Using Upper-Body Motions to Control Power Wheelchairs for Individuals with Tetraplegia

Elias B. Thorp[1,2], Farnaz Abdollahi[2], David Chen[3], Ali Farshchiansadegh[1,2], Mei-Hua Lee[2,4], Jessica Pedersen[3], Camilla Pierella[5,6], Elliot J. Roth[3,5], Ismael Seanez Gonzalez[1,2], and Ferdinando A. Mussa-Ivaldi[1,2,5] *Member, IEEE*

***Abstract*— Current power wheelchair control interfaces often do not provide adequately customized control for severely paralyzed individuals. These interfaces frequently either do not provide a full vocabulary of control commands, or are not designed to take advantage of any residual movement that remains available to the user. Here, we have developed a wheelchair control interface that relies on small shoulder movements for individuals who have high level cervical spinal cord injuries. By manipulating the kinematics of their shoulders, users can continuously control both the speed and direction of a power wheelchair. A pilot study of three able-bodied individuals and three individuals with spinal cord injuries revealed that users are able to learn to control the power wheelchair accurately and safely. Joystick control was superior to our interface in terms of time to completion and smoothness. This is likely due to the greater familiarity of our subjects with the joystick controller. However, using the body machine interface, subjects moved along paths that were similar to those obtained using a joystick and all users were able to complete the all of maneuvers. This proves the efficacy of the developed system as an alternative power wheelchair control scheme for individuals with little to no hand and arm movement.**

I. INTRODUCTION

High-level spinal cord injuries (SCIs), specifically injuries to the cervical spinal cord, result in motor deficits including weakness and uncoordinated movements. Despite this, many individuals retain some movement ability, allowing for control of assistive devices such as power wheelchairs. The majority of power wheelchair users rely on a hand-controlled joystick. Some SCI survivors, however, do not have sufficient arm and hand control or coordination to manually operate a joystick. The organization of the spinal cord is such that the motor neurons that innervate the proximal muscles of the upper body (neck and shoulders) exit the spinal cord at a higher level than the motor neurons that innervate more distal muscles (arms and hands). From this, it is common for individuals who have severely limited arm and hand coordination to maintain significant shoulder mobility.

There exist a number of commercially available alternative wheelchair control systems that do not rely on coordinated arm or hand movements, the most prevalent being the sip-and-puff and head array. While these systems can provide a means of power wheelchair control for individuals unable to use a joystick, they are often not intuitive or can be difficult to use. The major drawback of such systems is that they only provide the user with a discrete set of commands, severely limiting the possible maneuvers the user can achieve. In a recent survey, nearly 50% of power wheelchair users reported difficulties or an inability to perform maneuvers using their current control interface [1]. Moreover, these systems require the user to conform to the device, and do not provide individualized control that could increase usability and maximize efficient maneuverability.

There has been recent progress in the field of non-invasive brain machine interfaces as a means for power wheelchair control. Specifically, there are numerous electroencephalography or electromyography based wheelchair controllers [2 - 7]. The advantage of such systems is that they do not require any residual body motion, as they rely solely on neural activity to generate control commands. This type of controller would be ideal for extreme cases of paralysis such as locked-in syndrome or progressive conditions such as amyotrophic lateral sclerosis. However, similar to the commercially available alternative wheelchair controllers, many of these systems do not provide the user with proportional control. The user can specify the direction of motion, typically in one of the four cardinal directions, but has no control over the speed of the wheelchair [3], resulting in a more limited set of possible maneuvers. Also, these systems can have long latency periods due to the amount of time needed to classify brain or muscle activity, which is inherently noisy. Specifically, information transfer rates of non-invasive brain computer interfaces are usually below 0.5 bit/s [8, 9], while recent studies have shown that body movements operate on a much faster time scale [10]. While there has been progress in shared control algorithms that allow for a more continuous and complete vocabulary of commands [4] users may desire to have complete control of the wheelchair and not have to rely on the consistency of external sensors.

