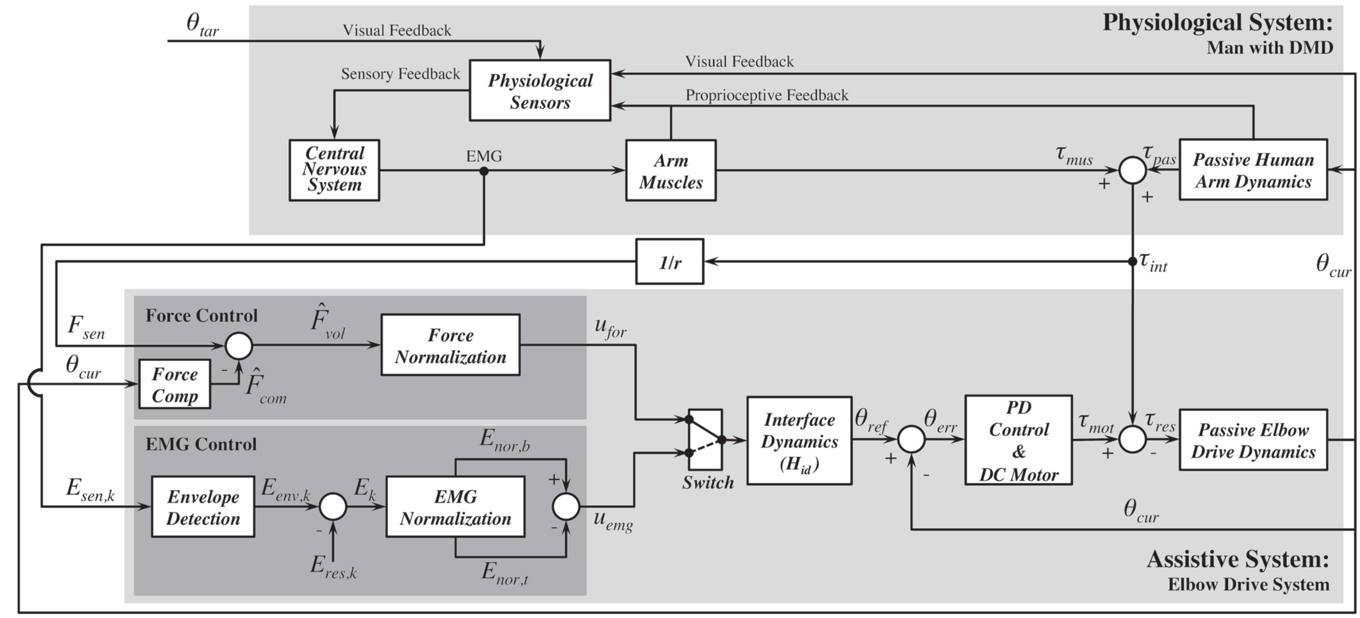

Fig. 4. Control diagram of the physiological and assistive system. Either force (F_{sen}) or EMG signals (E_{sen}) are used to derive the motion intention of the user and control the Elbow Drive system. The interaction torque (τ_{int}), which is a combination of the muscular torque (τ_{mus}) and the passive/intrinsic human arm torque (τ_{pas}) is measured by a force sensor (F_{sen}) at a distance r from the axis of rotation of the elbow joint. The task of the user is to reach the target angle (θ_{tar}) with the Elbow Drive (θ). The subject has visual feedback of the elbow and target angle. After the signal processing and the virtual dynamics (H_{id}), the resulting angle reference signal (θ_{ref}) is send to a low-level position feedback controller that operates the motor. The resulting torque (τ_{res}) generated by the dc motor (τ_{mot}) together with the interaction torque (τ_{int}) moves the passive Elbow Drive's and human arm dynamics.

order to inform the participant that he was in the correct position. Three target angles were linearly distributed over the elbow's ROM of the participant. Note that from this distribution of target angles two movement lengths can be distinguished: long movements, from minimum to maximum ROM, and short movements, from minimum or maximum to half of the ROM. The target angles located near the limits of the ROM were placed, within the ROM, 5 degrees from the limits in order to prevent collisions with the mechanical end-stops placed at the limits of the ROM. A video of one of the participants performing the discrete position-tracking task using the force-based control interface can be found in [8] as additional file 5.

For each interface the participant had 5 minutes to perform free movements, followed by 12 training trials and 36 evaluation trials. Each trial represented a single tracking task to a target angle. The target angles of the training and evaluation trials were ordered with the same number of long and short movements (i.e., 18 movements) and same number of ascending and descending movements.

