APPLICATIONS OF ADVANCED CONTROL INTERFACE TECHNOLOGY FOR INDIVIDUALS WITH UPPER LIMB IMPAIRMENTS

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There are likely a quarter of a million individuals who cannot use power wheelchairs because of an inability to use control interfaces. There are likely even more who desire computer access and whose impairments preclude them from being effective users. Historically, isometric controls were thought to have limited application for individuals with movement disorders due to their sensitivity to unintentional movements. The work in this thesis is a series of studies that demonstrate the potential of an alternative method of control—isometric technology. Our work shows that individuals with upper limb impairments can perform just as well with isometric controls as with conventional proportional control, and in some cases individuals with tremor actually perform better with isometric controls. We also introduce work on adaptive control algorithms that can correct errors in movement made with control interfaces and improve performance.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

PREFACE

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1.0 INTRODUCTION

This thesis is comprised of several fundamental studies that have been the foundation of my work on applications of control interfaces. The first section reviews the previous literature on control interface technology. The second section outlines a study that compares the performance of three different control interfaces for individuals with upper limb impairments during power wheelchair driving, investigating the utility of isometric (force-sensing control) as an alternative control interface. This section also demonstrates the utility of applying advanced customization algorithms to joysticks. The third section describes the control strategies that individuals with disabilities employ while using various types of control interfaces. Sections four and five describe the application of custom filters and other advanced signal processing algorithms for individuals with tremor and other movement disorders. Finally, I close with a section discussing the implications of our work and future aims.

1.1 A REVIEW OF ADVANCEMENTS IN POWER WHEELCHAIR JOYSTICK TECHNOLOGY

In community settings there are over 2 million users of wheeled mobility devices[1]. About ten-percent of these users operate electric power wheelchairs (EPWs) or scooters. Fehr, et al[2], in a survey study, demonstrated that the number of devices users is growing[3] and that there are likely a quarter of a million individuals who cannot use EPWs because of a variety of impairments in motor function, sensation, or cognition. The authors concluded that about half of the individuals who cannot currently operate an EPW by conventional methods could benefit if new technology were developed that could accommodate their needs.

Unimpaired individuals such as surgeons, pilots, and computer operators have been the subject of most control interface literature to date[4-6]. These studies demonstrated the ability of control interfaces to distinguish between intentional or unintentional movements. While there are some conventional control interfaces on the market today that can compensate for some unintentional movements like small amplitude tremor during tasks like EPW driving, we do not yet have the technology that can accommodate many severe movement disorders like larger amplitude tremor or errors that occur due to abnormalities in muscle tone. Combinations of hardware and specialized software applications could theoretically be used to create a customized device for each individual user. Indeed, Riley and Rosen[7], showed that customization of a joystick to an individual user can significantly improve an individual's performance with that control interface.

Advanced customization of joysticks for EPWs is not yet a reality in clinical practice. The standard EPW control most commonly prescribed in the clinic is the proportional movement sensing joystick (MSJ), so named because the device's output (here, the wheelchair's velocity) increases as the joystick is progressively moved away from center. Similar controls used by unimpaired individuals are automobile accelerator pedals and video game joysticks. Efficient use of proportional controls requires adequate proprioception, joint mobility, and dexterity.

Isometric controls, on the other hand, are non-compliant devices that sense force exerted on them; they do not change position perceptively when a subject applies force. An automobile brake pedal is an example of an isometric control. Once the brake engages, the pedal barely moves, but pressing harder with the foot proportionally increases the braking action. Another example is an "eraser head" mouse in the center of some laptop computer keyboards. Nudging the eraser head controls the velocity and direction of the mouse cursor. Isometric joysticks (IJs) can detect intent of motion by sensing force without the need for large displacements of the stick. That is, IJs require only the production of a simple, graded muscle force rather than the movement of multiple joints in the forearm and hand. Zhai, et al[8] have shown that although isometric controls are initially less intuitive to use, once mastered, they may be less fatiguing and produce smoother movement trajectories.

A few previous studies[9-12] have compared IJs to MSJs and have shown that IJs have better accuracy in tasks such as computer target acquisition. However, these studies also reported that because IJs lack damping features, they were very poor at attenuating unintentional movements. Thus, IJs were initially thought to be too sensitive to minor movement disorders and even to normal physiologic tremor. For those individuals already familiar with MSJs, it was thought that proportional control may still be the best option. Yet, because of the desirable features of IJs that allow them to be operable by those with limitations in range of motion and motor control, our work has focused on improving these types of controls.

We have developed The Human Engineering Research Laboratories (HERL) IJ[13-18] (see Figure 1) which uses a programmable embedded microcontroller that provides some flexibility in how the user's input is interpreted. The HERL IJ has been tested in both computer access tracking tasks and in real world driving by subjects with and without upper limb impairments. In prior work, subjects who used the IJ had quicker trial times and movement errors when they used the MSJ for forward and circular driving[14, 15].

Figure 1: Engineering Research Laboratories Isometric Joystick

The next section outlines our first study, which compared our customizable IJ with both a standard IJ and an MSJ during EPW driving for subjects with a variety of movement disorders. Spaeth, et al. developed a Force Sensing Algorithm (FSA)[19] that allows the IJ to act as a simple isometric device. The Variable Gain Algorithm (VGA)[19] allows the IJ to simulate many of the features of an MSJ.

It is important to note here that it is also possible to apply advanced algorithms to standard MSJs. However, for the purposes of our initial work, we wished to compare customized IJs to standard of care, that is, to conventional joysticks that consumers are prescribed in the clinic. Later in this thesis, we will review a study comparing an IJ and a MSJ that have both been customized.

[This review has been published in more detail in the *American Journal of Physical Medicine and Rehabilitation*[20].]

1.2 EFFECTS OF ISOMETRIC JOYSTICKS AND SIGNAL CONDITIONING ON DRIVING PERFORMANCE

1.2.1 Overview

The first study analyzed the performance outcome measures collected in the work of Spaeth, et al[19]. Using a Fitts' law paradigm[21, 22], we evaluated for differences among the MSJ, IJ with FSA, and IJ with VGA with respect to Reaction Time (RT), total trial time or "Movement Time (MT)", and Driving Accuracy (DA). We hypothesized that the IJ with VGA would have a similar RT to the MSJ, while the IJ with FSA would have a shorter RT than the MSJ because of its minimal damping effect on movements. We also hypothesized that the IJ with either the FSA or VGA algorithms would produce longer MTs to both near and far targets than the MSJ likely due to them being novel devices for users. We further hypothesized that the VGA software would provide better DA than the FSA software because of its emulation of the familiar MSJ.

1.2.2 Joystick Technology Description

The Flightlink MSJ used in this study has a post that can be displaced to a maximum angle of about 18 degrees and is prevented from moving further by a circular "stop ring," a mechanical barrier or template. The user obtains distinct, positional feedback when the stick contacts the template, indicating maximal excursion has been reached. For those with intact proprioception, this barrier is very useful to guide forward cruising or tight turning, allowing the user to sense the position of the stick in space. The IJ, being rigid, provides no mechanical feedback when the user reaches the maximum signals. The microcontroller monitoring the force sensors creates a “virtual template” by locking the signal on each axis when the maximum allowable force is reached. This lack of mechanical feedback may cause inexperienced operators to apply excessive force if they are unaware that it is no longer effective or necessary. Experienced IJ users judge the maximum force required by the behavior of the output device, e.g. wheelchair, since the device’s ballistics provide the feedback.