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[1] Department of Biomedical Engineering, Northwestern University, Evanston, IL 60208

[2] Sensory Motor Performance Program, Rehabilitation Institute of Chicago, Chicago, IL 60611

[3] Rehabilitation Institute of Chicago, 345 E. Superior Street, 1146, Chicago, Illinois 60611

[4] Department of Kinesiology, Michigan State University, East Lansing, MI 48864

[5] Department of Physiology, Physical Medicine and Rehabilitation, Northwestern University, Evanston, IL 60208

[6] Department of Informatics, Bioengineering, Robotics, and Systems Engineering at the University of Genoa, Genoa, Italy

(phone: 312-238-2696; email: eliasthorp@u.northwestern.edu, fabdollahi@ric.org, dchen@ric.org a-farshchiansadegh@northwestern.edu, mhlee@msu.edu, jpedersen@ric.org, camilla.pierella@northwestern.edu, ejr@northwestern.edu, i-seanez@u.northwestern.edu, sandro@northwestern.edu).

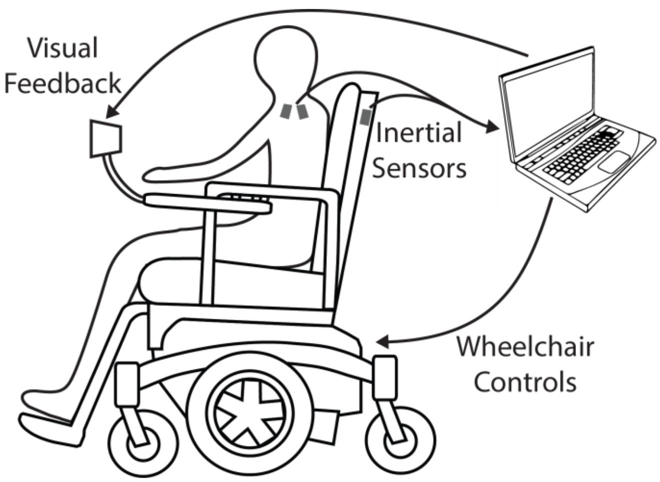

Figure 1. Diagram of the Body Machine Interface. Shoulder kinematics are recorded by inertial sensors, processed by an on board computer and commands are sent to the power wheelchair.

To overcome some of the challenges facing alternative wheelchair controllers for individuals with SCIs, leveraging residual body motion to generate control commands could provide a faster and more complete method to control a power wheelchair. It has been shown that individuals with SCIs can learn to control a two dimensional cursor by manipulating shoulder kinematics [10, 11]. This approach can also add rehabilitative benefits to power wheelchair control in that users practice engaging shoulder muscles, which may lead to increased functional ability of the upper body. Here we use dimensionality reduction techniques to map shoulder kinematics measured by inertial measurement units (IMUs) into a two dimensional control signal, which allows individuals with no arm or hand control to accurately and continuously control both the speed and direction of a power wheelchair.

II. METHODS

A. Body Machine Interface

The underlying principle for the Body Machine Interface (BoMI) lies in mapping high dimensional shoulder motions into low dimensional control as described in [11]. The components of the specific wheelchair interface described in this paper are shown in Figure 1. To track shoulder kinematics, two IMUs (Xsens Technologies B.V., Netherlands) were placed on each shoulder of the user (four IMUs total). IMUs use tri-axis accelerometers and gyroscopes to estimate the pose of each sensor. In this setup, we continuously recorded changes in the roll and pitch of each sensor, which equates to rotations about two orthogonal axes perpendicular to the direction of gravity. This setup generated eight control signals (roll and pitch from 4 IMUs) that capture a wide range of shoulder movements. The sensor configuration can be readily expanded by placing additional IMUs over different body locations.