For all participants, the free-movements and the training trials were performed using first the EMG-based interface and afterwards the force-based interface. Subsequently, the subject rested for 10 minutes and performed the evaluation trials first with the EMG-based interface, and secondly with the force-based interface, with a resting period of 10 minutes in between the evaluation of the two control interfaces.

F. Questionnaire

To evaluate the experience of the participants and the acceptance of the EMG- and the force-based control interfaces, each participant was asked to answer seven questions (see Table III) after completing the experimental protocol.

G. Data Analysis

Data analysis was performed on metrics derived from the elbow angle displacements as function of time while reaching towards the short and long targets. We measured 18 evaluation trials for each of the three subjects using EMG- and force-based control interfaces. The first two evaluation trials were removed from the analysis since they showed clear start-up effects, thus the dataset resulted in a total of 16 trials for each subject, control interface and target distance. The movement performance was evaluated in terms of task completion time (TCT) and rate (TCR), overshoot (OS), trajectory efficiency (TE), and smoothness (SM; see Table II for definitions). Additionally, to better understand the metric of task completion time, this metric was divided in two parts: rising time (RT) and settling time (ST). Short and long movements were analyzed separately since they present different index of difficulty [29]. The chosen performance descriptors are common measures used in studies that evaluate the performance of control interfaces [21]–[23].

Departure from normality and severe heteroscedasticity that was not alleviated by transformation were present in the distribution of all metrics, which prevented the use of multi-way repeated measures ANOVA to test for significance of the grouping by control system and subjects. We thus used a suite of nonparametric Friedman tests with a level of significance fixed at $p < 0.05$. A more detailed inspection of the distribution of all metrics by control system for each subject and target distance was performed through box plots combined with nonparametric Wilcoxon rank-sum tests (significance level: $p < 0.05$). In the box plots, the data points above or below 1.5 times the interquartile range (IQR) are shown as outliers with a "+" symbol. Furthermore, we explored the distribution of trials in all bivariate

TABLE II
PERFORMANCE METRICS

Performance metric	Description
1. Rising Time (s):	Time needed to move the elbow from the start position to the target area.
2. Settling Time (s):	Time needed to stabilize the elbow from the first time that the elbow reaches the target area to the completion of the task.
3. Task Completion Time (s):	Total time needed to complete the task.
4. Task Completion Rate (%):	Number of trials completed over the total number of trials in function of time.
5. Overshoot (%):	Maximum normalized distance traveled outside of the target area during a trial.
6. Trajectory Efficiency (%):	Shortest distance to the target angle divided by the total distance traveled by the elbow during a trial.
7. Smoothness (NZC):	Number of times that the elbow angle changed direction i.e. number of zero crossings (NZC) of the velocity signal from the time that the elbow leaves the previous target area, and excluding the last two seconds of dwell time.

TABLE III
QUESTIONS AND PREFERENCES OF PARTICIPANTS

	EMG Preference	Force Preference	No Preference
1. Which interface could control the Elbow Drive most accurately?	S1	S2 S3	
2. Which interface could control the Elbow Drive fastest?	S1 S2	S3	
3. Which interface was the easiest to control (did react best to your motion intention)?	S1	S2 S3	
4. Which interface was least fatiguing to use?	S1 S2 S3		
5. Which interface was the easiest to set up/install?		S1 S2	S3
6. Which interface was the most comfortable to use?	S2		S1 S3
7. Which interface has your overall preference for controlling the Elbow Drive?	S1	S2 S3	

planes among all metrics in search of patterns related to control interfaces and subjects. The statistical tests were performed with R software (R Development Core Team 2015).

III. RESULTS

Fig. 5 presents the average and ± 1 SD of the elbow angle displacements and the task completion rates as function of time using the EMG- and the force-based control interfaces of Subjects S1, S2, and S3. While the movements with EMG-based control were considerably slower in Subject S1 and less smooth in Subject S3, subjects present similar movements with force-based control.

A. Task Completion Rate and Time

Average task completion rate of all trials at 15 s was higher for force-based control (100%) than for EMG-based control (95.8 $\pm$ 5.1%). Actually, a rate of 100% was already achieved at 9 s with force-based control, while 85.4 $\pm$ 19.6% was achieved with EMG-based control at the same time span (Fig. 5). For Subject 3, 35 s were necessary to achieve a 100% task completion rate with the EMG-based control due to two slow trials. Average task completion time of all trials was 6.8 $\pm$ 4.8 s and 5.1 $\pm$ 1.8 s for EMG- and force-based control respectively. For more details of the elbow angle displacements and task completion rates as function of time of each subject, control interface and target distance, see Fig. 9 in Supplementary Material.

