

The Optimal Controller Delay for Myoelectric Prostheses

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Abstract—A tradeoff exists when considering the delay created by multifunctional prosthesis controllers. Large controller delays maximize the amount of time available for EMG signal collection and analysis (and thus maximize classification accuracy); however, large delays also degrade prosthesis performance by decreasing the responsiveness of the prosthesis. To elucidate an “optimal controller delay” twenty able-bodied subjects performed the Box and Block Test using a device called PHABS (prosthetic hand for able bodied subjects). Tests were conducted with seven different levels of controller delay ranging from nearly 0–300 ms and with two different artificial hand speeds. Based on repeated measures ANOVA analysis and a linear mixed effects model, the optimal controller delay was found to range between approximately 100 ms for fast prehensors and 125 ms for slower prehensors. Furthermore, the linear mixed effects model shows that there is a linear degradation in performance with increasing delay.

Index Terms—Box and Block Test, delay, myoelectric, myo-pulse control, prostheses, prosthesis, prosthetics.

I. INTRODUCTION

THE ideal multifunctional prosthesis is one that will quickly and accurately respond to the commands of the user. Multifunctional prosthesis controllers have shown higher classification accuracies when electromyographic (EMG) feature extraction and pattern recognition are performed on time windows of longer duration [1]–[3]. Unfortunately, there is a limit to the time over which EMG data can be collected and analyzed before the controller delay (i.e., the amount of time between the user’s command and the actuation of the device) resulting from this collection and analysis causes the control of the prosthesis to become cumbersome.

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In fact, Hogan [4] specifically described a problem with the control of prosthetic elbows that he attributed to large controller delays. While he did not specify the length of the delay, Hogan observed an oscillatory behavior that was created when a large controller delay caused users to repeatedly overshoot the desired elbow position and then have to compensate for this overshoot. Kyberd, *et al.* [5] found that, for high-speed devices, the controller delay made it very difficult to open a prosthetic hand partway. The large controller delay caused the hand to open fully by the time a “stop” command could be generated. Otto Bock’s (Duderstadt, Germany) new Sensor Hand Speed has a speed of opening and closing of 4 radians per second but also exhibits a similar lack of responsiveness due to the long filter time constants of the Otto Bock electrodes.

The goal of this experiment was to define the “optimal” delay for a prosthesis controller. We define the optimal delay as the maximum amount of time that can be used by the controller for data collection and analysis (so as to maximize classification accuracy) without affecting quantitatively observed prosthesis performance.

II. BACKGROUND

Whereas the effects of delays on performance in remote manipulation tasks and virtual environments have previously been examined [6]–[14], Paciga *et al.*’s work [15] is the only study the authors are aware of that objectively attempted to quantify the effect of a controller delay on prosthesis performance. Paciga, *et al.* examined the ability of users to produce five different levels of an EMG signal for a five-state prosthesis controller. It was shown that introducing a 200 ms delay into the visual feedback path provided to the subject would increase the measured errors six-fold from 1.1% to 6.6%.

Whereas relatively little work has objectively examined the impact of controller delays on prosthesis performance, the length of controller delay that can exist before prosthesis control degrades has been debated. Childress and Weir [16] believe that controller delays should be no longer than 50 ms to optimize performance and ensure these delays are imperceptible to the user. This is based upon anecdotal evidence from users that felt as if their hands “were being operated in molasses” if larger controller delays were present. However, members of a group at the University of New Brunswick (UNB) state that delays as large as 300 ms are not perceivable by the user [17], [18] and are acceptable for prosthesis control. Graupe *et al.* made contradictory statements in two of their efforts. In an early work, they stated that delays of greater than 200 ms are unacceptable [19] and in a later paper they stated that delays of 100–300 ms are satisfactory [20]. Additionally, Hefftner [21] *et al.* stated that controller delays must be kept below

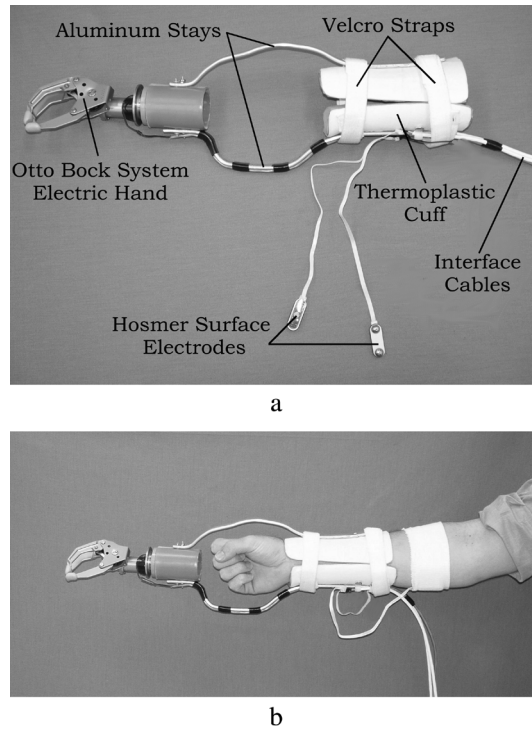


Fig. 1. (a) Photograph of PHABS detailing the components of the device. (b) Photograph of an individual wearing PHABS.

300–400 ms to prevent a noticeable delay. Chu, *et al.*, [22] designed a real-time controller and attempted to keep the controller delay below 300 ms so as to not affect performance. In summary, the existing literature shows that an eight-fold (50–400 ms) difference exists in the published estimates of the acceptable delays for prosthetic limb systems.

The length of acceptable controller delay may affect what signal processing and pattern recognition algorithms are used by multifunctional prosthesis controllers and thus it would be beneficial to establish this value for future investigations. Therefore, experiments were designed to discover the longest period of controller delay that does not significantly affect prosthesis performance and can thus be dedicated to EMG collection and analysis.

