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Published in:
IEEE Transactions on Neural Systems and Rehabilitation Engineering

DOI:
10.1109/TNSRE.2018.2866956

Evaluation of Control Interfaces for Active Trunk Support

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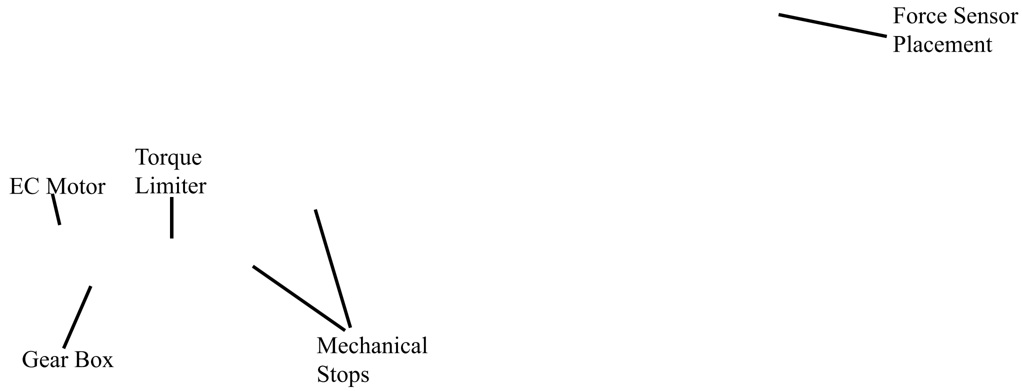
***Abstract*— A feasibility study was performed to evaluate the control interfaces for a novel trunk support assistive device (Trunk Drive), namely, joystick, force on sternum, force on feet, and electromyography (EMG) to be used by adult men with Duchene muscular dystrophy. The objective of this paper was to evaluate the performance of the different control interfaces during a discrete position tracking task. We built a one degree of freedom flexion–extension active trunk support device that was tested on 10 healthy men. An experiment, based on the Fitts law, was conducted, whereby subjects were asked to steer a cursor representing the angle of the Trunk Drive into a target that was shown on a graphical user interface, using the above-mentioned control interfaces. The users could operate the Trunk Drive via each of the control interfaces. In general, the joystick and force on sternum were the fastest in movement time (more than 40%) without any significant difference between them, but there was a significant difference between force**


Force Sensor
Placement
Torque
Limiter
EC Motor
Gear Box
Mechanical
Stops

(a)

connected to the computer and sent the calculated voltage to the motor driver (ESCON 90/50) through the same data acquisition card which then provided the appropriate current to the motor (Maxon EC 90 brushless).

D. Signal Processing

The real-time signal processing was adapted from Lobo-Prat *et al.* [10]. The joystick was a simple potentiometer with a resistance of 1k and 5V-feed from the data acquisition card. The following equation describes the joystick's signal processing::

$$J_{vol}(i) = J_{sen}(i) - J_{res} \tag{1}$$

$$u_{joy}(i) = \begin{cases} \dfrac{J_{vol}(i)}{J_{mvi,f}}, & \text{if } J_{vol}(i) < 0 \\ \dfrac{J_{vol}(i)}{J_{mvi,e}}, & \text{if } J_{vol}(i) > 0. \end{cases} \tag{2}$$

J_{sen} is the sensed intention of movement, J_{res} is the average signal amplitude of a resting period of two seconds, J_{vol} is the voluntary movement and u_{joy} is the joystick control signal. $J_{mvi,f}$ and $J_{mvi,e}$ are the maximum voluntary inclination (MVI) of the joystick when pushing the joystick forward (flexion of trunk) and pushing the joystick backwards (extension of trunk) for two seconds, respectively. Finally, the (i) represents the i^{th} time step of the signal.