The eight dimensional control commands were then mapped onto a two-dimensional task space vector. This was achieved using principal component analysis (PCA) [11]. PCA decomposed high dimensional shoulder movements into orthogonal components that best describe the variability of the data. This allowed us to create a customized interface that is tailored to the user's residual motion. During initial calibration, users performed a "dance" in which they were instructed to move their shoulders in as varied a motion as possible, while avoiding large movements. PCA was then performed on the collected IMU data. The two principal components (PCs) that account for the highest percentage of variance, were used to define a map (A). The control commands (p) were then calculated by

$$p = \begin{bmatrix} p_1 \\ p_2 \end{bmatrix} = \begin{bmatrix} a_{11} & a_{12} & \dots & a_{18} \\ a_{21} & a_{22} & \dots & a_{28} \end{bmatrix} \begin{bmatrix} h_1 \\ h_2 \\ \vdots \\ h_8 \end{bmatrix} = A\,h, \quad (1)$$

where A is the map defined by the PCA that transforms high dimensional shoulder configurations (h) into control commands (p). To achieve control of this interface, users must learn an inverse mapping from low dimensional control command, up to a high dimensional shoulder configuration. Specifically, users must learn an appropriate B (a right inverse of A) where

$$Bp = h \quad (2)$$

$$AB = I_2. \quad (3)$$

By this method, we were able to map an 8 dimensional vector of IMU measurements (h) down to a two-dimensional output vector (p). The mean posture during the calibration dance was set to be the origin of the output vector. This ensured that the user would be able to move equally in all directions and that the user could achieve the zero of the output vector by moving to a comfortable configuration. In this arrangement, the BoMI could be used to control any 2-degree of freedom device including a wheelchair or computer cursor. The dimensionality reduction calculations described

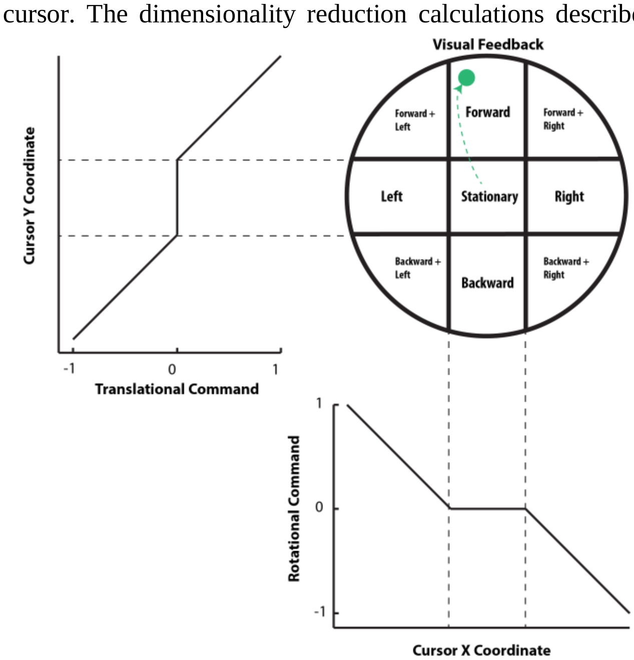

Figure 2. The top right shows the visual feedback display where the green cursor reflects the current state of the two control commands (reflected by the two plots). The green dashed line is for demonstration only and shows a potential path of the cursor from the mean posture. The two plots show how the cursor coordinates reflect both the rotational (x-axis) and translational (y-axis) command signals.

above were done using custom software designed using Matlab (Mathworks, Inc. Natick, MA)

To control a powered wheelchair, the output vector was proportionally mapped to signal voltages used to set the wheelchair velocity. Power wheelchairs are externally driven by providing two independent control commands [13], the translational command that specifies the forward speed (both wheels rotate with the same angular velocity), and the rotational command that specifies the angular velocity with respect to the vertical axis (the wheels rotate with opposite angular velocities). For simplicity, in the rest of the paper a control command of 1 equates to maximal forward or left rotation and a control command of -1 equates to maximal reverse or right rotation by the wheelchair. A command of 0 specifies no motion. Additionally, speed increases or decreases proportionally as the control command deviates from 0. By allowing the user to specify the full range of translational and rotational commands, a full set of wheelchair maneuvers can be achieved. In our system, the translational command was specified by the second value of the output vector (p_2) and the rotational command is specified by the first value of the output vector (p_1). To drive the wheelchair, we generated the necessary signal voltages by using a Phidget Analog 4-Ouput device (Phidgets, Inc., Alberta, CA). This device communicates with a computer via USB and outputs up to four independent analog voltages with a range of ±10 V and a resolution of 4.8 mV.