User input is a combination of two channel variables, Speed (Forward-Backwards) and Direction (Left-Right). Power output is proportional to the hypotenuse of these two orthogonal input signals. The mechanical template of the MSJ used in this study is pentagonal. The IJ with FSA software combines the speed and direction axes. The FSA software has a square virtual template because the signal locking of one axis is carried out independent of the other. At 45 degree angles, both axes produce their maximum signals (at corners of the square template) whereas the pentagonal template of the MSJ limits the maximum output during diagonal movements. The resulting power vector during diagonal inputs on the IJ with FSA is much larger than power in the vertical or horizontal directions alone, resulting in over response to diagonal inputs. This can cause unexpected results for a driver who does not expect a boost in speed when pushing the IJ in diagonal directions. This diagonal bias was corrected in the VGA software. The microcontroller reads the user’s Speed and Direction signals and passes them through an Arc Tangent function to obtain the proper resultant vector. This eliminates most of the diagonal bias and creates an approximately circular template.

The MSJ used does not have a programmable filter feature, but it does have some inherent damping due to a rubber boot. The IJ, regardless of algorithm applied, uses a second order Butterworth 100 Hz low pass filter for damping of higher frequencies such as those from road vibrations.

Another contrasting feature between the MSJ and the IJ is the configuration of the **Dead Zone**. The Dead Zone is the region near the joystick center that permits small inputs from force exertion on the stick but generates no output. The purpose of the Dead Zone is to provide the user with the ability to maintain hand contact with the joystick handle without producing output from movements that are unintentional, such as those from physiologic tremor. The MSJ has a distinct Dead Zone due to the centering spring having some preload. This is in contrast to a Dead Zone that would be set by software or electronics. Until the operator applies enough force to exceed this preload, the post does not move, and thus no output is generated. There is also some Dead Zone built into the electronics in the event that the centering spring does not return the post exactly to center. The IJ with FSA has no equivalent threshold, so any force applied to the post activates the sensors and generates an output signal, making it sensitive to tremor or spastic movements. This was corrected in the IJ with VGA by using software to block the low level signals, thus creating a "virtual" Dead Zone.

Another important feature of a control interface is the **Gain**. Gain is the force to power ratio, or how much effort the subject must apply to the device, in this case the stick, to produce a given amount of output, i.e. wheelchair speed or acceleration. When the IJ with VGA was created, we measured the Gain of the MSJ and duplicated it on the IJ through software amplification. Interestingly, when we changed the Gain of the IJ with VGA to match that of the MSJ, clinicians told us it was too sensitive in our pilot testing. Lower gain may have been needed in order to overcome the lack of proprioceptive feedback that occurs with a rigid device. Obtaining full power from an MSJ also requires stick displacement and overcoming both the inertia and the damping effect of the rubber bellows. We experienced the same problem with

FSA software and set the FSA Gain to one fourth that of the MSJ. Perhaps due to better template management, a 50% reduction in Gain proved adequate for the IJ with VGA. The overall objective was to create VGA software that would emulate the familiar and convenient features of the MSJ by creating virtual equivalents to the MSJ's template, Dead Zone, and Gain.

1.2.3 Subject Testing

The subjects in this study were recruited through therapist referrals and with advertisements in group homes and local disability organizations. The procedures followed were in accordance with the ethical standards of the Institutional Review Board at the University of Pittsburgh and The Department of Veterans Affairs.

Inclusion Criteria for subjects were that they used an EPW as their primary means of mobility with hand operated joystick control at least 20 hours per week, they were able to tolerate testing for two and a half hours, they were between the ages of 18 and 80 years, they were able to transfer in and out of a test wheelchair or willing to be assisted by clinician, and they were able to use the test EPW with a 45.7 cm x 45.7 cm (18 in by 18 in) seat size and a sling style wheelchair back. Subjects were excluded if they had an open pressure sore because of concerns that prolonged testing could worsen wounds.

After providing informed consent, subjects transferred into a Quickie P300 EPW (Sunrise Medical, 1994) with a 45.7 cm x 45.7 cm (18 in x 18 in) seat and unoccupied weight of 100 kg. The investigator configured the wheelchair for each subject by adjusting the armrests, legrests, footrests, and joystick position, and added chest straps or bolsters for trunk support as needed. Subjects used their own cushions. One of two test joysticks was mounted to the EPW in random order: a commercial Flightlink MSJ or HERL IJ. The IJ was equipped with a switch that allowed either the FSA or the VGA algorithm to be selected, for a total of three joystick interventions. A custom joystick mount[15] allowed quick interchange of joysticks. The joysticks were mounted on the same side of the test chair as on the subjects' personal chair. The MSJ and IJs were connected to the wheelchair's controller with a standard cable.

Nine circular, black vinyl targets 155 cm in diameter were taped to the floor in the pattern indicated in a double semicircular array. Near targets were 305 cm, and far targets were 538.5 cm from the starting position. The calculations that yielded appropriate target width and distance from starting position were developed previously according to Fitts Law[15].

A Dell Inspiron 7000 laptop computer was mounted on the back of the test chair. We wrote our own data collection program for the laptop using the C++ language and Borland Builder Version 5 integrated development environment [23]. The program created a folder for each new subject on the hard drive and generated a list of the randomized trials. The laptop saved data from recorded trials as text files.

At the start of each trial, the target to be acquired was displayed on the laptop screen, and the investigator announced this to the subject. The software generated an audible "beep," prompting the subject to drive and concurrently launched event timers. Subjects drove the test EPW from the mid point of the array to the various target announced. Subjects were asked to drive as quickly and accurately as possible. Target acquisition was detected by four optical sensors on the undercarriage of the EPW in a rectangular format (122 cm by 43.2 cm), positioned under the footrests and behind the rear tires. A small logic board monitored the optical sensors and signaled the computer whenever all four of the sensors were over a target. When the subject reached the target, the optical sensors stopped the clock and launched a second clock to verify the sensors remained over the target for two seconds. The laptop sampled joystick inputs and outputs every 12 milliseconds through a serial connection. Subjects then returned to starting position and repeated the trial, moving to a different target.

Subjects were allowed nine practice trials using the IJ. Then, the first joystick used for recorded trials was randomly selected. The joystick being used was randomly changed after every nine trials. Subjects drove from the starting position to each of nine targets three times with each of three joysticks for a total of 81 recorded trials. To minimize learning effect and fatigue, the investigators set the randomization software to allow no more than two consecutive presentations (18 trials) with any particular joystick.

The IJ and MSJ have identical speed controls. All but one subject elected to use the joysticks with the speed control set to its maximum (turned fully clockwise). For the subject requesting a lower setting, we were able to match the settings of the IJ and MSJ.