III. HARDWARE AND PROTOCOL

A. PHABS

A bypass prosthesis called “PHABS” (Prosthetic Hand for Able-Bodied Subjects) was created to allow able-bodied subjects to operate a prosthetic terminal device. PHABS consists of a plastic cuff with adjustable straps and two aluminum stays that connect the user to the terminal device (Fig. 1). Two Hosmer surface electrode units (Hosmer Dorrance Corporation, Campbell, CA) were placed on the proximal third of the forearm over the flexor and extensor muscle bellies. The Hosmer electrodes perform very little filtering while amplifying the signal. The electrodes use capacitive decoupling to eliminate any direct current component of the signal and have a low pass cutoff frequency of 1500 Hz. The electrodes were fixed in place using Compressogrip #21/2 (Knit-Rite, Inc., Kansas City, KS) compression bandage that was then covered in standard kitchen plastic wrap to keep the electrode contact areas moist. Fig. 1 shows PHABS

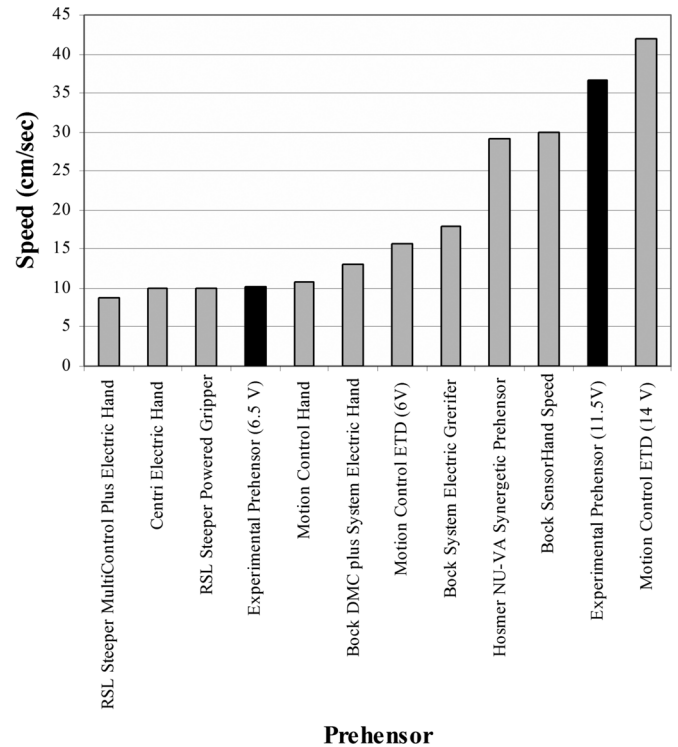


Fig. 2. Linear speeds of all commercially available prostheses sold in the U.S. [23], [24], as well as the measured speeds for both the “fast” and “slow” prehensors used in these experiments. There are three prehensors that are marginally slower than our “slow” prehensor and the Motion Control ETD (when operated at 14 V) is the only device that has a higher speed than the “fast” prehensor. Based on this data it is believed that the spectrum of commercially available prehensor speeds was fairly well represented in these experiments.

equipped with an Otto Bock System Electric Hand (Otto Bock Healthcare, Duderstadt, Germany).

For these experiments, PHABS was equipped with the Hosmer NU-VA Synergetic Prehensor (Hosmer-Dorrance Corp., Campbell, CA). This prehensor was chosen because of its high speed of opening and closing. One of the goals of these experiments was to investigate the effect of prehensor speed on the optimal controller delay. The supply voltage to the Synergetic Prehensor was varied to adjust the speed of the device. Supply voltages of 6.5 and 11.5 V were used (+/−2.5 V from the standard operating voltage). The time from full open to full close for the 6.5 V and 11.5 V supply voltages were 430 ms and 240 ms, respectively. The speeds of the prehensor were calculated to be 10.2 cm/s (2.15 radians/s, 123 degrees/s angular velocity) for the 6.5 V supply voltage and 36.7 cm/s (3.85 radians/s, 221 degrees/s angular velocity) for the 11.5 V supply voltage. For a comparison of these speeds against the speeds of the other prehensors that are commercially available in the U.S., see Fig. 2. The fastest prehensors on the market are the Sensor Hand Speed from Otto Bock (Otto Bock Healthcare, Duderstadt, Germany) and the Motion Control ETD (when run at 14 V) (Motion Control, Salt Lake City, UT) which are reported to have an average maximum speed of 30.0 cm/s [23] and 41.9 cm/s [24], respectively. The slowest commercially available prehensor is the RSL Steeper Multicontrol Plus Electric Hand (RSL Steeper, Leeds, U.K.) which has an average maximum speed of 8.75 cm/s [24]. The data in Fig. 2 shows that the prehensor speeds that were used in these experiments

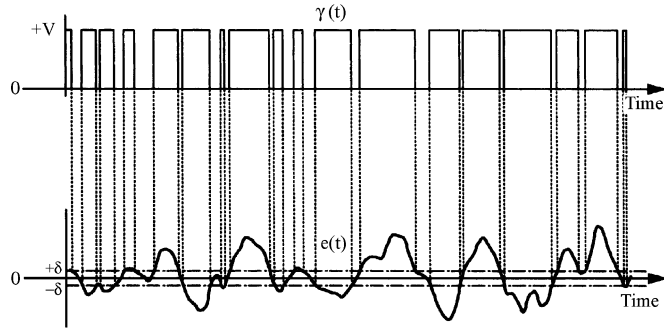


Fig. 3. Graphical illustration of myo-pulse control. Myo-pulse control provides proportional control of a motor by varying the pulse width and timing of a digital control signal. Whenever the EMG signal (bottom trace) is lies outside a predetermined threshold, the motor drive signal (top trace) is turned “on.” The ratio of motor “on” command time to motor “off” time determine the velocity of the motor. Used with permission of the Northwestern University Prosthetics Research Laboratory (NUPRL).

are representative of the two ends of the spectrum of prehensor speed available in current commercial devices.