To help the users achieve more accurate maneuvers, real time visual feedback was displayed over a small monitor placed in front of them. Figure 2 depicts the visual feedback system. A small screen was mounted on the wheelchair that displayed a cursor within a circle. The location of the cursor in the circle represented state of the output vector (p), with the first component (p_1) mapped to the x-axis of the screen and the second component (p_2) mapped to the y-axis. Thus, the forward speed of the wheelchair was proportional to the displacement away from the center of the screen in the y direction and turning speed was proportional to the deviation in the x direction. The visual feedback was analogous to how the tip of a joystick would be deflected to make maneuvers. To facilitate driving, a dead zone was enforced that spanned roughly 15% of the maximum movement along each of the first two PCs. In other words, for both command signals independently, if the cursor deviated by less than 15% of the maximal movement from the mean posture, the wheelchair would not move. This ensured that small, unintended movements away from the mean posture did not cause the wheelchair to begin driving. It also made it easier to execute single direction movements, such as driving forward or turning in place. The remaining portions of the movements were linearly mapped to the output commands as can be seen in the plots for Figure 2.

B. Wheelchair Motion Compensation

There exists an inherent problem in using inertial sensors in a non-inertial frame of reference. Specifically, any accelerations of the wheelchair will cause unintended changes in the measurements obtained by the IMUs. By using only the roll and pitch (rotations about two orthogonal axis perpendicular to gravity) from the IMUs, standard wheelchair maneuvers on a flat surface have no effect on the control commands. These maneuvers only affect the yaw angle or rotations about the axis parallel to gravity. However, if the wheelchair user were to drive on a ramp or along any uneven surface, an offset would be present in the control signals, making it very difficult or perhaps even impossible to accurately control the wheelchair.

To account for this effect, a reference sensor was placed on the wheelchair to measure orientation changes that were a result of only wheelchair motion and not body motion. Because the IMUs measure orientation with respect to different reference frames, in order to cancel wheelchair motions a rotation matrix is needed to transform the measurements from the reference sensor into the reference frame of each of the other sensors. The sensors were set up so that all sensors shared a common z-axis pointing in the direction of gravity. Accordingly, the transformation between the reference frames for each sensor was simply a rotation about the shared z-axis. Since the controller will need to be robust to magnetic interference, the yaw angle (rotation about the z axis) was not used because it relies on measurements from tri-axis magnetometers that tend to drift or provide unreliable measurements in the presence of strong or changing magnetic fields. To measure this angle, we adopted the following procedure. First, for each sensor we defined a two-dimensional vector expressing the roll and pitch as measured by that sensor. Subsequently, the transformation angle between any two sensors was found. This angle was then used to calculate a rotation matrix to transform the measurements from one IMU into the reference frame for any other IMU.

$$m_i = [\phi_i \ \theta_i] \tag{4}$$

$$\psi_{i,j} = \mathrm{atan}[\frac{m_i \times m_j}{m_i \cdot m_j}] \tag{5}$$

$$R_{i,j} = \begin{bmatrix} \cos(\psi_{i,j}) & -\sin(\psi_{i,j}) \\ \sin(\psi_{i,j}) & \cos(\psi_{i,j}) \end{bmatrix} \tag{6}$$

Here m_i is the vector of measurements for a single sensor *i*, and $\psi_{i,j}$ is the transformation angle between sensor *i* and sensor *j*. $R_{i,j}$ is then the rotation matrix used to move from the reference frame of sensor *i* to the reference frame of sensor *j*. Using this method, the measurements from the reference sensor were projected into the frame of each other sensor and subtracted to cancel wheelchair motions.

C. Virtual Navigation

For safety reasons, participants performed simulated driving tasks in a virtual environment prior to driving the actual wheelchair. The virtual environment used in this study was a slightly modified version of the McGill Wheelchair simulator [14]. For the purposes of the current study, the simulator was adapted to use the output of the BoMI as the control commands for the virtual wheelchair. Custom environments were also developed to encourage both free exploration as well as to practice the specific tasks tested using the real wheelchair. Participants explored two environments. One environment was a replica of the 13th floor at the Rehabilitation Institute of Chicago and allowed the participant to maneuver freely in and out of rooms in real world scenarios. The other environment reproduced a series of task features that mirrored those that the participant would need to perform in the actual chair. These tasks were an abridged version of the Wheelchair Skills Test (version 4.1, http://www.wheelchairskillsprogram.ca) and are described further below.