1.2.4 Outcome Measures

Dependent variables included Reaction Time, Movement Time and Driving Accuracy. **Reaction time** (RT) was defined as the interval between the computer's "start" beep and the first input signal detected at the joystick and was recorded in milliseconds. **Movement Time** (MT), recorded in milliseconds, was defined the interval from the end of RT until target acquisition was detected by the sensors. If the subject successfully remained on target for 2000 milliseconds, proof that the chair stopped and did not roll off the target, the **Driving Accuracy** (DA) was recorded as a "Hit." Otherwise the DA was recorded as a "Miss." If the subject did not acquire the target after 40 seconds, the computer halted the driving trial and scored it as a "time out," which we then converted into a "Miss."

1.2.5 Data Analysis

To evaluate for the presence of learning effect, data files were split so that data was organized by joystick and target so that any learning effect found could be controlled for in subsequent analysis. A repeated measures ANOVA was used to analyze the data, using RT and MT as within-subjects factors. We also organized data by RT and MT in the order that the trials were performed.

Average DA for each of three joysticks was calculated for each subject using all trials. Then, MT and RT data for trials resulting in missed targets was discarded so that average values for RT, and MTs for near and far targets for successfully acquired targets could be calculated for each of the three joysticks for each subject. Then, overall average RT, and MTs for near and far targets were calculated for each joystick. Four separate mixed models were used to address the hypotheses and so that fixed and random effects could be incorporated into the model. Analyses were completed using SAS[24]. Significance level was set at 0.05. There were repeated trials for each subject, so subjects were entered as the random factor. Models were established to evaluate differences between joystick type with regard to RT, MT for near targets, MT for far targets, and DA.

1.2.6 Results

Thirteen subjects gave informed consent to participate, and eleven subjects completed all trials. One subject with Cerebral Palsy (CP) became fatigued during the study, likely because his own chair had highly customized seating installed, and thus our test chair did not have optional positioning for his scoliosis. Another subject with CP had difficulty using the controls because she typically used a front wheel drive chair, while our test chair was rear wheel drive. The study was terminated for these subjects because their driving was considered unsafe and their data was not included. Subject number in each diagnostic category for the rest of the fully tested subjects was as follows: 4 (36.4%) with CP, 2 (18.2%) with Traumatic Brain Injury (TBI), 2 with Spinal Cord Injury (SCI), 1 (9.1%) with Muscular Dystrophy, 1 with Spina Bifida (SB), and 1 with Polio. Average age was 37.8 $\pm$ 10.9 years. Six subjects were male (54.6%), and 5 were female (45.5%). Six subjects were Caucasian (54.6%), 4 were African-American (36.4%), and 1 was Asian-American (9.1%). Subjects with TBI, SB, and Polio drove with their dominant hand, which was unaffected by their diagnosis but required special positioning in the EPW due to spasticity and contractures. All other subjects had impairments involving both upper limbs and also required special positioning.

Average results for RT, MT for near and far targets, and DA based on joystick type are listed in Table 1.

Table 1: Performance outcome variables based on joystick type

	RT (ms)	MT for near targets (ms)	MT for far targets (ms)	DA (% hits)
MSJ	1174.6 $\pm$ 884.0	6049.0 $\pm$ 2239.0	7728.8 $\pm$ 3241.4	90.6 $\pm$ 10.4
IJ with FSA	847.0 $\pm$ 610.7	6742.3 $\pm$ 2626.7	8615.0 $\pm$ 3837.9	93.3 $\pm$ 7.0
IJ with VGA	969.9 $\pm$ 626.7	6556.3 $\pm$ 2806.9	8360.1 $\pm$ 3463.1	90.6 $\pm$ 15.2

Learning Effect: Based upon Mauchly's Test of Sphericity, subjects' RT and MT scores did not significantly change after consecutive trials.

Reaction Time: The IJ with FSA had a significantly shorter RT than the MSJ ($p=0.0020$). The IJ with VGA was not significantly different than that of either the IJ with FSA or MSJ with respect to RT.

Movement Time: No significant differences in MTs were detected when analyzing the closer targets (1 through 5). With the four more distant targets (6 through 10), the IJ with FSA had significantly longer MT ($p=0.0159$) than the MSJ. However, the MT of the IJ with VGA was not significantly different than that of either the IJ with FSA or MSJ.

Driving Accuracy: No significant differences in DA were detected between any of the joystick pairs.

A power analysis was performed for this study based on previous work[14, 15]. Ten subjects yielded 80% power to detect differences at the 0.05 alpha level. Anecdotally, most subjects claimed that they preferred using the MSJ because it was a familiar device.

1.2.7 Discussion

Research done prior to this study demonstrated prolonged RTs and MTs when individuals with movement disorders used isometric devices. However, this study demonstrates that with appropriate customization, hypersensitivity to unintentional inputs can be attenuated. Our IJ with VGA had reduced sensitivity to small amplitude unintentional movements and high frequency environmental vibrations, similar to the Dead Zone and Filter features of an MSJ, while retaining the unique force reflecting qualities of an IJ.

The RT while using VGA software was similar to that of an MSJ, likely because of the built in Dead Zone. As expected, the IJ with FSA had a shorter RT than the MSJ, likely due to its minimal Dead Zone. Further work is needed to identify the Dead Zone parameters for an IJ that will optimize RT, especially in the presence of unintentional subject movements.

The results also support our hypothesis that the IJ with FSA software would produce a longer MT than the MSJ. This was true, however, only when far targets were tested. It is possible that a longer distance was needed in order to detect differences. Many of the subjects in the study were observed to stop prematurely when using the IJ. This may be because subjects are accustomed to coasting to a stop. Even so, the IJ with VGA exceeded our expectations and produced a MT statically indistinguishable from that of the MSJ. Further work could be done to develop an algorithm for an IJ that will allow more gradual deceleration. However, additional training may prove to be all that is needed for a new IJ user.

Indeed, this study shows that despite being familiar with MSJs, users can use an IJ accurately with very little practice time. Accuracy was greater than 90% for all three joysticks. It was impressive that the performance of the IJ with VGA and even FSA software was at least as accurate as the MSJ. Even though subjects did not practice with the MSJ in this study, and some subjects did not drive their own EPW, all subjects were experienced MSJ users and performed maneuvers typical of everyday driving.

A limitation of this study was that users were not tested in their own wheelchairs because the testing equipment was not adaptable to personal EPWs. Two subjects could not complete the study. Our future work will allow the equipment to be integrated and tested on personal equipment.

An additional consideration is that the ability to perform a Fitts' law task does not necessarily predict the ability to maneuver an EPW in a real environment. We also did not address reverse driving which requires different steering strategies. Future work in assessing these devices in real world environments is needed.

[This study has been published in more detail in the *American Journal of Physical Medicine and Rehabilitation*[20].]

1.3 FORCE CONTROL STRATEGIES WHILE USING MOVEMENT AND ISOMETRIC JOYSTICKS

1.3.1 Overview

In this study we examined the time-series data recorded in the previous study to characterize subjects' force control strategies. In the prior study, we noted that when using the IJ with sophisticated signal processing [13, 25], subjects exhibited RT, MT, and DA statistically indistinguishable from the MSJ. When using the IJ with a basic control function, subjects had lower performance than when using the MSJ. However, we observed that, regardless of the control software installed, users tended to exert more force than necessary on the IJ. The propensity to hyper-activate isometric controls clearly has implications for users with profound weakness or fatigue and may bias outcome studies comparing IJs and MSJ, if not accounted for.