B. Controller

1) *Theory and Implementation:* Myo-pulse control ([25], [26]) is a commercially available control strategy (Hosmer–Dorrance Corporation, Campbell, CA) and, while it is infrequently used in clinical practice, it was chosen as the control algorithm for these experiments because it provided an easy way to both minimize the delay inherent in the controller and to set the controller delay to any value above this minimum. Myo-pulse control can be thought of as a combination of both pulse width modulation (PWM) and pulse period modulation (PPM) because it provides proportional control of motor speed by varying the pulse width and timing of a digital control signal. The prehensor’s motor and the inertia of the fingers then act as a mechanical low-pass filter to smooth the pulse train and create a smooth movement.

A myo-pulse controller compares the absolute value of the EMG signal to a predefined threshold. If the EMG is above the threshold the motor will be turned “on” and if it is below the threshold the motor will be turned “off.” A graphical example of one-channel myo-pulse control is shown in Fig. 3.

If the EMG signal is treated as a Gaussian signal whose variance is related to the strength of the contraction, as the strength of a contraction increases so does the probability that the amplitude of the EMG will lie outside the thresholds and thus increases the amount of time the signal is “on.” Variable speeds of opening and closing are created by the ratio of “on” time to “off” time.

In typical EMG processing, the EMG is rectified and low-pass filtered to provide the envelope of the EMG signal as the input to the controller. Low-pass processing methods introduce a substantial delay between EMG onset and controller output due to the long time constants of these enveloping filters. Myo-pulse modulation avoids these delays by allowing the inertia inherent in the prehensor mechanism to perform the filtering, which makes for a very responsive EMG control system. In fact, the delay introduced by the controller is limited to its sampling period. For our purposes, myo-pulse control allows the inherent

controller delay to be kept as close to zero as possible to create an ideal baseline condition.

For these experiments the controller was created in Matlab’s (The Mathworks, Inc., Natick, MA) Simulink and executed using Simulink’s Real Time and XPC Target Toolboxes. The myo-pulse controller was executed at 1 kHz and thus the inherent controller delay was only 1 ms when no additional artificial delay was added. The Simulink software allowed artificial controller delays to be added to the controller and variable myo-pulse thresholds to be implemented.

2) *Myo-Pulse Creates the Ideal Baseline Condition:* Fig. 4 shows the elements that could introduce a delay in a multi-functional prosthesis control system. These delays include: the amount of time it takes from the intent of movement to the development of EMG, the time constant of the analog filters contained in the EMG preamplifiers, the analog to digital sampling period, the time required to collect the EMG signal for feature extraction, the time required to perform the EMG feature extraction, the time required to execute the pattern recognition on the extracted features and the time required to actuate the component. By implementing a myo-pulse controller, the delays associated with analog low-pass filtering, EMG signal collection, feature extraction, and pattern recognition were eliminated. Additionally, the combination of the A/D sampling period and signal processing has been reduced to 1 ms. Therefore, for the baseline condition, our controller delay (time from EMG development to motor drive signal) is only 1 ms.

In these experiments up to 300 ms of additional delay was artificially added to the myo-pulse controller. This additional delay can be thought of as representing any combination of the variables that are eliminated by using the myo-pulse controller (e.g., EMG signal collection, analog filter time constant).

Admittedly there is some delay between the intent of movement and the development of EMG. For example, Bereitschaftspotentials, or very preliminary neurological precursors of movement, can appear up to 1.5 s prior to voluntary finger movement [27]. In these experiments, we are not investigating or attempting to control the neurological component shown in Fig. 4.

Finally, as mentioned previously in the description of PHABS, we will be examining the effect of the time necessary for the actuator to produce a movement by varying the speed of the prehensor.

C. Box and Block Test

The Box and Block Test was used to quantify prosthesis performance. This test was chosen because it is quantitative, quick and easy to administer, sensitive [28], and normative data has been collected [29], [30]. It has been used frequently to quantify the effect of treatment on upper limb function for disorders such as cerebral palsy [31], multiple sclerosis [32], and stroke [33]–[35].

The testing apparatus for the Box and Block Test (Sammons Preston Inc., Bolingbrook, IL) consists of two adjacent compartments separated by a 6-in barrier (Fig. 5). The Box and Block Test is a 60-s timed test in which subjects are instructed to pick up blocks from one compartment, transport them across the barrier, and release them in the opposite compartment as quickly as possible. Several rules govern the scoring of the test, i.e., blocks thrown across the barrier do not count, if two blocks are moved

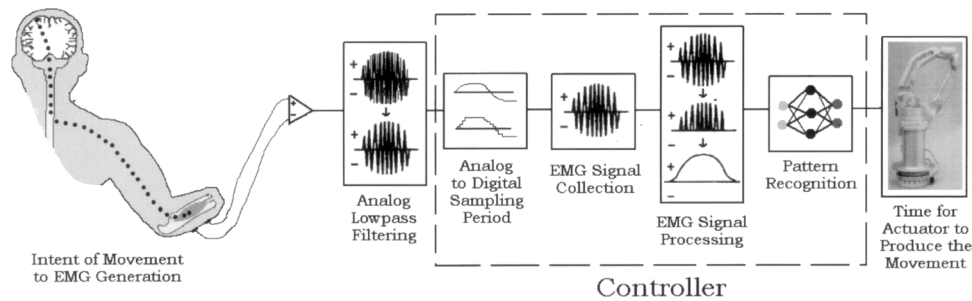


Fig. 4. Elements composing the time from the intention of movement to the completion of the movement by the prosthesis.