Regarding the force-based control interface on the sternum, it is necessary to distinguish the voluntary forces of the user from external forces such as gravity or joint stiffness and to compensate for these. The external force is obtained before the actual measurement by measuring the sternum interface forces during a slow flexion of the trunk (0.05 rad/s) while the subject is fully relax. The compensated force $F_{com}(\theta)$ is a function of the trunk angle (θ) and it is subtracted from the sensed forced $F_{sen}(i)$ to realize the intended movement as is shown in the following equations:

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

$$u_{force}(i,\theta) = \begin{cases} \dfrac{F_{vol}(i,\theta)}{F_{mvc,f}}, & \text{if } F_{vol}(i) < 0 \\ \dfrac{F_{vol}(i,\theta)}{F_{mvc,e}}, & \text{if } F_{vol}(i) > 0. \end{cases} \tag{4}$$

Where $u_{force}(i,\theta)$ is the force control signal, $F_{mvc,f}$ and $F_{mvc,e}$ are the two seconds abdominal (flexion) and iliocostalis (extension) maximum voluntary contraction (MVC), F_{res} is the average signal amplitude of a resting period of 2 seconds and J_{vol} is the voluntary movement. The force plate signal processing can be described as:

$$FP_{vol,}(i) = FP_{sen}(i) - FP_{res} \tag{5}$$

$$u_{forceplate}(i) = \begin{cases} \dfrac{FP_{vol}(i)}{FP_{mvi,f}}, & FP_{vol}(i) < 0 \\ \dfrac{FP_{vol}(i)}{FP_{mvi,e}}, & FP_{vol}(i) > 0. \end{cases} \tag{6}$$

Envelope detection was applied to the raw EMG signal with a high-pass Butterworth filter at 40 Hz, a full wave rectifier and a low pass Butterworth filter at 2 Hz [16]. Furthermore, additional signal processing was performed to normalize the two signals from the agonist and antagonist muscles. In the following equation, $E_{nor,k}(i)$ represents the normalized EMG and $u_{emg}(i)$ the control signal:

TABLE I
ADMITTANCE CONTROL PARAMETERS

Control Interface	Virtual Mass	Virtual Damping
joystick	0.3	1
force on sternum	0.1	0.5
force on feet	0.3	1.5
EMG	0.5	4

$$E_{nor,k}(i) = \frac{E_{env,k}(i) - E_{res,k}}{E_{mvic,k}} \tag{7}$$

$$u_{emg}(i) = E_{nor,t}(i) - E_{nor,g}(i) \tag{8}$$

Subscript k represents the abbreviation of the tibialis (t) and gastrocnemius (g) muscles. $E_{env,k}$ is the envelope of the raw EMG, $E_{res,k}$ is the average of the signal amplitude in a rest period of two seconds, E_{mvc_k} is the maximum value during 2 seconds of maximum voluntary isometric contraction.