1.3.2 Background

The literature on force control strategies is limited, but some research has been done on unimpaired individuals exposed to high acceleration environments, similar to those experienced by Air Force pilots. This type of environment causes subjects to misjudge the amount of force needed for isometric control and to apply excess force to the stick and must rely on proprioception to fine tune their force exertion [26-28]. When vibration is added to reduce proprioception, subjects apply an insufficient force magnitude to the IJ but often can continue to apply appropriate directional force. Although these concepts may not apply to subjects who are not in high acceleration environments, they illustrate the fundamental role that proprioceptive and other afferent sensory pathways play in fine-tuning force production.

Deflecting the stick of an MSJ to the extent that it rests against the mechanical template provides unequivocal feedback to the driver that peak power has been selected. Some templates also include "notches" that guide the stick toward locations along the template that are theoretically useful for tight turns and driving long distance at high speeds. However, the extent to which users rely on the MSJ template and the notches within it to guide driving, especially short distance driving, is not well understood. If drivers use feedback from the position of the MSJ stick for closer quarter maneuvers, rather than using the template, they may have difficulty applying appropriate force when using an IJ in similar maneuvers, since the IJ does not offer positional feedback. When the user reaches maximum signal boundary, the lack of positional feedback may be one reason users exert too much force on the stick and may explain why isometric devices are often less intuitive to use [29].

The purpose of this study is to better characterize the force exerted on the three different joystick presentations in the prior study to determine how much subjects rely on the MSJ's template. We also wished to determine how much excess force was used on the IJs, and whether that excess force was associated with improved time, improved accuracy, or simply was wasted. We hypothesized that subjects would exert higher average and excess forces on the IJs than the MSJ since they are novel devices with lower gain and provide little positional feedback. We also hypothesized that subjects would exert less force on the IJs over subsequent trials. We finally hypothesized that using higher average force is related to quicker trial times and poorer accuracy, regardless of joystick used.

1.3.3 Methods

This study analyzed the results from the protocol in the previous study. The actual mechanical and virtual templates (VTs) are discussed in the prior section. For the IJ with FSA, gain was set to 100 mV/N up to the VT; forces that subject exerted that exceeded 100 mV/N were recorded but truncated to VT values. For the IJ with VGA, gain was set to 200 mV/N up to the VT; forces exceeding 200 mV/N were recorded but truncated to VT values. The VGA dead zone was set such that low amplitude inputs (< 0.9 N) occurring from unintentional/resting movements produces no output. Actual gain on the MSJ was measured at 400 mV/N.

Joystick gain and dead zones are depicted in Figure 2. Operational range, or the range of force for which the control functions are defined, are listed in Table 2 in the results section. We plotted instantaneous force versus time to create force-time curves for each trial. We calculated the average applied force (F_a) used during each trial by dividing the area under the force time curve by the total trial time (t_{total}). We defined excess force (F_e) for each trial as the average of the differences between the actual applied force and the IJ's VT when the VT was exceeded. This was not calculable for the MSJ since there is no stick excursion beyond the mechanical template. We defined Control Efficiency (CE) as the Newton·seconds (N·s) expended within the operational range divided by the total N·s of each trial; thus CE was represented as a percentage. We calculated an overall average percent (P) of trial time in which subjects exerted force equal to or greater than the VT for the IJs. We quantified hyper-activation (A) in N·s by integrating the area between the user's applied force curve and the curve representing the VT and summed all hyper-activation episodes that occurred during each trial.

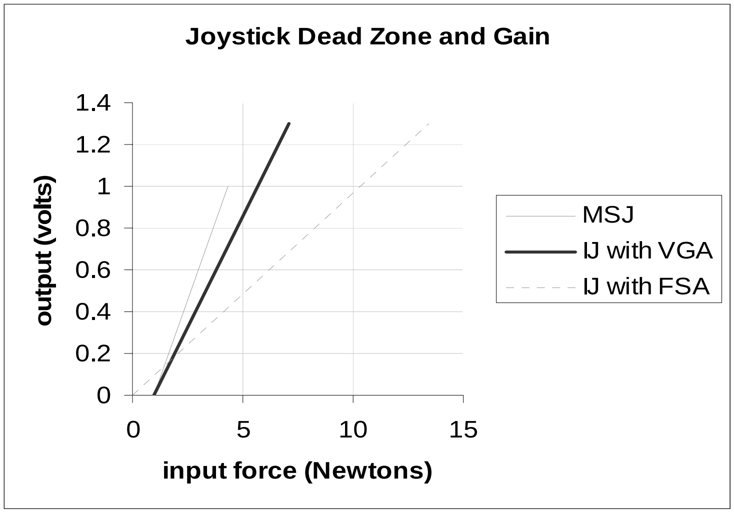

Figure 2: Dead Zone and Gain of Joysticks

1.3.4 Statistical Analysis

We set all alpha at values equal to 0.05 *a priori*. We used SPSS [30] for univariate analyses and the following non-parametric analyses. We used non-parametric statistics since F_a and trial time were not normally distributed. To evaluate for a learning effect for F_a used in subsequent trials, we performed a Friedman Test. Given no learning effect was observed, we calculated an overall F_a for each joystick. We used Mann-Whitney Tests to determine if F_a for each trial was related to driving accuracy. We then performed Spearman Rho correlations to evaluate the relationship between F_a and trial time for successfully acquired targets. We used SAS [31] to complete a Mixed Model Analysis on trials with successful target acquisition to evaluate differences among joysticks with respect to F_a, F_e, CE, P and A.

1.3.5 Results

Table 2 shows differences among joysticks with respect to outcome measures. The IJ with FSA and the IJ with VGA had a significantly higher F_a than the MSJ ($p<0.0001$, $p<0.0001$). The IJ with VGA had a significantly higher F_e than the IJ with FSA ($p=0.0058$). The IJ with VGA had a significantly lower CE than the IJ with FSA ($p<0.0001$). The IJ with VGA had a significantly higher P than both the IJ with FSA ($p=0.0042$) and MSJ ($p=0.0006$). The IJ with VGA had significantly higher A_n than the IJ with FSA ($p=0.0264$). The IJ with VGA did not have significantly higher A_f than the IJ with FSA ($p=0.0554$).

Table 2: Comparison of Joysticks with respect to Force Outcome Measures

		IJ with FSA	IJ with VGA	MSJ
OR	Operational Range (N)	0 to 13.4	0.9 to 7.1	0.9 to 4.3
F_a	Average Force (N)	5.8 ± 1.9	5.5 ± 2.4	2.0 ± 0.3
F_e	Excess Force (N)	0.7 ± 1.1	3.0 ± 3.0	N/A
CE	Control Efficiency (%)	98.0 ± 4.4	88.5 ± 16.0	N/A
P	Average Percent of trial driven in hyperactivation state	3.9 ± 6.1	20.6 ± 20.0	N/A
A_{near}	Area under the Excess Force Curve for Near Targets (N·s)	0.1 ± 0.1	6.5 ± 9.0	N/A
A_{far}	Area under the Excess Force Curve for Far Targets (N·s)	0.4 ± 0.5	9.1 ± 14.1	N/A
DA	Driving Accuracy (%)	93.3 ± 7.0	90.6 ± 15.2	90.6 ± 10.4

There was no learning effect seen for F_a for any joystick (FSA $p=0.184$, VGA $p=0.117$, MSJ $p=0.804$). F_a was negatively correlated with trial time for all 3 joysticks for successfully acquired near and far targets, as shown in Table 3. F_a was not related to DA for any of the joysticks (FSA $p=0.306$, VGA $p= 0.126$, MSJ $p=0.340$).

