Available online at www.sciencedirect.com

SCIENCE DIRECT®

Human Movement Science 25 (2006) 165–180

HUMAN MOVEMENT SCIENCE

www.elsevier.com/locate/humov

Speed–accuracy trade-offs in myocontrol

Eric J. Fimbel [a,b,*], Martin Lemay [c], Martin Arguin [b,d]

[a] *Electrical Engineering Department, École de technologie supérieure, 1100 Notre Dame West, Montréal, Qué., Canada H3C 1K3*

[b] *Institut universitaire de gériatrie de Montréal, Canada*

[c] *Department of Psychology, Université du Québec à Montréal, Canada*

[d] *Department of Psychology, Université de Montréal, Canada*

Available online 7 February 2006

Abstract

Myoelectric (EMG) signals are used in assistive technology for prostheses, computer and domestic control. However, little is known about the capacity of controlling these signals. Specifically, it is unclear whether *myocontrol*, i.e., the control of myoelectric signals, obeys the same laws as motor control. Neurologically intact adult participants performed pointing tasks with EMG signals captured from the forehead or the hand in two modalities (sustained: stabilize the signal amplitude in the target; impulsion: produce an impulse and return to resting level). In the sustained modality, the time to reach the target (reach time) increased logarithmically with target amplitude, which is compatible with the predictions of Fitts' law. The rate of failure was not significantly affected by target amplitude. In the impulsion modality, the reach time and the rate of failure followed a bow-shaped pattern as a function of target amplitude. Stabilization time in the sustained modality followed a convex (bow-shaped) pattern for the forehead and a concave pattern for the hand. This was the only significant effect of electrode placement in this study. These findings suggest that myocontrol obeys laws that are distinct from those determining motor control, and that the muscular and intra-muscular synergies that produce EMG signals are specific of each pointing modality and target amplitude.

* Corresponding author. Address: Electrical Engineering Department, École de technologie supérieure, 1100 Notre Dame West, Montréal, Qué., Canada H3C 1K3. Tel.: +1 514 396 8670; fax: +1 514 396 8684.

E-mail address: efimbel@ele.etsmtl.ca (E.J. Fimbel).

0167-9457/$ - see front matter © 2005 Elsevier B.V. All rights reserved.
doi:10.1016/j.humov.2005.12.001

PsycINFO classification: 2330; 4010; 2530

Keywords: EMG; Reaching; Speed–accuracy; Fitts' law; Motor control; Myocontrol

1. Introduction

Myocontrol refers to the control of myoelectric signals such as in electromyography (EMG). Myoelectric signals (EMG) are used in biofeedback (Balliet, Shinn, & Bach-Y-Rita, 1982; Cox & Matyas, 1983; Jacobs & Felton, 1969; Lee, Hill, Johnston, & Smieho-rowski, 1976; Lee, Wong, Tang, Chang, & Chiou, 1996; Webb, 1977; Wieselmann-Penkner, Janda, Lorenzoni, & Polansky, 2001), functional electric stimulation (gait: William & Durfee, 1994; muscle retraining: Thorsen, Spadone, & Ferrarin, 2001) and prosthetics (Sears & Shaperman, 1991; Silcox, Rooks, Vogel, & Fleming, 1993; Zecca, Micera, Carrozza, & Dario, 2002). EMG signals are also used in hand-free computer interfaces for healthy (Trejo et al., 2003) and disabled users (Barreto, Scargle, & Adjouadi, 2000). For instance, a quadriplegic person may type with a virtual keyboard in which the keys are automatically scanned. When the desired key is highlighted, the user generates an EMG impulse (using forehead muscles). The same technique may be used for moving the pointer in a graphical interface, by using a pad with arrow keys instead of letters (Steriadis & Constantinou, 2003).

The EMG signal is often interpreted as a dual-state switch, i.e., *on* and *off*. However, the throughput of the interface can be significantly increased by dividing the range of possible EMG amplitudes in several intervals, each one corresponding to a different command (McKenzie, 1995). To that end, the EMG signal may be controlled visually in real time, by means of a feedback bar, e.g., Cyberlink (Doherty, Bloor, & Cockton, 1999). Command entry may be viewed as a pointing task (called *visuo-EMG pointing task*) in which a "target" is reached by producing an EMG signal of the proper amplitude.

Speed and accuracy are obviously important issues in attempting to apply myo-electric signals in computer interfaces and/or prosthesis. The goal of the present study was to determine whether myocontrol obeys the same laws as motor control. More specifically, we wanted to determine whether visuo-EMG pointing tasks are ruled by speed–accuracy trade-offs such as in regular pointing movements.