E. Control

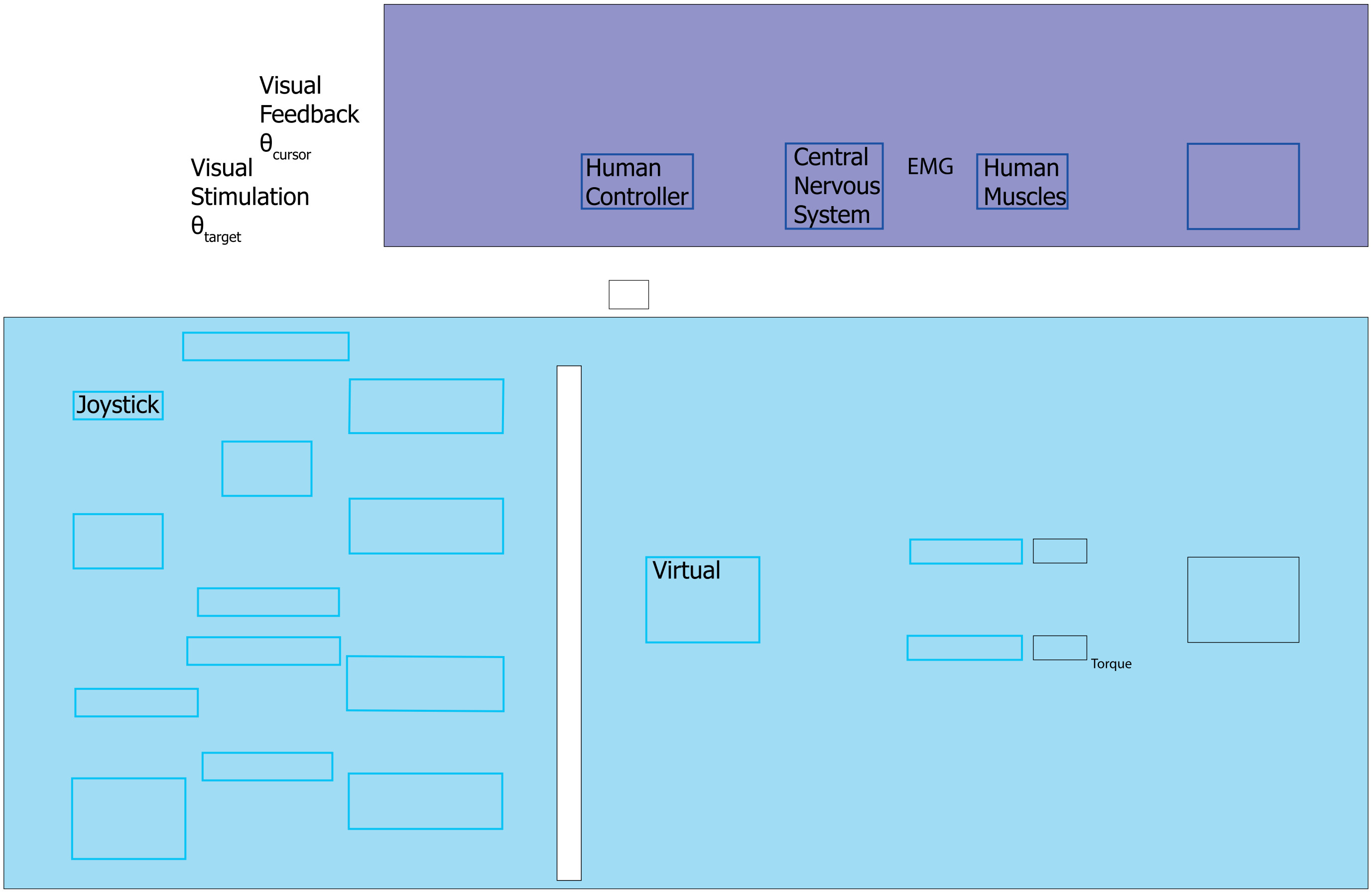
The control architecture consists of two levels: higher and lower level control. The higher level control is a second order admittance model with virtual mass A and virtual damping B where A, B were tuned with the different control interfaces in order to achieve low movement time and low overshoot. The values were kept constant between the subjects and they were chosen in such a way that the time to complete a task and the overshoot were minimized in a pre-experimental procedure with 2 subjects that were not included in the experiment. The admittance model generates the reference position from the intention detection signal. The position reference is followed by the lower level control. The lower level controls the position of the two motors using one PD controller for each motor. The PD values of the controller were tuned manually (Fig. 3, Table I).

III. EXPERIMENTAL DESIGN

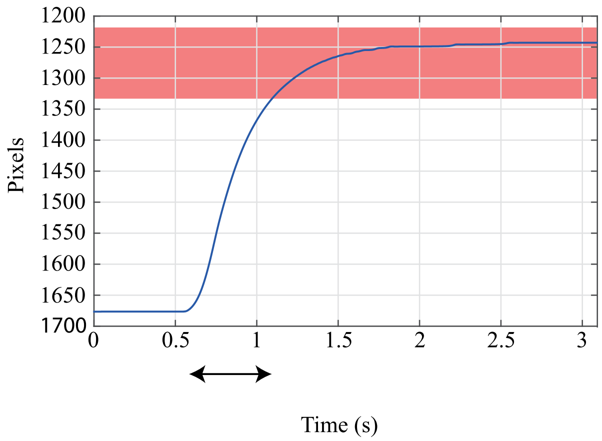
The experiment was based on the approach described by Fitts [17], who identified a predictive model describing human speed accuracy trade off in a tapping task. A one-dimension serial position-tracking task was presented to the subject by means of Python custom-made Graphical User Interface (GUI) on a 1050 Ã— 1680 pixels display.