Table 3: Correlation Statistics for Average Force and Trial Time for Joysticks.

	Near Targets	Far Targets
IJ with FSA	r = -0.392 p = 0.000	r = -0.337 p = 0.000
IJ with VGA	r = -0.176 p = 0.015	r = -0.177 p = 0.023
MSJ	r = -0.388 p = 0.000	r = -0.572 p = 0.000

1.3.6 Discussion

Interestingly, subjects rarely used the MSJ template to guide driving but rather used continuous re-positioning of the stick inside of the template. It seems plausible that proprioceptive feedback would be important, then, for this type of control. Indeed, Sand, et al. [28] showed that proprioception is important for fine tuning of movement when the force exerted is below 5 N. Since the operational range of the MSJ in this study was 0.9 to 4.3 N, proprioception is likely an important aspect of using an MSJ in environments involving short distance driving with turns, such as some indoor maneuvers.

Although we did not formally test proprioception in our subjects, the work of Sand et al. shows that loss of proprioception results in under-production of force on an IJ [28]. Therefore it is unlikely that proprioceptive loss affected IJ use in this study since subjects exerted over twice the average force on the IJs as on the MSJ, and over five times the force needed to produce any output at all.

On the IJ with FSA, subjects exerted on average more than 5 N. Yet, subjects rarely exceeded or met threshold force when using this joystick, and when they did exceed threshold it was only by about 0.7 N. Furthermore, the control efficiency of 98% was impressive. These findings suggest that the force subjects were using on the IJ with FSA was appropriate, albeit higher than that required to operate an MSJ, and likely due to the lower gain of this device.

On the IJ with VGA, subjects used on average more than 5 N as well. However, they spent over 20% of the average trial time meeting or exceeding the VT, and used about 3 N of excess force. Control efficiency was approximately 89%. Since the IJ is operable with lower force, further work is needed to determine whether subjects used more force simply because positional feedback is not as important for joystick control with inputs above 5 N. An alternative explanation is that, since subjects were blinded to the IJ algorithms used and since the FSA required more force to operate, subjects may have applied the force required to operate the FSA when the VGA was used. Further explanations are that there was inadequate training or suboptimal gain adjustment. As we mentioned previously, we set the current gain of the VGA software based on feedback from clinicians who had determined it to be too sensitive in our initial pilot work. Further customization of gain for individual users may reduce the propensity for fatigue.

Surprisingly, we observed no learning effect for average force. Longer training sessions specifically with the IJ may be needed to allow subjects to learn how much force to apply from the ballistics of the wheelchair. Previous research on unimpaired subjects demonstrates that subjects internalize a model of the ballistic response of the objects they are acting upon [32, 33] and this may be sufficient feedback to influence control strategies.

Exerting higher average force on all joysticks was correlated with decreased trial time, likely due at least in part to increased speed. However, the correlation for the IJ with VGA was low, again suggesting some of the force used on this IJ was excessive. Although this is not a true Fitts' Law task since subject's maximum speed was constrained to the EPW's top speed, subjects spent the majority of the trials below threshold force. Thus, speed was likely freely chosen for a majority of the time. We did not confirm our hypothesis that using higher average force is associated with poorer accuracy. Hence, directional steering may compensate for some errors in judgment of appropriate force exertion. Indeed, Sand's work [28] has shown that directional IJ control is not as sensitive as to positional feedback as force magnitude.

The larger forces used, possibly an adaptation to the lack of positional feedback or a result of lower gain, may be influenced by further training with such a novel device. Yet, even though the IJ required more force to operate than the MSJ overall, subjects' driving performance was not affected, indicating that it may be useful to explore additional customization parameters for the IJ. The next section outlines a study in which we move one step further and add a custom filter.

[This study has been published in more detail in *IEEE Transactions on Neural Systems and Rehabilitation Engineering*[34].]

1.4 APPLICATION OF A TREMOR FILTER

1.4.1 Overview

The Weighted-frequency Fourier Linear Combiner (WFLC) filter has been used to cancel tremor effectively in microsurgery[5, 35]. The purpose of this study was to compare an MSJ, IJ, and IJ with the WFLC filter in individuals with tremor while they performed a virtual driving task.

1.4.2 Background

Tremor is an involuntary, oscillatory motion[36] that can make the use of a control interface difficult. Human tremor and its effect on control interfaces have been studied for over 30 years. Stiles, Randall, and Rietz [37-42] have described the range of frequency and amplitude of normal and pathologic human tremor. Hefter, et al[43] also validated a concept that altering hand mechanics affects tremor properties. Riley and Rosen[7, 44] compared isometric and standard motion sensing controls in a target selection task and found that no single control type worked best for those with tremor, but rather customization of the interface for individuals users produced the best results.

Part of the initial design concept for our IJ was to attempt to the amount of positional changes of the hand and limb needed to operate the device. Limiting changes in hand motion should theoretically make intentional tremor easier to filter, since there should be fewer changes in tremor properties.

The conventional means for filtering is a simple low-pass filter. While a low pass filter may be effective for most users with physiological tremor and for vibrations transmitted to the device from the ground when using a wheelchair, tremor frequencies in individuals with Multiple Sclerosis (MS) can be as low as 3.5 to 5.0 Hz[45]. Setting the cut-off frequency this low will introduce a phase lag and potentially eliminate intentional commands. Notch filters, on the other hand, can suppress only the tremor frequency and reduce distortion of intentional signals[35]. Adaptive filters, those that self-adjust their parameters, have an added benefit because tremor is not always constant[35]. An adaptive notch filter, such as the weighted Fourier linear combiner (WFLC)[5, 35], can filter a small band of frequencies without adding significant delay[5]. In fact, prior studies have shown improved performance with a WFLC compared to a low-pass filter when individuals with CP used an IJ to perform virtual wheelchair driving tasks[46]. In that particular study, a high-pass filter was added as a safe-guard to ensure that the WFLC did not track intentional movements.

In the current study, our primary objective was to evaluate the performance of individuals with tremor while using an IJ with WFLC filter, an IJ without a filter, and our joystick in MSJ mode. We hypothesized that in a virtual driving environment, the subjects' performance statistics would be ranked from best to poorest in this order: IJ with filter > IJ > MSJ.

1.4.3 Methods

This study was approved by the VA Pittsburgh Medical Center Institutional Review Board. We recruited subjects who attended the 2007 National Veterans Wheelchair Games in Milwaukee, WI. Individuals who approached our information booth and indicated an interest in participating were provided an informed consent document and reviewed for eligibility.