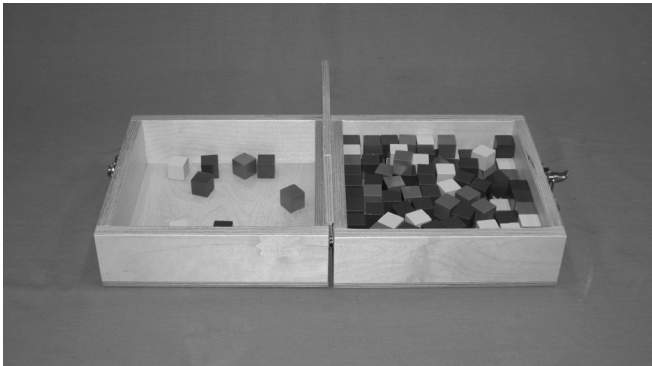


Fig. 5. Photograph of a Box and Block apparatus. Subjects will pick up blocks from one side of the box, transport them over the barrier, and then release them on the other side of the box. The number of blocks transported over the barrier in 1 min is the score that the subjects receive.

across the barrier together they only count as one, etc. For more details on these rules and the test itself, see [30].

D. Testing Protocol

1) *Pretest Procedures:* Each testing session began with an explanation of the test procedures and the acquisition of informed consent from the subject. For simplicity, the two prehensor speeds will now be referred to as “fast” and “slow.” Half of the subjects were given the slow prehensor speed first and half were given the fast speed first. The first eighteen subjects flipped a coin to determine which prehensor speed would be used during the first set of trials. The last two subjects were simply assigned the speed that they would complete first to ensure that an equal number of subjects used each speed first.

Two EMG sites were located on the forearm over the wrist flexor and extensor groups according to standard clinical prosthetics practice. This procedure involves having subjects repeatedly produce contractions of similar intensity and then iteratively moving a testing electrode around the flexor or extensor surface of the forearm. The different sites are examined in an effort to find the two sites that provide the strongest signal for the test movement (i.e., wrist flexion or extension) without substantial coactivation from the opposite movement or interference from the brachioradialis, which is used to support the weight of the PHABS.

The subjects donned the PHABS with the electrodes placed at the previously identified locations. Threshold levels in the

myo-pulse controller were altered so that they were as low as possible without allowing the baseline noise contained in the EMG channels to elicit movement of the prosthesis.

The subject was then allowed time to practice picking up several blocks and transporting them across the barrier. An Otto Bock Quick-Disconnect Wrist (Otto Bock Healthcare, Duderstadt, Germany) allowed the amount of pronation/supination of the terminal device to be altered to a position that was preferred by the subject to increase visibility, correlate movement of the anatomic wrist to the prosthetic finger, etc. Once the subject felt comfortable controlling the PHABS the instructions for the Box and Block Test were read to them and the subjects then completed two 1-min practice trials of the Box and Block Test.

2) *Test Administration:* Seven levels of additional controller delay (i.e., delay that was added to simulate EMG collection and processing) were investigated in this study, ranging from 0 to 300 ms in 50 ms increments.

It was observed that there was a substantial improvement in the Box and Block scores during the first several trials as the subjects developed strategies as to how to best complete the Box and Block Test and became more comfortable controlling the PHABS. In all but one of the studies listed earlier, the Box and Block Test was only administered once per testing session. One study [28] had each subject perform the test three times per sitting but made no comments regarding any improvement in performance from the first to the third trials.

As a result of this learning effect each subject performed seven trials at the beginning of testing (trials #1–7) that were timed and recorded but not included in the final analysis. Each of the seven delay levels was introduced once within the first seven “ignored” trials. The subject then performed three more groups of seven trials (trials #8–28) in which each delay level was randomized and presented once per group (three times total). The randomization of the trials was done to ensure that any further learning/fatigue effects were distributed equally across all conditions.

Each of the first 28 trials was conducted using the hand speed that was determined by the coin flip. After trial #28 the speed of the hand was switched and the subject was allowed to do two practice trials to become familiar with the new prehensor speed (trials #29–30). After completing the first 28 trials, the subjects were found to adapt quickly to the new speed of the prehensor. The subject then performed another three sets of seven trials (trials #31–51) to complete the protocol. Subjects were required to take a 5-min break after every 10–12 trials to reduce the effects of fatigue.

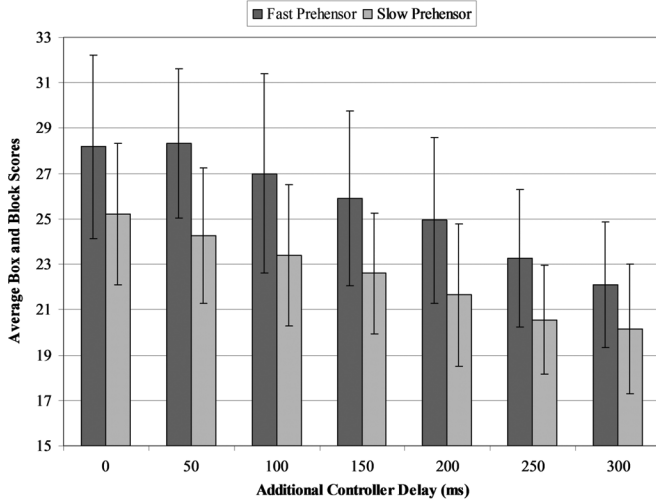


Fig. 6. Average Box and Block test scores ($n = 20$) when using the “fast” and “slow” prehensors. As the additional controller delay increases, the average scores on the Box and Block Test decreases. Additionally, as expected, subject performed better when using the “fast” prehensor.