D. Participants

Three people with SCIs participated in this study. The specific injury levels can be seen in Table 1. Generally, all participants sustained injuries that resulted in loss of fine motor control of their hands and arms and complete paralysis of the lower trunk and legs. All participants with SCIs were expert power wheelchair. Additionally, three able-bodied control participants also participated in the study. No subject had any prior experience with the BoMI described here prior to participating in the test. All subjects provided informed consent approved by the Northwestern University Institutional Review Board.

TABLE I.

	Injury Details		
	Injury Level	***Time Since Injury***	***Current Wheelchair Control Method***
S1	C6	2 years	Goal Post Joystick
S2	C6	11 years	Joystick
S3	C6	8 years	Joystick

E. Experimental Protocol

Control participants trained to use the BoMI in five 1 hour training sessions over three weeks prior to driving the wheelchair. Participants with spinal cord injuries trained over 24 1 hour long training sessions that spanned roughly four months. All subjects practiced bi-weekly. Training involved using the BoMI to control a computer cursor in the same manner as they would control the wheelchair. Instead of controlling the translational and rotational commands, the output vector controlled the coordinates of the cursor on the screen. Subjects completed tasks that relied on a combination of time and accuracy. Specifically, participants used the BoMI to perform virtual center out reaching tasks, type a sentence on a virtual keyboard, play games such as pong or solitaire, and perform virtual driving in the environments described above.

To assess driving ability, subjects performed a modified version of the Wheelchair Skills Test. Specifically, they performed tasks that relied on maneuvering the wheelchair and not tasks that involved maintenance of the wheelchair or accessing objects outside of the wheelchair. Subjects completed seven tasks from the Wheelchair Skills Test: 1) drive forward 10 meters, 2) drive backward 5 meters, 3) drive forward and turn right, 4) drive forward and turn left, 5) drive backward and turn right, 6) drive backward and turn left, and 7) drive through a doorway. Subjects also drove twice through a slalom obstacle course that involved driving around 4 cones. The final task was to assess the ability to make fluent maneuvers in which the control commands had to be constantly changed.

Control participants completed the above driving tests using a Quantum Q6 Edge (Pride Mobility, Exeter PA). Participants with SCIs completed the maneuvers using their personal power wheelchair. All subjects performed the driving test using both the BoMI and a conventional joystick. The maximum speed of the wheelchair differed slightly depending on the specific power wheelchair, however for all tests the maximum speed was kept as close to 1 mph as possible. For data analysis, results were normalized by the maximum speed to account for this difference.

F. Data Analysis

The performance metrics used to quantify driving ability were the path length for each maneuver, the time to complete a specific maneuver, and the smoothness of each maneuver. The smoothness was calculated as the inverse of the number of peaks in the velocity profile. For a given maneuver, peaks were taken only when the velocity was greater than 25% of the maximum velocity and at least 500 ms apart. Prior to peak detection, the velocity profile was smoothed using a 10 sample (0.2 seconds) moving average filter to eliminate high frequency noise. The ability to