B. Friendman Tests

Table IV in Supplementary Material presents the results of the significance of grouping trials by control interface and subjects for all metrics. Significance was much higher (p-value much lower) for grouping by subjects than by control systems in all metrics. The only metric that showed a significant difference among control methods was overshoot ($p = 0.0011$). Significant differences between subjects were found for all metrics ($p < 0.001$).

C. Box Plots and Wilcoxon Rank-Sum Tests

Subject S1 presented a significantly longer task completion time ($p = 0.0007$) for EMG-based control (7.6 $\pm$ 2.8 s) due to a longer rising time (5.2 $\pm$ 2.2 s, $p = 0.0002$) compared to force-based control (TCT: 4.9 $\pm$ 1.1 s, RT: 2.7 $\pm$ 1 s; Fig. 6 and Table V in Supplementary Material). For short targets, the trajectory efficiency was significantly higher ($p = 0.019$) in EMG-based control (89.7 $\pm$ 9.1%) than in force based-control (78.5 $\pm$ 13.5%; Fig. 7 and Table VI in Supplementary Material). Subject S2 showed significantly higher task completion time (EMG: 4.3 $\pm$ 0.8 s, force: 4.8 $\pm$ 0.7 s, $p = 0.039$) due to a significantly higher rising time (EMG: 1.9 $\pm$ 0.4 s, force: 2.5 $\pm$ 0.6 s, $p = 0.003$) for the movements towards the long targets when using force-based control compared to EMG-based control. Subject 2 also presented a significantly higher number of zero crossings for the long targets (EMG: 5.6 $\pm$ 1.6, force: 7.3 $\pm$ 0.7, $p = 0.0008$) when using force-based control. Subject S3 presented large significant differences ($p = 1.2 \cdot 10^{-6}$) in the smoothness of the movement. The number of zero crossings for this subject were nearly two-fold higher in EMG-based control (37.4 $\pm$ 2.6) than in force-based control (18.1 $\pm$ 0.9), which compromised trajectory efficiency (EMG: 67.2 $\pm$ 2.6%, force: 81 $\pm$ 11.1%, $p = 0.04$) and overshoot (EMG: 12.1 $\pm$ 6.2%, force: 2.9 $\pm$ 4.9%, $p = 0.0002$) of the movements when using EMG-based control. In Subject S3, differences in trajectory efficiency were larger for the movements towards the short targets ($p = 0.004$), for which EMG-based control presented a significantly higher task completion time than force-based control (EMG: 9.7 $\pm$ 4.1 s, force: 5.3 $\pm$ 1.5 s, $p = 0.0007$; Fig. 7).