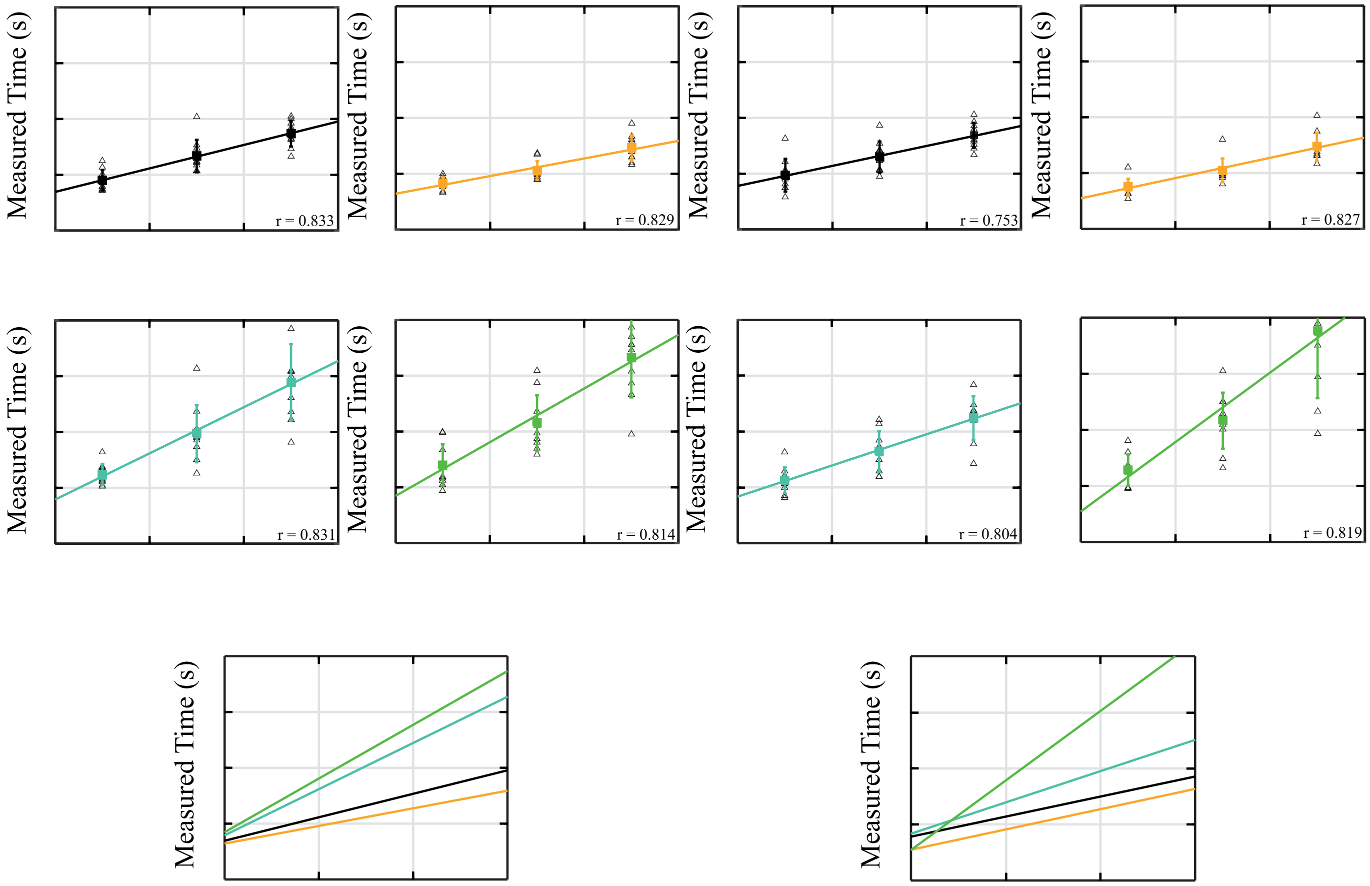
The user has to steer a cursor (yellow circle in Fig. 4) from a home position (blue circle) to a target (red circle). The cursor was coupled with the encoders of the left motor. The Index of Difficulty (ID, 9) from Shannon's form [18] was used to characterize targets. A target was presented at one of the three different locations at a distance of 395, 791 and 1583 pixels from the cursor starting position. The target radius remained constant at 70 pixels plus 30 pixels for cursor correction [19]. The GUI is shown in Fig. 4.