IV. RESULTS

A. Box and Block Scores

Twenty subjects participated in the study and the average Box and Block scores for all subjects are shown in Fig. 6. As expected, the “fast” prehensor had higher scores, on average, than the “slow” prehensor. Also as expected, as the amount of controller delay increases the average Box and Block score tends to decrease. The only exception to this rule is the slight increase from 0 to 50 ms with the “fast” prehensor condition.

B. Statistical Analysis

1) *Repeated Measures ANOVA*: A single-factor repeated measures ANOVA was used to compare statistical intrasubject differences in the average Box and Block scores for the various controller delays. The null hypothesis was that there was no difference between the delay conditions. Additionally, a Bonferroni correction for multiple comparisons was made.

The data passed Mauchly’s Test of Sphericity for both the “slow” data ($p = 0.950$) and the “fast” data ($p = 0.174$) meaning that the variance of the data for each of the conditions can be considered equal and is, therefore, adequate for repeated measures ANOVA analysis. Additionally, the data for all of the conditions passed the Shapiro–Wilk test of normality ($p > 0.05$), ensuring that the distribution of the data was appropriate for an ANOVA analysis.

The results of the repeated measures ANOVA are presented in Fig. 7. Those comparisons that produced statistically significant differences ($p < 0.05$) are highlighted in gray. The most important comparisons to examine are those between the smallest controller delays, i.e., 0 and 50 ms and the larger delays. There is no statistically significant difference between the 0/50 ms delay conditions and the 100 ms condition but there is a difference between the 0/50 ms conditions and all controller delays of 150 ms or larger. These differences existed for both the “fast” and “slow” prehensors. All of the comparisons produced the same results for both prehensor speeds with the exception

Slow Prehensor						
Delay Level (ms)	50	100	150	200	250	300
0	1.000	0.070	0.001	0.000	0.000	0.000
50		1.000	0.002	0.000	0.000	0.000
100			1.000	0.098	0.000	0.000
150				0.680	0.006	0.000
200					0.999	0.025
250						1.000

Fast Prehensor						
Delay Level (ms)	50	100	150	200	250	300
0	1.000	1.000	0.010	0.000	0.000	0.000
50		1.000	0.001	0.000	0.000	0.000
100			0.594	0.007	0.000	0.000
150				0.051	0.002	0.000
200					0.037	0.000
250						0.265

Fig. 7. Results of the repeated measures ANOVA analysis on the Box and Block data. Each cell represents the probability that there is no difference between the two conditions. Those combinations that produce a statistically significant result ($p < 0.05$) are highlighted in gray.

of the 100 ms versus 200 ms and 200 ms versus 250 ms levels. These differences were not statistically significant for the slow prehensor but were statistically significant for the fast prehensor.

2) *Linear Mixed Effects Model*: In addition to attempting to determine simply *statistical* differences between the different conditions, a linear mixed effects model was constructed to determine a *clinically* significant change in the Box and Block scores. This type of model does not attempt to determine the presence of differences between experimental conditions but instead attempts to create a model that best explains the observed data. From this model, the amount of controller delay necessary to produce a change of n blocks in the Box and Block Test can be calculated.

Prior to beginning the study, we determined that a change of three blocks is a clinically significant change in a Box and Block score. This estimate was based off of the results from our pilot experiments, where a change in score of three blocks was equal to a 12.4% change from the overall average score of all trials.

The linear mixed effects model includes both standard fixed effects (i.e., intercept, controller delay, prehensor speed, and an interaction term) as well as random effects to account for inter-subject differences and takes the following form:

$$\text{Score} = \beta_0 + b_j + (\beta_1 + b_{1j}) * \text{Delay} + (\beta_2 + b_{2j}) * \text{Speed} + (\beta_3 + b_{3j}) * \text{Delay} * \text{Speed} + e \quad (1)$$

Score	Score on the Box and Block test.
Delay	The delay added to the controller [0, 50, 100, 150, 200, 250, 300].
Speed	Speed of the prehensor [10.2 or 36.7 m/s]
b_{xj}	Random effects, i.e., the expected differences of the intercept, delay, speed and interaction coefficients from the population means, given a particular subject j (1-20).
e	Residual effects.

The null hypothesis was that each of the β coefficients in (1) was equal to zero.

All coefficients were found to be statistically significant (β_0 , $p > 0.005$; β_1 , $p > 0.005$; β_2 , $p > 0.005$; β_3 , $p = 0.018$). If

the calculated β values are entered into (1) the final model takes the form

$$\text{Score} = 23.69 - 0.0157 * \text{Delay} + 0.143 * \text{Speed} - 0.000163 * \text{Delay} * \text{Speed}. \quad (2)$$

A statistically significant interaction term implies that the amount of delay required to cause an n block change in the score will be dependent upon the prehensor speed. For a given prehensor speed, the change in delay that would be required to change the Box and Block Test score by n blocks can be determined by

$$\Delta \text{Delay} = \frac{\Delta \text{Score}}{0.0157 + 0.000163 * \text{Speed}}. \quad (3)$$

For a fast prehensor (Speed = 36.7 cm/s), a three block change can be obtained with a delay of 138 ms. For the slow prehensor (Speed = 10.2 cm/s) a three block difference will be produced with a delay of 173 ms. These results show that users can tolerate more delay with slower devices but at the expense of lower overall performance.

The 138 ms and 173 ms values are estimates of the delays that cause a three block change in the performance for an *average* subject i.e., 50% of the subjects show a three block change for a 138 ms (fast prehensor) or 173 ms (slow prehensor) controller delay. When designing a controller, it should be effective for the majority of users, rather than the average subject. To obtain an estimate of the optimal delay for 90% of the population a model was constructed for each individual subject. The delay required to create a change of three blocks for the fast prehensor was observed to range from 80 to 268 ms for the 20 subjects. **Based on these data, we were then able to conclude that 90% of the population (mean + 1.282 SD) would show a change of less than three blocks when a delay of 99.8 ms (which we round up to 100 ms) was present. Likewise, for the slow prehensor, it was found that 90% of the population would show a change of less than three blocks with a 123 ms controller delay (which we round up to 125 ms).**

V. DISCUSSION

A. Average Box and Block Scores

The average Box and Blocks scores show that there is a decreasing trend in Box and Block score as the amount of controller delay increases. A linear relationship was observed which is consistent with similar linear trends between performance and controller delay observed by others [6], [7].