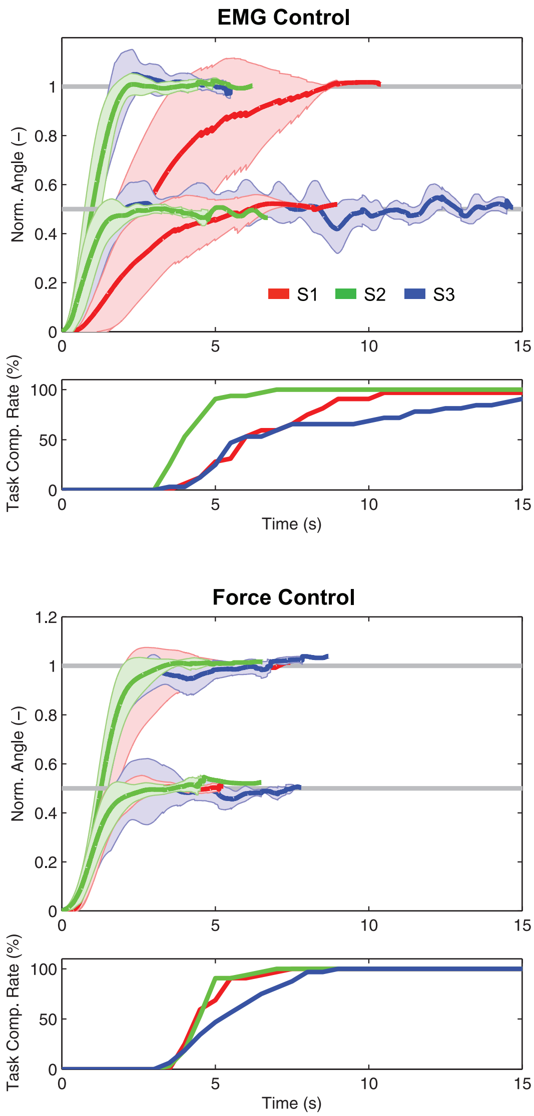

Fig. 5. Average (solid line) and ± 1 SD (area with faded color) of the elbow angle displacements, and task completion rate as function of time of the 32 evaluation trials of Subjects S1 (red), S2 (green) and S3 (blue) using the EMG- (top) and the force-based (bottom) control interfaces while reaching towards the long (norm. angle $= 1$) and short (norm. angle $= 0.5$) target angles (grey solid line). Three slow trials of Subject S3 [long targets: (1),(2); short targets:(4)] and one of Subject S1 [long targets: (3)] using EMG-based control have been excluded from this figure for clarity. The slow trials can be seen in Fig. 9 in Supplementary Material.

D. Bivariate Plots

Fig. 8 shows that trials cluster in three groups according to smoothness and that this grouping is mainly by subjects: trials corresponding to Subject S2 at one extreme, trials corresponding to Subject S3 with EMG-based control to the other extreme, and a third cluster in the middle including both the trials corresponding to Subject S1 and those corresponding to S3 with force-based control only. Smoothness is minimum (maximum number of zero crossings) for trials done by Subject S3 using EMG control and maximum (minimum number of zero crossings) for trials done by Subject S2, while trials done by Subject S1 are intermediate. It is worth noting that while trials done by Subject S3 with EMG-based control have a very low smoothness, those performed by the same subject with force-based control have similar levels of smoothness than those of Subject S1.

As a summary, the analysis of the performance metrics indicated that there is more, and more consistent variability among subjects than between control interfaces, with Subject S2 presenting a higher movement performance than Subjects S1 and S3. For Subjects S1 and S2 both control interfaces were equally performant, but the force-based control system clearly performed better for Subject S3.

E. Questionnaire

The results of the questionnaire are summarized in Table III. While Subjects S1 and S3 showed a clear preference for EMG- and force-based control respectively, Subject S2 did not. All subjects agreed that EMG-based control was least fatiguing. For the rest of the questions, the answers of the participants showed different preferences.

IV. Discussion

A. Feasibility, Performance and Acceptance

In the context of assistive devices, the performance descriptor that we consider most meaningful is the task completion rate with a reasonable time limit, which clearly captures the ability of subjects to perform the task. Additionally, descriptors that measure accuracy and efficiency of the movement, such as trajectory efficiency, overshoot and smoothness, become relevant to analyze and compare the movement performance between control methods in more detail. Results of questionnaires provide additional insights on usability and acceptance of the technology.

Despite the fact that all three participants had not been able to perform any voluntary movements with their arms for the past 3–5 years, all of them were able to perform 100% of the series of tracking tasks with both control modalities (mean task completion time EMG: 6.8 ± 4.8 s, force: 5.1 ± 1.8 s). These results indicate that EMG- and force-based interfaces are feasible solutions for the control of active elbow supports in adults with DMD. While healthy individuals perform the tabletop to mouth movement within a second, the task completion times achieved by the participants of this study were considerably longer. We expect that the task completion time of adults with DMD would decrease with training and make this technology acceptable.

All participants were capable of continuously controlling the amplitude of their EMG and force signals to perform the series of discrete position-tracking tasks with a high overall movement performance. Even though the EMG and force signals of the participants presented a very low amplitude (see MVIC values in Table I; compared to signals of healthy individuals [19]), they were measurable with standard signal acquisition equipment, and contained meaningful information of their motion intention. In a recent uncontrolled exploration, we detected that a 37-year-old man with DMD was still able to generate usable EMG signals with his biceps and triceps muscles.