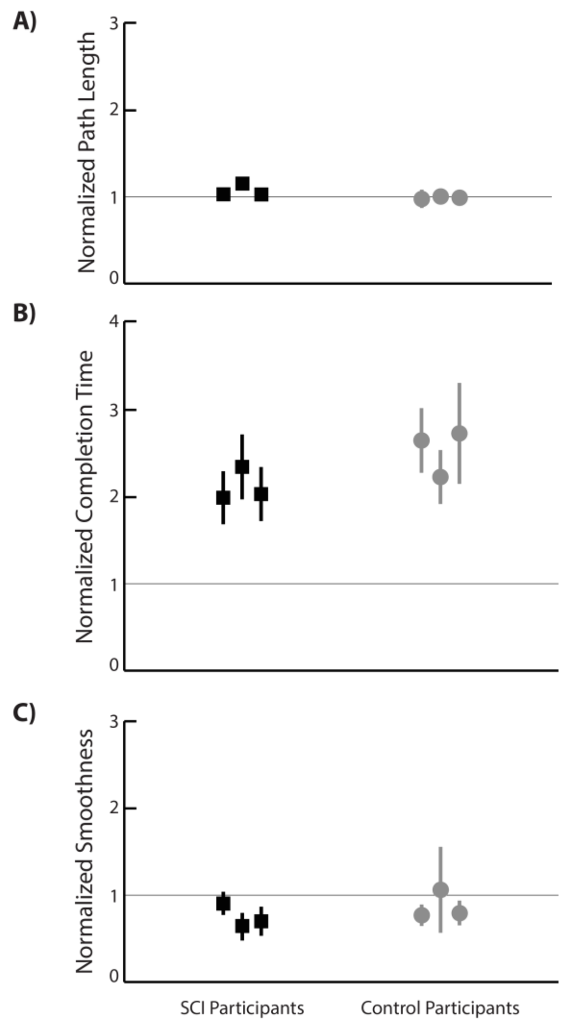

Figure 3. Mean performance metric ratios for each subject. Different bars represent different subjects. The red dashed line is at 1 and indicates no difference between the BoMI and joystick. Error bars represent 95% confidence intervals

maximize smoothness and minimize the number of independent submovements to accomplish a maneuver is a good indicator of the ability to safely and effectively drive a power wheelchair and has been shown to be significantly different between novice and expert power wheelchair users [14].

To normalize across maneuvers, for each maneuver, the ratio of a performance achieved when using the BoMI to the performance achieved when using the joystick was calculated. From this, a single value could be obtained for each maneuver that was independent of the length and difficulty of the maneuver and represented the performance level using the BoMI with respect to maximal performance (using the joystick). A ratio equal or close to 1 indicates that there was no difference in the performance value between BoMI trials and joystick trials, while a ratio greater than 1 indicates the performance using the BoMI was worse than performance using the joystick.

III. Results

All participants were able to accurately control the power wheelchair using only shoulder motions. Figure 3 shows the three performance metrics for each subject averaged across the nine different tasks. The average path length ratio can be seen in Figure 4A. There is no significant difference between the path length ratio and the value 1, indicating that subjects realized paths using the BoMI that were roughly equal to the paths achieved when using the joystick. Across all subjects and maneuvers, there was only a 1.85% increase in the path length (STD = 0.1446). Additionally, all subjects were able to successfully complete all of the maneuvers on the first attempt. Figure 5 shows an example path generated by one of the spinal cord injured subjects driving the power wheelchair through the slalom using the joystick (red) and the BoMI (blue). There are only minor deviations between the paths taken using the joystick and BoMI, despite the subject being an expert joystick user. The paths for the remaining subjects showed similar trends, consistent with the results in Figure 3A.

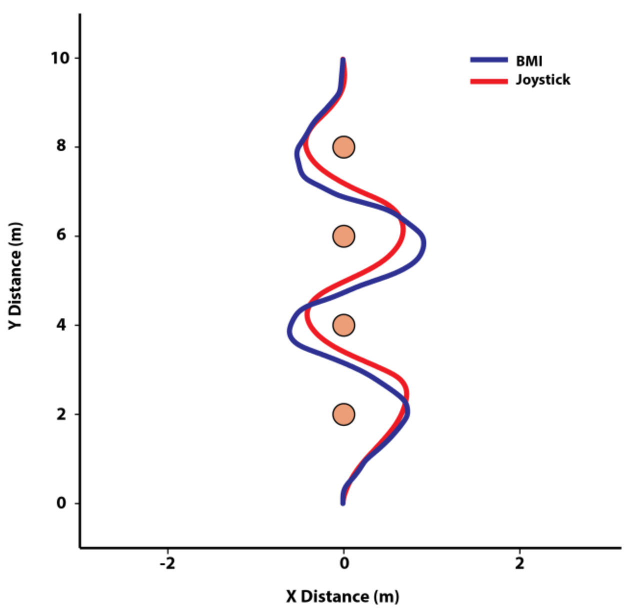

Figure 4. Sample wheelchair trajectory for one of the spinal cord injured participants using the body machine interface (blue) and the joystick (red). The task was to weave in and out of 4 traffic cones. Cones were places 2 m apart.