This relationship is monotonic for the “slow” prehensor and broadly monotonic for the “fast” prehensor results. The one exception for the fast prehensor is the small increase in the Box and Block scores for the 50 ms condition when compared to the 0 ms condition. However, this increase was not statistically significant ($p = 1.000$). This “knee” may be due to the limitations in the human operator rather than the prosthesis controller.

The average Box and Block scores also indicate that the subjects tended to score higher with the faster prehensor. This difference was shown to be statistically significant by the linear mixed effects model (β_2 , $p < 0.0005$).

B. Statistical Analysis

1) *Repeated Measures ANOVA*: While there was a general trend of decreasing performance on the Box and Block test as

the controller delay increased, the difference between the delay conditions did not become statistically significant until the additional controller delays reached at least 150 ms. For both the “fast” and “slow” prehensors, there was a statistically significant difference between the two smallest controller delay increments (0 and 50 ms) and the 150 ms condition. Additionally, there was no difference between the 0 and 50 ms delay increments and the 100 ms delay condition.

The conclusion that can be made from these ANOVA results is that, on average, a delay of 100 ms does not cause a statistically significant decrease in performance on the Box and Block Test. However, a delay of 150 ms will cause of statistically significant decrease in performance when compared to the 0 ms condition. These results indicate that the controller delays should be kept between 100 and 150 ms for the average subject to avoid producing a statistically significant decrease in performance.

2) *Linear Mixed Effects Model*: The disadvantage of the ANOVA analysis is that, while it does tell us which conditions statistically differ from one another, it does not permit calculation of user-defined clinically significant effects of controller delay on performance. Additionally, the ANOVA does not take advantage of the obvious relationship that exists between delay levels. The linear mixed effects model demonstrated that, depending on the speed of the prosthesis, a prosthesis controller should have a delay of less than 100–125 ms to avoid producing any clinically significant change in performance on the Box and Block Test for 90% of the population.

Even though a statistically significant difference between the groups may not have been detected by the ANOVA until 150 ms of delay was present, since β_1 was significant the linear mixed effects model suggests that any delay will decrease the Box and Block Test score. Implications of this observation for single degree-of-freedom prosthesis performance is that smaller delays will improve performance on the Box and Block Test. However, in the case of pattern recognition based multifunctional prosthesis control, it has been shown that larger analysis windows (and thus larger delays) increase classification accuracy. Therefore, a balance must be struck between increased speed of response and increased classification accuracy when considering pattern recognition based multifunctional prosthesis controllers.

The coefficient of the speed term β_2 was positive, indicating that as the speed of the prehensor increases the Box and Block scores tend to increase. This is not surprising since the “fast” prehensor opens and closes more quickly.

The statistical significance of the interaction term β_3 indicates that the affect of controller delay on the Box and Block performance is a function of prehensor speed. Additionally, the interaction term β_3 was negative, which indicates that as the controller delay increases the difference in the scores for the fast and slow prehensors will decrease. The effect of the interaction term makes intuitive sense because as the controller delays are increased, the controller delay, and not the speed of the prehensor, will dictate the responsiveness of the device.

C. Applicability to Other Forms of Controllers

In these experiments, myo-pulse control was used to create a controller with minimal delay. Until the last decade or so

most commercially available prehensors have been relatively slow and therefore did not require a responsive controller. As faster devices were created, more responsive signal processing techniques have been implemented to control these devices. For example, Hosmer Dorrance's Synergetic Prehensor uses myo-pulse control whereas the Motion Control ETD uses an adaptive filtering technique developed by Park and Meek [36] that modifies the time constant of the EMG filter based upon the EMGs' rate of change. These relatively new processing methods have been shown to be effective means of providing quick, responsive control of high-speed prehensors.

The task that was performed in these experiments required the subject to open and close the terminal device as quickly as possible in order to move the maximum number of blocks in the allotted time. It is reasonable to assume that the terminal device was being operated in a bang-bang fashion with intense contractions being produced to drive the hand at top speed. Given this scenario, the authors perceive little advantage to using longer time constant filters to reduce signal variability because the input signals are all likely above the maximum velocity threshold. Therefore, we believe that our results using myo-pulse control with a pure delay are a good estimate of the results that would be obtained using a low-pass filter based controller and therefore these results are applicable to other forms of controllers.

D. Applicability to Multiple Degree of Freedom Systems

There is an apparent tradeoff when examining the results from this study in the context of currently evolving pattern recognition systems. Past research in pattern recognition systems has shown a clear advantage to using larger analysis windows to increase classification accuracies [1]–[3]. While no performance analysis has been done with one of these systems implemented in a clinical fitting on an amputee, one would infer that there would be an increase in prosthesis performance with increased classification accuracy. However, longer analysis windows also lead to increased controller delay, which should degrade performance.

We have shown a linear decrease in prosthesis performance with increasing controller delay. Englehart, *et al.* [1] showed a nonlinear increase in classification accuracy with increasing window length. The "knee" in the error rate versus record length occurs somewhere below 100 ms. This curve also shows diminishing returns with respect to classification accuracy for record lengths longer than 100 ms. Additionally, Zardoshti-Kermani, *et al.* [37] showed that using window sizes greater than 100 ms did not substantially increase the separability of EMG signal features.