The relative performance of the control interfaces depended on the subject (Table IV in Supplementary Material). For Subject S2, the differences of performance between both methods were small for all metrics, small but statistically significant for task completion time and smoothness, and even not significant

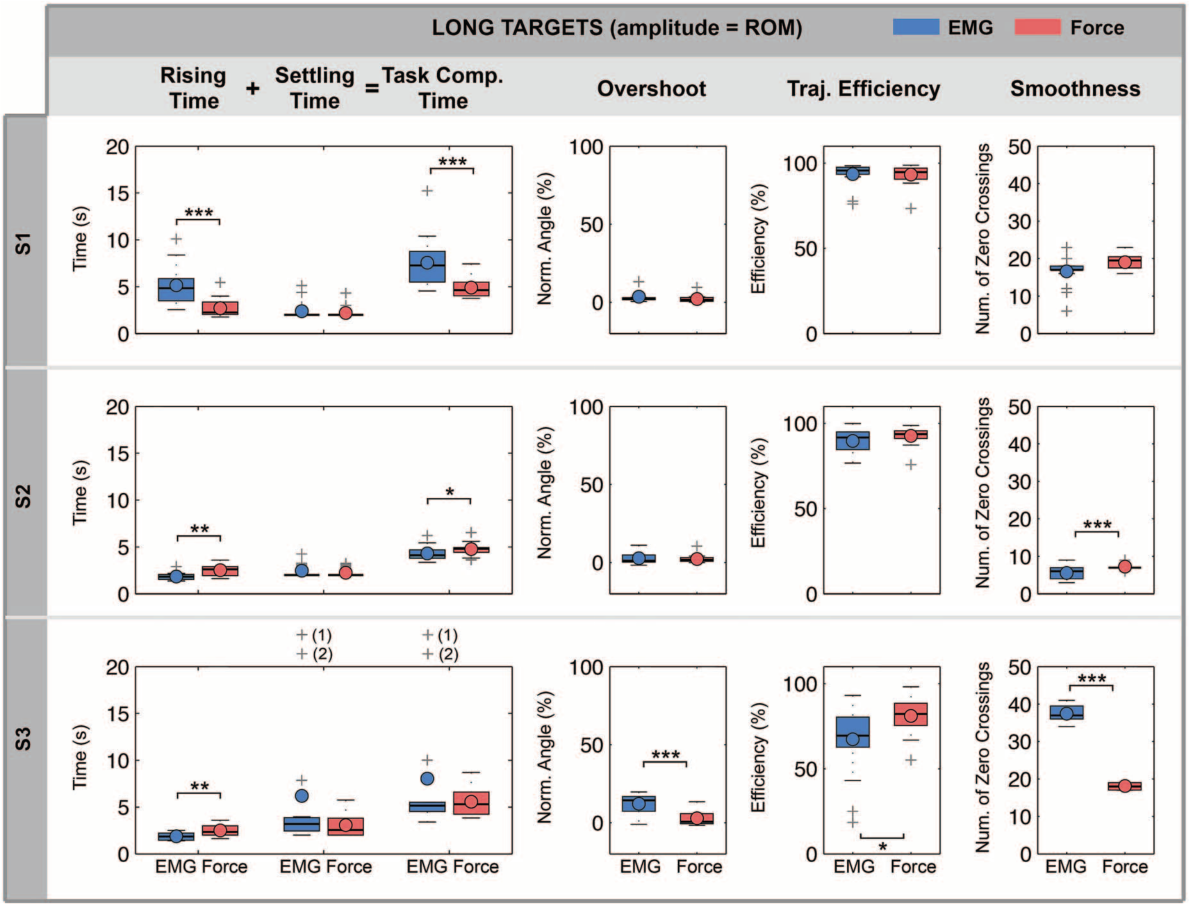

Fig. 6. Boxplots for each movement performance metric of Subjects S1, S2 and S3 using the EMG- (blue) and the force-based (red) control interfaces during the tracking of the long target. Outlier (1) has the value of 32 s and outlier (2) of 25 s. (*) indicates $p < 0.05$. (**) indicates $p < 0.01$. (***) indicates $p < 0.001$.

for trajectory efficiency and overshoot. The movements of Subject S1 were considerably faster with the force-based interface, while the rest of performance metrics presented similar results for both control interfaces. Finally, all metrics (especially smoothness) clearly indicated that Subject S3 performed better with force-based control than with EMG-based control (Tables V and VI in Supplementary Material).