Subjects were required to be between the ages of 18 and 80 and possess a pathological tremor. If a person was not able to sit upright for three hours, had an active pelvic or thigh wound, or had a history of seizures in the last 90 days (since seizures could theoretically be induced by video game like tasks), he or she was excluded. Subjects filled out questionnaires discussing daily living activities, as well as previous computer and wheelchair use. A physiatrist performed a brief neurological examination, including a tremor assessment.

The IJ was customized for each subject by using custom tuning software to set a dead zone, bias axes, and establishing optimal gain for each subject. During the virtual driving tasks, subjects sat in their own wheelchairs, and those who were ambulatory sat in a desk chair. The driving tasks consisted of two tracks viewed at Bird's Eye level, one simulating a left hand turn and the other a right hand turn. See Figure 3.

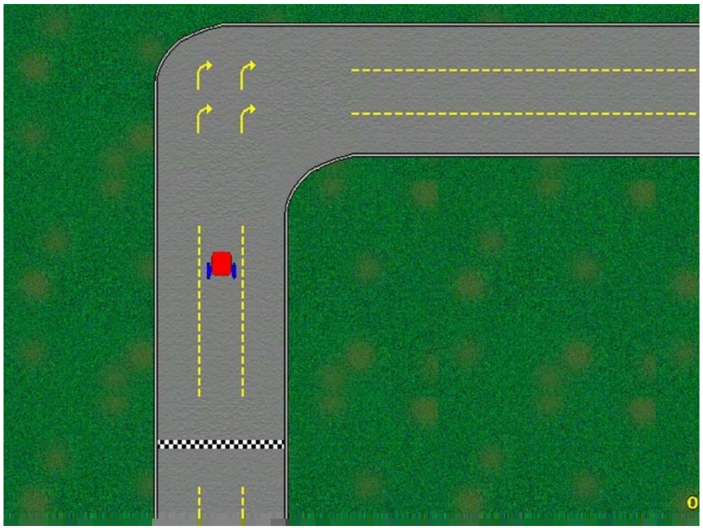

Figure 3: Bird's Eye View of Virtual Driving Task in Tremor Filter Study

Subjects were instructed to complete trials as quickly and accurately as possible. Subjects sat in front of a computer screen to perform the tasks and used the interchangeable joysticks mounted on a table with an adjustable height. Subjects practiced with the IJ and the MSJ for 36 trials, or up to 30 minutes.

The WFLC filter was customized for each subject by using data from the practice trials. We set the initial estimate of maximum frequency based on the expected frequency range of the subject's diagnosis, using tremor associated with MS at 3-5 Hz, tremor associated with Parkinson's Disease (PD) at 3-7 Hz, and other pathological tremor at 2-8 Hz [45, 47].

Next, we selected optimal filter parameters for each subject using a validated protocol[5,46]. This involved visually inspecting frequency and power output curves generated from practice trials and adjusting the filter parameters so that the output to the controller with filter applied most closely matched the user's input on the device. For each subject, these parameters were applied only to the IJ with WFLC. We also added a high pass filter to the IJ with WFLC, which was set at 2 Hz for both the speed and direction axis.

Subjects then performed 20 trials with each of three joysticks in random presentation. We sampled joystick input and output at 59.4 Hz. From sampling data, we calculated total trial time, boundary violations that occurred when subjects drove outside of the path, and root mean square error, defined as the average deviation from the center of the path measured in pixels starting at the center point of the body of the wheelchair icon and following the shortest distance to the path midline. The boundary was four times the width of the virtual wheelchair.

1.4.4 Statistical Analysis

We calculated average outcome variables over trials using MATLAB[48]. All alpha levels were set to 0.05 *a priori*. SPSS and R were used to perform all analyses[30, 49]. Log scale transformations were used to stabilize the variance and effects of outliers. Mixed Model Analyses were run to evaluate for differences among joysticks with respect to RMSE and time, using subject as the random factor effect, and joystick as the fixed effect. Generalized Estimating Equations were used to evaluate for differences among joysticks with respect to boundary violations because this variable was not normally distributed.

1.4.5 Results

Four men and one woman with an average age of 61.2 +/- 12.2 years participated in this study. All were Caucasian. Two subjects had a diagnosis of MS, one had paraplegia from a spinal cord injury, one had PD, and one possessed tremor due to medication. Two subjects were wheelchair users, and only one used a computer regularly. Two subjects had intention tremor, and three had both intention and resting tremor. Average results of outcome measures are listed in Table 4. No subjects were excluded.

Although the average RMSE was lowest for the IJ, there were no significant differences among joysticks with respect to RMSE ($p=0.5316$).

The IJ produced significantly lower trial times than the IJ with filter and the MSJ ($p=0.0425$ and $p<0.001$). The average driving time for the IJ was 32.4 seconds, while the average driving time for the MSJ was 41.7 seconds, illustrating an approximate 10 second difference, or a 22% reduction in trial time. No differences were found between the other joysticks.

The IJ produced significantly fewer violations than the MSJ ($p<0.001$). There were no significant differences between the other joysticks.

Table 4: Performance Outcome Measures by Joystick in Tremor Filter Study

	IJ	IJ with Filter	MSJ	p-value
Average Root Mean Square Error	11.4 ± 4.0	11.6 ± 4.8	12.5 ± 5.5	0.284
Average Trial Time (s)	32.4 ± 11.4	39.1 ± 22.2	41.7 ± 20.9	0.0425* <0.001**
Average Number of Violations	0.71 ± 1.3	0.75 ± 1.3	0.97 ± 2.4	<0.001***

*significance in trial time between IJ and IJ with filter.
**significance in trial time between IJ and MSJ.
***significance in collisions between IJ and MSJ.

1.4.6 Discussion

The results did not support our hypothesis that the filter would improve driving performance. The WFLC may have damped some of the subjects' intentional movements in this study. Since the WFLC has been used effectively in studies on handwriting and microsurgical tools[5, 35], it may actually be most effective on higher frequency, lower amplitude tremor seen in unimpaired individuals.

One limitation of this study was the lack of homogeneity of tremor etiology. We anticipate future studies on larger groups of subjects. Because our technology now allows us to record a wide range of motor control parameters as individuals use joysticks, using them as a classification tool in order to quantify tremor is now feasible and will be the aim of future work. This may be particularly useful in measuring response to pharmacologic treatment in PD.

In our prior work, we have shown how IJs can perform just as well as MSJs in a real driving environment[20]. However, this is the first study in which we have been able to show that the IJ (without filter) performed better than the MSJ. This suggests that either the tuning software, which helps to eliminate some resting tremor and excess force exerted beyond that needed for control, or the rigid handle unique to the IJ design, may have a beneficial effect on performance. It is indeed likely that the rigid handle of the IJ may have some tremor damping effect. The next section describes a study that investigates the utility of adding tuning parameters to a conventional MSJ. In that study we also evaluate the application of other advanced algorithms that may be useful to address performance errors of individuals with movement disorders.

[This study has been published in more detail in the *Journal of Rehabilitation Research and Development*[50].]

1.5 ADDITIONAL ADVANCED ALGORITHMS

1.5.1 Overview

This study's aim is to compare performance of the two algorithms versus no algorithm at all in a computer based tracking task. We hypothesized that application of the algorithms would result in improved performance compared to no algorithm, regardless of joystick used.