These observations are encouraging when examined in the light of the results obtained in the preceding experiments. We have demonstrated that prosthesis performance can show a clinically significant decrease with a controller delays of 100 ms or greater and Englehart, *et al.*, has shown that there is relatively little classification accuracy gained when more than 100 ms of EMG is collected for classification purposes. Therefore, while 100 ms may be a slightly conservative estimate of the optimal controller delay in light of the potential classification accuracy increases with larger analysis windows, we assume that the optimal controller delay for multiple degree-of-freedom devices

utilizing pattern recognition will be similar to what we have reported here.

Finally, when extending these results to multijoint devices, the speed of different prosthetic components must be considered. While a 100–125 ms delay was found to be the optimal controller delay for the range of commercially available prehensors speeds, this value may be higher for larger inertia components such as prosthetic elbows. However, even if slower devices are found to have a larger acceptable delay, one of the degrees-of-freedom inevitably being controlled by the multifunctional controller would be the prehensor. Since there is no way to tell when an "elbow flexion" command would be given instead of a "hand open" command, we need to design for the most responsive component. Therefore the ideal controller delay for the prosthesis controller would still be 100–125 ms, unless separate controllers are used for each degree-of-freedom of the prosthesis.

E. Other Observations

The fact that delays of greater than 100 ms cause a decrease in performance is consistent with some clinical observations that have been made in our laboratory. When controller delays were greater than 50–100 ms users noted that it felt as if they were operating their prosthesis 'in molasses.' The data collected in these experiments reinforces our clinical experience that the delays should be kept at or below 100 ms.

Keeping controller delays below 100 ms is a *goal*. If reliable control of a multifunctional prosthesis can be demonstrated but only by using larger controller delays, then users may choose to have a sluggish device that gives them robust control over several degrees of freedom. However, our results show that any additional delay results in decreasing performance and, therefore, an ideal multifunctional prosthesis would accurately respond to the users' commands and do so as little time as possible.

VI. CONCLUSION

In an effort to define the "optimal controller delay," (i.e., the maximum amount of time that can be used by the controller for data collection and analysis to maximize classification accuracy without affecting prosthesis performance) twenty subjects performed the Box and Block Test with a variety of controller delays. We have shown that both a repeated measures ANOVA analysis and a linear mixed effects model analysis give us estimates of the optimal controller delay that lie between 100 and 175 ms for the average user. However, if the controller is designed to accommodate the 90th percentile user, we show that the optimal controller delay lies in the neighborhood of 100–125 ms, depending on the prehensor speed. In addition we show a linear decrease in operator performance for any additional delay.

REFERENCES

- [1] K. Englehart, B. Hudgins, and P. A. Parker, "A wavelet-based continuous classification scheme for multifunction myoelectric control," *IEEE Trans. Biomed. Eng.*, vol. 48, no. 3, pp. 302–311, Mar. 2001.
- [2] S. Du and M. I. Vuskovic, "Temporal vs. spectral approach to feature extraction from prehensile EMG signals," in *Proc. IEEE Int. Conf. Inf. Reuse Integration*, Nov. 8–10, 2004, pp. 344–350.
- [3] M. Yamada, N. Niwa, and A. Uchiyama, "Evaluation of a multifunctional hand prosthesis system using EMG controlled animation," *IEEE Trans. Biomed. Eng.*, vol. 30, pp. 759–763, 1983.