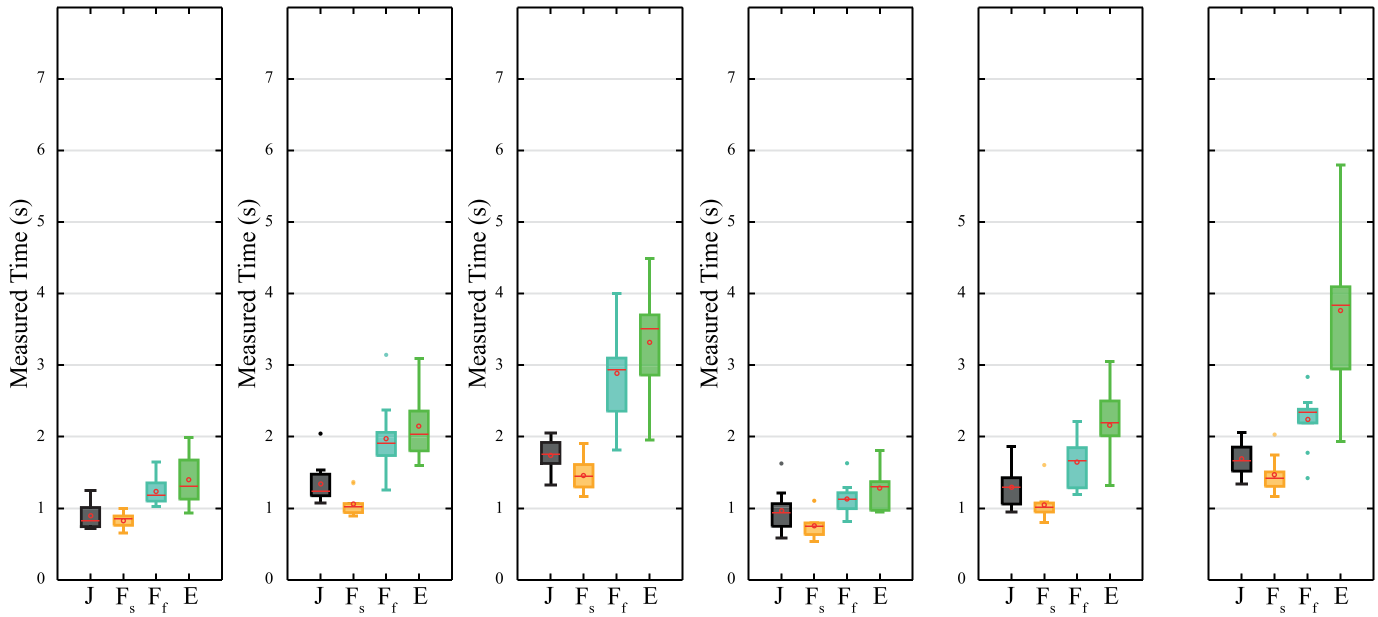
$$ID = \log_2\left(\frac{D}{R} + 1\right) \tag{9}$$