1.5.2 Background

We also developed advanced algorithms that can be applied to various interfaces to correct for subject errors in movement. The Proportional Integral Derivative (PID) controller[51] corrects error in a trajectory by measuring differences between ideal path and cursor location, the degree to which the movements overshoot or undershoot the ideal path, and the degree of oscillation about the ideal path. A Least Mean Squares (LMS) algorithm[52-55] uses the speed error, direction error, and frequency variation to correct tracking trajectory.

1.5.3 Methods

We recruited subjects between ages 18 and 80 years with upper limb spasticity or rigidity. Subjects used a conventional Quickie[56] brand MSJ and our IJ[20, 34]. The joysticks were positioned according to subjects' preferences using a custom-made mounting device clamped to the desktop.

We tuned both the IJ and the MSJ for each subject according to our tuning protocol[20], setting a dead zone, biasing the directional axes of the joystick, adjusting the maximum force the controller would recognize, and customizing the gain. The two algorithms used in this study are explained in more detail in Appendix A.

Subjects used control interfaces to trace 3 different shapes on a laptop screen. The shapes had a path width of 50 pixels and as a subject moved the cursor along the track, the “desired” or ideal position for the cursor was defined as the point on the track closest to the actual cursor position at each instantaneous sample point. After a 10-minute practice period, subjects were tested using a random order of joysticks, algorithms, and shapes for a total of 48 trials. We sampled at a rate of 17 Hz, to ensure sampling rate greater than physiologic tremor frequency, and the following outcomes were recorded. Trial Time: The time in seconds from the beginning to end of the trial. Offset: the average deviation of sample points from the path in pixels. Error: the average deviation of the absolute value of the distance of the sample points from the path in pixels. Variability: the standard deviation in distances of the sample points from the mean distance of sample points. These outcome measures were adapted from prior work by MacKenzie, et al[57].

Statistical analyses were performed using R[49]. All alpha levels were set to 0.05. We used Mixed Model Analysis to evaluate for differences among control algorithm conditions with respect to performance outcome variables. We used log (for MSJ data) and inverse (for IJ data) transformations on all variables to correct for skewed distributions. We also used this model to evaluate for an effect on outcomes measures due to tremor or Barthel Index scores (higher score indicates higher functional ability).

1.5.4 Results

Fifteen subjects were recruited. Average age was 52.1+/-9.3 years. Three subjects were female. Diagnoses included MS (n=7), cervical SCI (n=6), Wilson's disease (n=1), and PD (n=1). One subject with SCI also had a TBI. Ten subjects were regular computer users.

Results for the MSJ are listed in Table 5. There were no significant differences in trial time depending on algorithm applied. The PID produced lower offset compared to the LMS ($p=0.004$) and no algorithm ($p<0.0001$). The PID and LMS both produced lower error than no algorithm ($p<0.0001$ for both) and lower variability than no algorithm ($p<0.0001$ for both).

Table 5: Performance Outcome Measures for Motion Sensing Joystick per Algorithm

	No Algorithm		LMS		PID	
	mean	sd	mean	sd	mean	sd
trial time (s)	5.45	2.04	5.66	2.57	5.68	2.30
offset (pixels)	6.29	3.80	4.01	2.10	3.48	1.36
error (pixels)	7.97	4.88	4.91	2.66	4.37	2.05
variability (pixels)	7.18	4.49	4.21	2.04	3.95	1.79

Results for the IJ are listed in Table 6. The PID and LMS both produced slightly longer trial times than no algorithm ($p=0.04$ for both). The PID and LMS produced lower offset than no algorithm ($p<0.0001$ for both), lower error than no algorithm ($p<0.0001$ for both), and lower variability than no algorithm ($p<0.0001$ for both).

Table 6: Performance Outcome Measures for Isometric Joystick per Algorithm

	No Algorithm		LMS		PID	
	mean	sd	mean	sd	mean	sd
trial time (s)	5.80	2.60	6.39	2.53	6.25	2.81
offset (pixels)	5.18	3.57	3.58	2.86	3.57	3.06
error (pixels)	6.41	4.70	4.32	3.67	4.29	3.90
variability (pixels)	5.88	4.23	3.90	3.50	3.78	3.28

For the MSJ, Barthel index was not associated with any outcome variables. For the IJ, higher Barthel score was associated with longer trial time (p=0.046) and lower offset, error, and variability (p=0.003 to 0.004). Presence of tremor did not have a significant effect on any of the outcome measures.

1.5.5 Discussion

Trial time may be slightly longer for those using the IJ with control algorithms because the algorithms kept the cursor trajectory closer to the ideal trajectory and discouraged subjects from “cutting corners” to complete tasks more quickly. Because the time difference was small, this is likely to be clinically insignificant. This effect was not seen with the MSJ, possibly because some of the subjects already had some skill using an MSJ for PWCs and may have had better fine motor control with this device. However, subjects performed well with both joysticks and clearly had better performance when the algorithms were applied compared to using no algorithm at all.

2.0 CONCLUSIONS AND FUTURE DIRECTIONS

2.1.1 Conclusions

Historically, isometric controls were thought to have limited application for individuals with movement disorders due to their sensitivity to unintentional movements. The work in this thesis is the first to demonstrate the real potential of isometric controls for use in this population. Subjects with various upper limb impairments can perform just as well with isometric controls as with conventional proportional control, and individuals with tremor actually perform better with isometric controls. While filters appropriate for physiologic tremor may not attenuate pathological tremor, the inherent design of isometric controls may make them an exciting alternative control for this population. Although not necessarily intuitive to use, IJs probably have a shorter learning curve than we once thought. While the propensity to experience fatigue due to excessive force production remains a significant obstacle for their use in populations prone to fatigue or with weakness, additional training with these devices may be sufficient to promote appropriate force control. Finally, further customizing either proportional or isometric controls with other types of advanced control algorithms is proving to be the key to addressing errors in performance due to movement disorders.

2.1.2 Future Directions

We have collected large amounts of pilot data in individuals with athetoid and spastic CP, TBI, tremor, and in control subjects, and future work is aimed at comparing tuning parameters to identify overarching themes that are common to each diagnostic group. Identification of such parameters would allow for development of a series of default tuning parameters that could be available as software packages for particular diagnostic groups and which then could be further customized according to individual user needs. We are currently working on analyzing our vast amounts of data to garner a better understanding of driving styles of different individuals. Some individuals drive with continuous, smooth trajectories. Some individuals drive with pulse width modulation (on/off) type control strategies. Understanding the techniques of drivers will allow us to design better interfaces that meet their needs. We also plan to update the electronics in our current IJ.

One useful addition to the IJ may be haptic feedback in the form of vibration or audio signals that indicate when the user has produced force to about half of the template maximum, or when they are exceeding the threshold. This may allow gain to be adjusted to that users do not need to apply as much force and may counteract some of the effects that are due to limiting proprioceptive feedback.

It may also be worthwhile to consider using a joystick with variable compliance as a training tool, that is, in transitioning an individual from an MSJ to IJ. Future studies could evaluate whether varying compliance as a transitional rehabilitation tool is useful and which diagnoses derive the best benefit.