The large performance differences found between the control methods in Subject S3 resulted from the lower signal-to-noise ratio of the EMG compared to the other Subjects (S3 E_{mvic_t}: 0.001 mV, S1 E_{mvic_t}: 0.004 mV). The envelope of the EMG signal while resting (i.e., noise level) presented maximum peaks of 0.0007 mV for all three subjects, which clearly affected the EMG-based control of Subject S3 since the amplitude of the signal noise was equivalent to 70% of his E_{mvic_t}. Jerkiness of the movements performed when using the EMG-based control interface could have been improved by fine-tuning the system for Subject S3: modifying the cutoff frequency of the filter used to detect the EMG envelope, or increasing the mass constant of the interface dynamics (H_{id}). We decided not to manipulate the parameters of the interface dynamics during the experiment in order to keep the experimental conditions consistent among participants.

The results of the questionnaire showed mixed preferences between participants which indicates that both EMG- and force-based control were perceived as possible solutions for the control of active arm supports.

B. Fatigue and Joint Stiffness

All three participants indicated in the questionnaire (Table III question 4) that EMG-based control was less fatiguing than force control. One fundamental difference between EMG and force-based control is that in the latter the measured force signal also contains static and dynamic components (i.e., intrinsic forces) of the arm such as stiffness, viscosity, inertia and gravitational forces. Distinguishing the voluntary component of the force signal requires proper estimation of the other components. Gravity and joint stiffness forces are challenging to estimate accurately due to their nonlinearity, time-variance and pose dependency [30]–[32]. Additionally, the low voluntary forces of adult men with DMD, which are several orders of magnitude smaller than their arm's weight, and the increase of joint stiffness due to disuse of the arms [33] make the distinction of the voluntary forces even more challenging. The higher fatigue levels experienced by the participants when using the force-based control interface might be due to small errors on the estimation of the gravity and joint stiffness compensation force.