Despite showing no difference in path length, Figure 3B indicates that time to completion was significantly higher when using the BoMI than when using the joystick for all subjects. On average, subjects took 2.58 times (STD = 1.00) longer to complete each maneuver using the BoMI compared to when they completed the same maneuver using the joystick.

Figure 3C shows the smoothness for each maneuver. For 4 out of 6 participants, the maneuvers using the BoMI were less smooth than maneuvers using the joystick. A closer inspection of Figure 4 reveals that the path taken by one of the participants using the BoMI (blue) is less smooth than the joystick driven trajectory (red). When controlling the wheelchair with the BoMI, this subject frequently stopped traversing to adjust the heading of the wheelchair and then continue moving forward. This behavior is consistent with the findings in Figure 3C and is representative of the other participants.

IV. Discussion

In this paper we describe the development of a BoMI, which uses shoulder kinematics to control the speed and direction of a power wheelchair. The results indicate that users are able to learn to accurately control a power wheelchair to perform basic maneuvers using only small shoulder movements. In fact, all subjects were able to successfully complete all nine maneuvers on their first attempt. Because the path lengths achieved using joystick control and BoMI control were not significantly different, this suggests that users were able to make the maneuvers they intended to without any intervention. This is especially promising for the spinal cord injured participants, who are all experienced joystick users. While continually making shoulder movements may be more fatiguing than making small hand movements, the forward map (A) was designed so control movements would be small and comfortable. More work is needed to quantify how performance may degrade over time, however no participants expressed any discomfort or fatigue after the 1.5 hour training sessions.

We did, however, see a stark contrast in the time to completion and the number of peaks in the acceleration profile for all subjects between joystick control and BoMI control. This is likely attributable to two main factors. The first is the level of confidence and previous experience with the two input devices. Joysticks are ubiquitous in video and computer games and are one of the most popular input devices for manual 2D control. It is therefore not surprising that subjects were more hesitant when using the BoMI, a completely novel control interface with which subjects had little experience, which resulted in not only slower speed but also frequent stops to adjust the heading of the wheelchair. We expect that the gap between BoMI and Joystick performance will shrink over practice using the BoMI, as users become more comfortable controlling the power wheelchair with upper-body motions. Specifically, we expect that users will begin to learn the inverse mapping from wheelchair maneuvers to cursor movement, which users cannot learn from simply using the computer interface.

The other factor that may explain the increase in the number of peaks in the acceleration profile for BoMI control compared to joystick control is the difference in the passive dynamics of the two systems. Joysticks are built with a series of springs that provide resistance against deflection away from the center position. This resistance both makes it easier to maintain a constant position, and acts to smooth any movements, resulting in a naturally smoother velocity profile. The movements of the body in the BoMI system are unimpeded. Accordingly, the control cursor is subject to more rapid position changes. In line with this, users initially found it somewhat difficult to maintain a fixed body posture while the wheelchair was moving. In addition to improvement from training, we expect that implementing a low pass filter or adding some dynamics to the cursor movement will be reduce this effect.

In this study, we have considered the joystick controller as the "gold standard" against which the performance of alternative devices ought to be compared. However, the concept of the body-machine interface has some important collateral features, beside the ability to drive the device. The BoMI is also a means to keep the residual body mobility engaged in performing coordinated motor control tasks. Unlike the brain-machine interface and joystick controllers, the body machine interface can also be programmed to promote physical exercise and to challenge the users to engage parts of the body that lie on the boundary of the paralysis or that tend to be underused. This benefit is critically important to promote recovery and prevent comorbidities in severe paralyses and can be obtained by programming the body-machine map and/or by placing the IMU sensors so as to target specific degrees of freedom of the user's body [15].

Overall, the results presented above provide a proof of concept to the idea that high dimensional shoulder movements can be used to effectively control the speed and direction of a power wheelchair. The results suggest that the wheelchair control scheme described in this paper can be an effective alternative wheelchair controller for individuals with injuries to the cervical spinal cord.

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