Future research will focus on creating a "smart filter" that could continuously monitor the vibration frequencies from various sources and adjust accordingly. Previous research[58] on dynamic coupling principles might also be applied to help overcome the effects of frequencies that are transmitted to the hand when EPWs are driven over rough terrain.

It is possible that in the future, people will be able to use a single device for not only their computers and wheelchair driving but also environmental control of many target devices like garage doors, keyless entry, home lighting, heating and air conditioning thermostats, and a universal remote for home entertainment products; portable augmentative communication aids; or alternative methods to operate a mobile phone. The development of integrated controls would offer third party payers the option of funding single multifunctional controls rather than separate, product-specific devices and would also justify more rehabilitation services to customize the device to the consumer.

Of course, as our control interfaces advance, so must the controllers. Controllers contain the electronics and software, modify the signals from the interface, and convert them to output which is passed along to the device being controlled. As more complex algorithms are developed, the need for more advanced controllers that can handle the mathematics will increase. Most current EPW controllers also ignore the wheelchair dynamics, making them unable to ensure the same performance with variations in loads or terrain. Cooper, et al. are developing an advanced controller based on a kinematic and dynamic model of a wheelchair (NIH R03 HD048465-01) that will be able to be augmented with advanced algorithms and that will accommodate a variety of interfaces.

Another worthwhile direction of future research is developing a device for recording the signals generated from control interfaces—not only joysticks, but also other input devices like switches. Currently, there is no technology that allows clinicians to observe and analyze the signals generated by the user. Access to such data would allow clinicians to custom tune the interface, identify algorithms that could be applied (e.g. tremor filter), identify targets for training (e.g. excessive use of force on a joystick that results in arm fatigue), or identify efficacy of treatment on functional use of an interface (e.g. medications or interventional injections for spasticity). We aim to design and develop a technology (I-Log) that can record and analyze input signals. The I-Log will include a low power microcontroller and a real time clock with a large capacitor backup. It will write data to a Secure Digital Flash disk which will also serve as its interface and data transfer. The firmware will record the time and log the input signals (e.g. displacement, force, switch activation) from the EPW interface. In the case of switch-based systems, the I-Log will simply log the time period and frequency that a signal was active. Host software will translate data written on the SD card to a format readable by the Analysis/Interpretation software. It will also communicate with the I-Log to configure its settings. Analysis/Interpretation Software will convert bits to motion or switch activations, display data in meaningful ways, perform standard analyses, and provide a useable graphical interface that shows, for example, the range of inputs used as a function of time, and allows for recommending bias axes, template, dead zone, gain settings, filter settings, and algorithm parameters to optimize the interface's output.

Developing better control interfaces that can be used despite motor impairments and movement disorders may improve computer access, mobility, community interaction, and ultimately quality of life[2].

APPENDIX A

LMS AND PID CONTROL ALGORITHMS

Our first algorithm is a simple adaptive **least mean squares (LMS) type algorithm**[52-55] which works by correcting trajectory using the difference (or error) between the desired and/or the actual speed and direction signals. These parameters are assumed to be independent for the purposes of the adaptation algorithm. This algorithm can be used to correct the trajectory of a sprite that a user directs by tracing a stationary shape on the screen, or alternatively, attempting to keep the sprite in a track box that is moving along a track on the screen at a given speed. In the study described in this thesis, the screen shape was stationary; thus only the direction signal was used in the algorithm. The desired or ideal location was defined as the closest point on the track with respect to the user's actual cursor position. However, for the purposes of completeness, equations are shown for both speed and direction signals.

Given these assumptions, the adaptive LMS algorithm will take the following form:

For speed control:

$$e_{Speed}(k) = P_{Fore\_Aft\_T}(k) - P_{Fore\_Aft\_S}(k) \tag{1}$$

$$K_S(k+1) = K_S(k) + \mu_{Speed} \bullet e_{Speed}(\mathrm{k}) \tag{2}$$

$$.99 \geq \mu_{Speed} \geq 0.01 \qquad K_{S\_\max} \geq K_S \geq K_{S\_\min} \tag{3}$$

Where $P_{Fore\_Aft\_T}$ is the fore-aft position (ahead or behind) of the trackbox, $P_{Fore\_Aft\_S}$ is the fore-aft position (ahead or behind) of the sprite, and e_{Speed} is the error signal between the center of the trackbox and the center of the sprite with respect to speed (moving ahead or falling behind). In Equation (2), μ_{Speed} is the update weight of the speed gain (K_S). It is tuned during the off-line testing to determine the values of μ_{Speed} to use during on-line tuning and to set initial values for K_S. Each sampling time is represented by k, and $k + 1$ is the next sampling time.

For directional control a similar update algorithm will be used:

$$e_{Direction}(k) = P_{Lateral\_T}(k) - P_{Lateral\_S}(k) \tag{4}$$

$$K_D(k+1) = K_D(k) + \mu_{Direction} \bullet e_{Direction}(\mathrm{k}) \tag{5}$$

$$.99 \geq \mu_{Direction} \geq 0.01 \qquad K_{D\_\max} \geq K_D \geq K_{D\_\min} \tag{6}$$

In Equation (4), $e_{Direction}$ is the directional error signal between the desired location on the trajectory and the sprite. The position (lateral) gain update weight $\mu_{Direction}$ is used to adapt the position gain (K_D).

The second algorithm, a **proportional-integral-derivative (PID) algorithm**[51] is a method of correcting error in a trajectory using feedback control. It works by measuring differences between desired location and actual sprite location, the degree to which the movements overshoot or undershoot the desired location, and the degree of oscillation about the trajectory. The algorithm then uses the weighted sum of these three actions to adjust the output of the device. By "tuning" the three parameters in the PID algorithm an investigator can fine tune the controller the individual user. The PID is given by:

$$MV(t) = P_{out} + I_{out} + D_{out}$$

The manipulated variable (MV) here is the sprite location. The proportional term makes a change to sprite location that is proportional to the sprite's positional error between the desired position and the sprite. The proportional term is given by:

$P_{out} = K_p e(t)$ where

P_{out}: **Proportional output**

K_p: **Proportional Gain**, a tuning parameter

e: **Error** between sprite and desired location

t: **Time** or instantaneous time (the present)

The contribution from the integral term is proportional to both the magnitude of the error and the duration of the error. Summing the instantaneous error over time (integrating the error) gives the accumulated offset that should have been corrected previously. The accumulated error is then multiplied by the integral gain and added to the controller output. The magnitude of the contribution of the integral term to the overall control action is determined by the integral gain,

K_i. The integral term is given by:

$I_{out} = K_i \int_0^t e(T)dT$ where

I_{out}: **Integral output**

K_i: **Integral Gain**, a tuning parameter

e: **Error**

T: **Time** in the past contributing to the integral response

The rate of change of the process error is calculated by determining the slope of the error over time (i.e. its first derivative with respect to time) and multiplying this rate of change by the derivative gain K_d. The magnitude of the contribution of the derivative term to the overall control action is termed the derivative gain, K_d.

The derivative term is given by:

$D_{out} = K_d \frac{de}{dt}$ where

D_{out}: **Derivative output**

K_d: **Derivative Gain**, a tuning parameter

e: **Error**

t: **Time** or instantaneous time (the present)

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