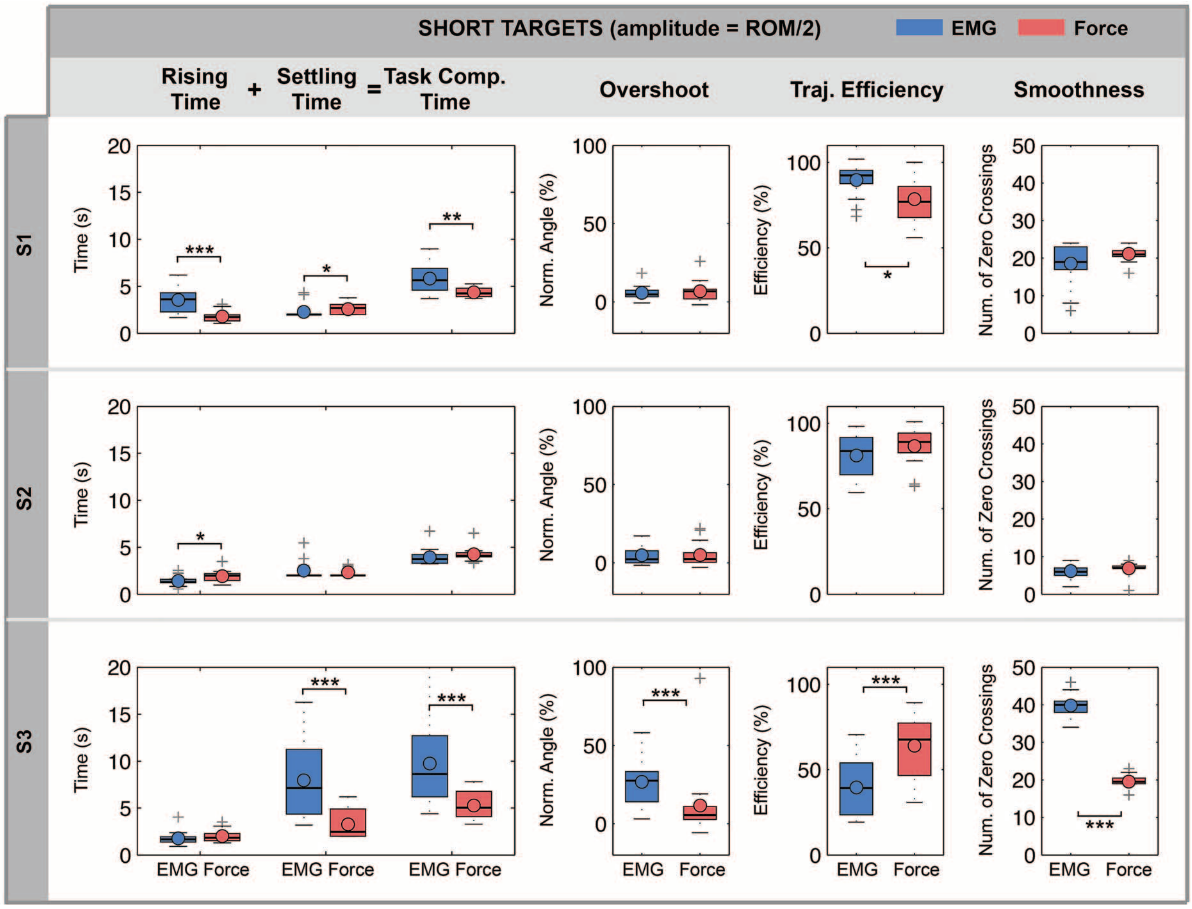

Fig. 7. Boxplots for each movement performance metric of Subjects S1, S2 and S3 using the EMG- (blue) and the force-based (red) control interfaces during the tracking of the short target. (*) indicates $p < 0.05$. (**) indicates $p < 0.01$. (***) indicates $p < 0.001$.

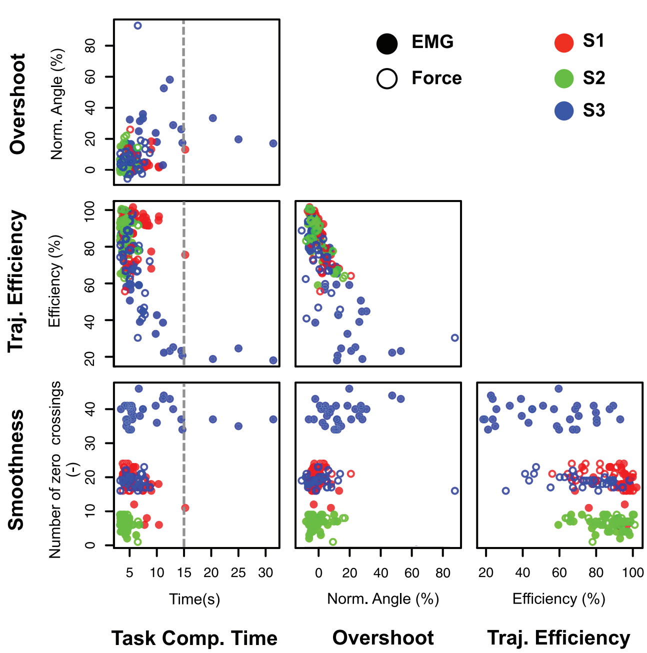

Fig. 8. Distribution of trials in all bivariate planes among all metrics: performance metrics task completion time, overshoot, trajectory efficiency and smoothness. All 16 trials from long and short targets of Subjects S1 (red), S2 (green) and S3 (blue) using EMG- (full circle) and force-based (empty circle) control are included. The vertical dashed line indicates the time limit of 15 s used to distinguish the slow trials.

C. Force- Versus EMG-Based Control

Taking into account the aforementioned difficulties of the force-based control interface to detect the voluntary forces of the user, we think that force-based control interfaces are appropriate for people that still have enough force to partially overcome the intrinsic forces of the arm. While EMG-based control might be less intuitive (especially for multi-DOF movements) than force-based control, since force is closer to the natural way of interacting with the environment, EMG signals are not affected by the intrinsic forces of the arm. Therefore we presume that EMG signals can better represent the movement intention of people with voluntary forces below the intrinsic forces of the arm. 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.

D. Participants and Experimental Protocol

Due to the low density of men with DMD [34] and the legal/ethical constraint that they can only participate in one study at the same time, the access to suitable subjects was limited. In the allowed time window of this study, we had access to three participants that met all criteria and were able and willing to participate. Since the main goal of this study to investigate the feasibility of EMG and force as control interfaces using a research platform, we performed the tests with three participants, which allowed for an exploratory assessment of the limitations and capabilities of the control interfaces. Therefore, our results indicate, but cannot demonstrate at population level, the feasibility of both control interfaces for the operation of active elbow supports for adults with DMD.

Adults with DMD experience strong training and fatigue effects due to the disuse of their arms. Therefore, the experimental protocol balanced the amount of training and evaluation trials: the protocol allowed the participants to learn how to control the Elbow Drive and perform the tracking task while having a considerable number of evaluation trials per condition (i.e., 36 trials) and keeping the overall length of the experiment below two hours to minimize fatigue. Due to the small number of participants included in this study and the high functional variability between adult men with DMD [5], randomizing the order of the control methods would not have been effective.