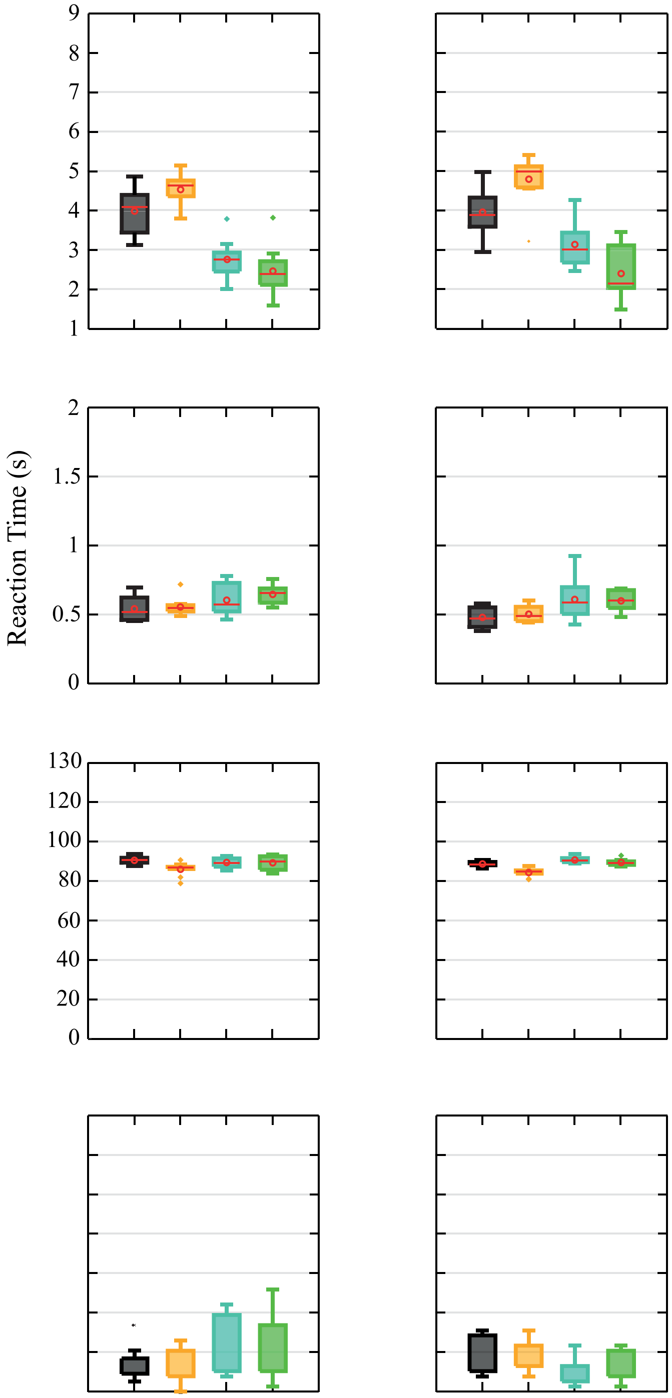


Visual
Feedback
θ_{cursor}
Visual
Stimulation
θ_{target}
Human
Controller
Central
Nervous
System
EMG
Human
Muscles
Joystick
Virtual
Torque


1200
1250
1300
1350
1400
1450
1500
1550
1600
1650
1700
Pixels
0
0.5
1
1.5
2
2.5
3
Time (s)


Measured Time (s)
r = 0.833
Measured Time (s)
r = 0.829
Measured Time (s)
r = 0.753
Measured Time (s)
r = 0.827
Measured Time (s)
r = 0.831
Measured Time (s)
r = 0.814
Measured Time (s)
r = 0.804
r = 0.819
Measured Time (s)
Measured Time (s)


Measured Time (s)
0
1
2
3
4
5
6
7
J
F_s
F_f
E



Reaction Time (s)
9
8
7
6
5
4
3
2
1
2
1.5
1
0.5
0
130
120
100
80
60
40
20
0

C. Overall Performance

Based on the results of the MT, we can conclude that force on feet and EMG were slower than force on sternum and joystick for all IDs indicating that the subjects could achieve faster flexion and extension movements (attaining more than 70% in some cases). The MT results are in agreement with the performance of throughput where force on sternum and joystick have a larger transfer rate than force on feet and EMG. This result can be explained by the fact that the motion of the trunk can be controlled more intuitively by the joystick and the force on sternum. In contrast, force on feet and EMG from the leg muscle are considered to be less intuitive. The RT, PE differences were small (< 0.2 seconds, 4% respectively) which indicates only a marginal performance difference. Finally, there were no significant differences for the OS.