- [4] N. Hogan, "A review of the methods of processing EMG for use as a proportional control signal," *Biomed. Eng.*, vol. 11, pp. 81–86, 1976.
- [5] P. J. Kyberd, O. E. Holland, P. H. Chappell, S. Smith, R. Tregidgo, P. J. Bagwell, and M. Snaith, "MARCUS: A two degree of freedom hand prosthesis with hierarchical grip control," *IEEE Trans. Rehabil. Eng.*, vol. 3, no. 1, pp. 70–76, Mar. 1995.
- [6] J. E. Conklin, "Linearity of the tracking performance function," *Perceptual Motor Skills*, vol. 9, pp. 387–391, 1959.
- [7] W. R. Ferrell, "Remote manipulation with transmission delay," *IEEE Trans. Human Factors Electronics*, vol. HFE6, pp. 24–32, 1965.
- [8] W. R. Ferrell and T. B. Sheridan, "Supervisory Control of Remote Manipulation," *IEEE Spectrum*, vol. 4, pp. 81–88, 1967.
- [9] I. MacKenzie and C. Ware, "Lag as a determinant of human performance in interactive systems," in *Proc. ACM Conf. Human Factors Comput. Syst.—INTERCHI '93*, 1993, pp. 488–493.
- [10] T. B. Sheridan and W. R. Ferrell, "Remote manipulative control with transmission delay," *IEEE Trans. Human Factors Eng.*, vol. HFE4, pp. 25–29, Sep. 1963.
- [11] K. U. Smith, W. M. Smith, and M. F. Smith, *Delayed Sensory Feedback and Behavior*. Philadelphia, PA: W. B. Saunders and Company, 1962.
- [12] W. M. Smith, "Feedback: Real-time delayed vision of one's own tracking behavior," *Science*, vol. 176, pp. 939–940, 1972.
- [13] W. M. Smith and K. F. Bowen, "The effects of delayed and displaced visual feedback on motor control," *J. Mot. Behav.*, vol. 12, pp. 91–101, 1980.
- [14] R. Held, A. Efstathiou, and M. Greene, "Adaptation to displaced and delayed visual feedback from the hand," *J. Exp. Psychol.*, vol. 72, pp. 887–891, 1966.
- [15] J. E. Paciga, P. D. Richard, and R. N. Scott, "Error rate in five-state myoelectric control systems," *Med. Biol. Eng. Comput.*, vol. 18, pp. 287–290, 1980.
- [16] D. S. Childress and R. F. Weir, "Control of limb prostheses," in *Atlas of Amputations and Limb Deficiencies: Surgical, Prosthetic, and Rehabilitation Principles*, D. G. Smith, J. W. Michael, and B. J. H., Eds., 3rd ed. Rosemont, IL: J. H. Bowker, 2004, pp. 173–195.
- [17] K. Englehart and B. Hudgins, "A robust, real-time control scheme for multifunction myoelectric control," *IEEE Trans. Biomed. Eng.*, vol. 50, no. 7, pp. 848–854, Jul. 2003.
- [18] K. Englehart, B. Hudgins, and A. Chan, "Continuous multifunction myoelectric control using pattern recognition," *Technol. Disability*, vol. 15, pp. 95–103, 2003.
- [19] D. Graupe, J. Salahi, and K. H. Kohn, "Multifunctional prosthesis and orthosis control via microcomputer identification of temporal pattern differences in single-site myoelectric signals," *J. Biomed. Eng.*, vol. 4, pp. 17–22, 1982.
- [20] D. Graupe, J. Salahi, and D. S. Zhang, "Stochastic analysis of myoelectric temporal signatures for multifunctional single-site activation of prostheses and orthoses," *J. Biomed. Eng.*, vol. 7, pp. 18–29, 1985.
- [21] G. Heffner, W. Zucchini, and G. G. Jaros, "The electromyogram (EMG) as a control signal for functional neuromuscular stimulation—Part I: Autoregressive modeling as a means of EMG signature discrimination," *IEEE Trans. Biomed. Eng.*, vol. 35, no. 4, pp. 230–237, Apr. 1988.
- [22] J. K. Chu, I. Moon, and M. Mun, "A real-time EMG pattern recognition based on linear-nonlinear feature projection for multifunction myoelectric hand," presented at the IEEE 9th Int. Conf. Rehabilitation Robotics (ICORR), Chicago, IL, 2005.
- [23] *Myoback Arm Components Catalog*. Duderstadt, Germany: Otto Bock Healthcare, 2004/2005.
- [24] C. W. Heckathorne, "Components for electric-powered systems," in *Atlas of Amputations and Limb Deficiencies*, D. G. Smith, J. W. Michael, and J. H. Bowker, Eds., 3rd ed. Rosemont, IL: American Academy of Orthopaedic Surgeons, 2004, pp. 145–171.
- [25] D. S. Childress, "An approach to powered grasp," in *Advances in External Control of Human Extremities/Fourth International Symposium on External Control of Human Extremities*, M. M. Gavrilovic and A. B. Wilson, Jr., Eds., Belgrade, Yugoslavia, 1973, pp. 159–167.
- [26] R. F. Weir, "Design of artificial arms and hands for prosthetic applications," in *Standard Handbook of Biomedical Engineering and Design*, M. Kutz, Ed. New York: McGraw-Hill, 2003, ch. 32, pp. 32.1–32.61.
- [27] L. Deecke, B. Grozinger, and H. H. Kornhuber, "Voluntary finger movement in man: Cerebral potentials and theory," *Biol. Cybern.*, vol. 23, pp. 99–119, 1976.
- [28] D. E. Goodkin, D. Hertsgaard, and J. Seminary, "Upper extremity function in multiple sclerosis: Improving assessment sensitivity with box-and-block and nine-hole peg tests," *Arch. Phys. Med. Rehabil.*, vol. 69, pp. 850–854, 1988.
- [29] J. Desrosiers, G. Bravo, R. Hebert, E. Dutil, and L. Mercier, "Validation of the Box and Block Test as a measure of dexterity of elderly people: Reliability, validity, and norms studies," *Arch. Phys. Med. Rehabil.*, vol. 75, pp. 751–755, 1994.
- [30] V. Mathiowetz, G. Volland, N. Kashman, and K. Weber, "Adult norms for the Box and Block Test of manual dexterity," *Am. J. Occup. Ther.*, vol. 39, pp. 386–391, 1985.
- [31] G. Goodman and S. Bazyk, "The effects of a short thumb opponens splint on hand function in cerebral palsy: A single-subject study," *Am. J. Occup. Ther.*, vol. 45, pp. 726–731, 1991.
- [32] D. E. Goodkin, R. A. Rudick, S. VanderBrug Medendorp, M. M. Daughtry, K. M. Schwetz, J. Fischer, and C. Van Dyke, "Low-dose (7.5 mg) oral methotrexate reduces the rate of progression in chronic progressive multiple sclerosis," *Ann. Neurol.*, vol. 37, pp. 30–40, 1995.
- [33] J. H. Cauraugh and S. B. Kim, "Chronic stroke motor recovery: Duration of active neuromuscular stimulation," *J. Neurol. Sci.*, vol. 215, pp. 13–19, 2003.
- [34] J. R. Carey, T. J. Kimberley, S. M. Lewis, E. J. Auerbach, L. Dorsey, P. Rundquist, and K. Ugurbil, "Analysis of fMRI and finger tracking training in subjects with chronic stroke," *Brain*, vol. 125, pp. 773–788, 2002.
- [35] J. Cauraugh, K. Light, S. Kim, M. Thigpen, and A. Behrman, "Chronic motor dysfunction after stroke: Recovering wrist and finger extension by electromyography-triggered neuromuscular stimulation," *Stroke*, vol. 31, pp. 1360–1364, 2000.
- [36] E. Park and S. G. Meek, "Adaptive filtering of the electromyographic signal for prosthetic control and force estimation," *IEEE Trans. Biomed. Eng.*, vol. 42, no. 10, pp. 1048–1052, Oct. 1995.
- [37] M. Zardoshti-Kermani, B. C. Wheeler, K. Badie, and R. M. Hashemi, "EMG feature evaluation for movement control of upper extremity prostheses," *IEEE Trans. Rehabil. Eng.*, vol. 3, no. 4, pp. 324–333, Dec. 1995.



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