E. Implications

Several arm supports that compensate the weight of the arms are commercially available [6] and have been shown to increase the independence and quality of life for people with muscular weakness [35], [36]. However, in adults with DMD, the decrease of muscle force combined with an increase of passive joint-stiffness [16], [33], due to the disuse of the arms, reduces the effectiveness of passive arm supports [7]. Table I presents the range of motion reached by the three participants using only passive weight compensation in the Elbow Drive. Subject S1, which was the strongest, could achieve only 15 degrees of elbow flexion with his maximum effort. At the last stage of the disease people with DMD can benefit more from active arm supports, which can provide extra assistance.

Compared to the large number of active arm prosthetic devices available for amputees [10], few active devices have been developed for supporting upper-extremity function of people with severe muscular weakness [16], [37], [38]. This feasibility study is the first step towards the development of a control interface specially designed to control active arm supports for people with DMD.

The results show a higher variability between subjects than between control interfaces (Table IV in Supplementary Material), which indicates that it is likely that the control interfaces have to be customized for each individual with DMD. This customization should include signal-filtering parameters, interface dynamics, and sensor placement, but also the physiological signals (i.e., EMG, force or combination) used to derive the motion intention of the user.

F. First One-DOF Control

Upper-extremity assistive devices that support the performance of activities of daily living require multiple-DOF. It is part of our continued investigations to determine how many DOF are needed and what is the most suitable way to control them. There are no studies that evaluated the feasibility of EMG and force as control interfaces in adults with DMD. Hence, our strategy was to carry out an evaluation with a one-DOF active elbow support to focus on the control capabilities of men with DMD before integrating the complexity of a multi-DOF assistive device.

G. Extension to Multi-DOF Control

The extension of the control interfaces to a device with multiple DOF will increase their complexity, which might lead to clear performance differences between the two control interfaces. While the calibration-based method proved to be effective for the distinction of the user's voluntary force to control the elbow movement, it might become too cumbersome for the compensation of gravity and joint stiffness in a multi-DOF arm support where the arm can acquire a large variety of poses. We have performed preliminary measurements of joint stiffness forces in the horizontal plane [39] that can be used for their compensation.

Several studies have shown simultaneous and proportional control of two-DOF in arm prosthesis by means of regression-based algorithms [12], [13] and three-DOF using intramuscular EMG electrodes [40]. Furthermore, there are studies that have correlated shoulder EMG to arm kinematics [41], [42]. We are currently working on the extension of the one-DOF proportional EMG-based control used in the Elbow Drive system, to multiple-DOF, and the implementation of regression-based control methods for arm prosthesis to active arm supports for people with DMD. Preliminary results of a proportional and simultaneous two-DOF EMG-based control interface for assisting planar movements in task-space, can be found in [43]. In [43], EMG signals from biceps/triceps were mapped to right/left translations of the hand and EMG signals of deltoid anterior/posterior were mapped to forward/backward translations of the hand.

V. Conclusion

This study presents a feasibility and performance evaluation of EMG- and force-based interfaces during a one DOF discrete position-tracking task with adults with DMD. All subjects were able to complete 100% of the series of tracking tasks with both control interfaces with an average task completion time of 6.8 $\pm$ 4.8 s and 5.1 $\pm$ 1.8 s for the EMG- and force-based control respectively. While movements with the force-based control were considerably smoother (difference between means: 19 zero crossings; $p < 0.001$) in Subject 3 and faster in Subject 1 (difference between means: 2.5 s; $p < 0.001$), EMG-based control was perceived to be less fatiguing by all three participants. In conclusion, we found important variability among subjects in terms of performance of EMG- and Force-based interfaces, and both methods are feasible for the control of one DOF active elbow supports in men with DMD with very limited arm function. Both EMG- and force-based interfaces should be further investigated for the control of multi-DOF active arm supports.

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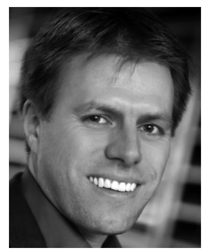
Prof. Veltink has been an Associate Editor for the IEEE TRANSACTIONS ON NEURAL SYSTEMS AND REHABILITATION ENGINEERING.



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When the Laboratory of Biomechanical Engineering was founded in 1990, he started working there as a Faculty Member. He worked on various topics related to the coordination of movement and with applications in rehabilitation, orthopedics and neurology. He now chairs the Biomechanical Engineering Group, which is embedded in the Institute for Biomedical Engineering (MIRA), University of Twente.