According to the subjects' responses to the questionnaire, the joystick was the easiest to control and the more accurate which contradicts with the results of the PE and OS metrics. The force on sternum was the fastest which is in line with the experimental results. On the other hand, the subjects found the force on sternum the most exhausting compared to the other 3 due to the fact that force signal contains both dynamic and static components of the upper body . Finally, the users' overall preference was the joystick which is not surprising because it is a very common control interface in, for example, video games.

D. Control Interfaces

The joystick was the first control interface to be used to allow the subjects to familiarize themselves with the system's dynamics. Hand- joysticks are commonly used as control interfaces for assistive devices (e.g., electrical wheelchairs) by individuals with muscular weakness and this was considered the easiest and most straightforward method to control an assistive device. Indeed, the joystick performed similarly to force on sternum although it was always the first control interface. The major drawback of using a joystick is that one hand's function is sacrificed whenever a person intents to move the trunk since it is a parallel system.

The force on sternum control interface was considered the most intuitive one. However, gravity compensation complicates matters as it acts on the upper body. Participants had difficulties in fully relaxing their muscles, which is important to achieve proper gravity compensation. Additionally, the interface was sensitive to respiration, which resulted in small oscillatory movements during the dwell time when the subjects were trying to keep their trunk at a certain angle.

The force on feet control interface was slower than the one using a sensor at the sternum, but gravity compensation was not necessary and the placement of the sensor not very critical. Although it is a parallel system requiring slight movements of the feet and it is slightly slower than the other control interfaces, it may be a solution for people who are seated in a wheelchair, because using the feet to control trunk movement is not a functional sacrifice.

EMG control performance was strongly dependent on where the sensor was placed on the muscles. It was easier to get a signal from the tibialis muscle to control the device. It was more difficult to find the right sensor location on the gastrocnemius so that the subject could control the device with ease. It should be mentioned though that at the beginning, the subjects needed some familiarization time to get used to the fact that they had to move their trunk by contracting their legs muscles. However, all the subjects were able to adapt to that procedure within minutes.

E. One-DoF vs Multi-DoF

The Peeters et-al study showed that the contribution of the trunk to achieve flexion tasks is divided equally between different segments (pelvis, lower lumbar, upper lumbar, lower thoracic, upper thoracic) [21]. As a consequence, an active trunk support should also be able to provide multi-DoF support, resembling natural reaching movements. Since no active trunk assistive devices exist, we decided to investigate the control capabilities of a relatively simple system before investigating a more complex assistive device. The DoF has to be increased for lateral bending as the current design restricts it.

An increase in DoF will introduce complexity not only to the mechanical design but also to the control. The mentioned control interfaces would have performed differently if the complexity of the control task had been increased.

F. Limitations

The first limitation is the setting of the admittance values which were tuned based on the performance on a pre-experiment with 2 subjects (not included in the results). The purpose of this pre-experiment was to find the optimal values of virtual mass and damping for each of the control interfaces. Even though the values of the virtual masses are close, the values of the virtual damping differ noticeably. This is due to the nature of the input signal in the admittance model. Control interfaces such as EMG need a bigger virtual damping value to attenuate the high input in the admittance model. Reducing the virtual damping value would result in a higher overshoot.

Second, the gear ratio backlash and the electronics of the motor resulted in mechanical play, giving 0 to 20 pixels in the GUI. Thus, not all the trials started at 0 or 1680 pixels (starting points for flexion end extension respectively), even though the subjects were asked to move the device to the mechanical end stops. We did not compensate for this non equality between trials since we considered it to be negligible.

VI. Conclusion

We investigated the performance of four different control interfaces on an experimental active trunk support device. The force on sternum and the joystick control interfaces were faster than the ones based on force underneath the feet and EMG. Regarding path efficiency, overshoot and reaction time, significant differences were found between the control interfaces but the differences in absolute values are negligible. Force on sternum was experienced as the most fatiguing interface by the participants, and they preferred the joystick. From the above results, we can conclude that all four control interfaces can be potentially used to control an active trunk support with different advantages and disadvantages. Further research on the performance of the control interfaces will be done with people with DMD.

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