In pointing tasks performed under visual control, Fitts' law is one of the most robust laws (Fitts, 1954). It predicts that the difficulty of a task is proportional to $Id = \log_2(2A/W)$, where A is the amplitude of the movement and W the precision demand, i.e., the width of the target. Thus, under Fitts' law, performance indicators like the execution time ET vary linearly with task difficulty, i.e., $ET = a\log_2(2A/W) + b$. The constants a, b are called *information transmission coefficients*. They depend on the particular task and the effector used (e.g., hand, head, foot, etc.). Fitts' law has been verified empirically either in its original form or with slight modifications in a variety of tasks and conditions including 2-D pointing (Bohan, Longstaff, Van Gemmert, Rand, & Stelmach, 2003; Smyrnis, Evdokimidis, Constantinidis, & Kastrinakis, 2000), non-dominant hand pointing (Bryden, 2002), drawing (Mottet & Bootsma, 2001; Mottet, Bootsma, Guiard, & Laurent, 1994) bimanual pointing (Riek, Tresilian, Mon-Williams, Coppard, & Carson, 2003), feet alternating pointing (Hoffmann, 1991), 3-D pointing (Murata & Iwase, 2001; Yang, Zhang, Huang, & Jin, 2002), head pointing (Radwin, Vanderheiden, & Lin, 1990), grasping (Bootsma, Marteniuk, MacKenzie, & Zaal, 1994; Jungling, Bock, & Girgenrath, 2002; Mon-Williams & McIntosh, 2000), real and virtual object rotation (Ruddle & Jones, 2001), isometric tasks (Billon, Bootsma, & Mottet, 2000) and throwing (Etnyre, 1998). Additional examples can be found in Plamondon and Halimi (1997).

However, some doubts arise with respect to the validity of Fitts' law in myocontrol. First, it is possible for participants to use different muscular synergies or synergetic subsets of motor units pertaining to a single muscle (Ter Haar Romeny, Van der Gon, & Gielen, 1984) during myocontrol tasks, and it is unclear whether the information transmission coefficients of these synergies can be subsumed in a single equation. Most relevantly, the use of different muscular synergies in motor control has been proposed as the cause of some violations of Fitts' law (e.g., Danion, Duarte, & Grosjean, 1999; Smits-Engelsman, Van Galen, & Duysens, 2002). Moreover, Fitts' law has been viewed as describing the relationship between the spatial and kinematic aspects of a movement, in which case it may be dependent upon biomechanical factors, e.g., joint stiffness, muscles and tendons elasticity and damping (Mottet & Bootsma, 2001; Viviani & Terzuolo, 1982). Another relevant factor is that, although myo- and motor control share nervous structures and pathways, their outputs are divergent. EMG can be controlled relatively independently from limb position, possibly by using the synchronicity of discharge of motor units (Thorsen et al., 2001). The proprioceptive signals that are relevant for controlling the output are therefore different in myo- and motor control. This means that proprioceptive feedback may be interpreted and/or used differently in these two types of control.

In order to determine whether visuo-EMG pointing tasks are ruled by speed–accuracy trade-offs, participants in the present investigation performed visuo-EMG pointing tasks in two different modalities. In the *impulsion modality,* the movement endpoint corresponded to the maximal amplitude attained by the feedback bar during an EMG impulse, without any stabilization on the target required. This is equivalent to a motor pointing task in which the limb does not stop in the target (see Fitts, 1954; Woodworth, 1899). It is worth emphasizing that EMG impulses are *not* similar to ballistic movements. The pattern of myoelectric amplitude during EMG impulses differs in both shape and duration from the triphasic EMG bursts that generate ballistic movements (e.g., Hallett, Shahani, & Young, 1974). In the *sustained modality*, the amplitude of the feedback bar had to be stabilized on the specified target for 1 s. This is equivalent to a motor pointing task in which the limb stops on the target. Whereas in regular pointing movements, Fitts' law has been verified whether the effector stops on the target (e.g., Fitts & Peterson, 1968) or not (e.g., Fitts, 1954), but the situation may be different in myocontrol because stabilizing the EMG amplitude is more difficult than stabilizing the position of a limb or a mechanical effector (Light, Chappell, Hudgins, & Engelbart, 2002).

In order to assess the effect of the electrode site, two groups of participants performed the task, one with an electrode placement on the forehead and the other on the thumb. Forehead is typically used for quadriplegic patients, and thumb is used in precision movements. It may thus provide the finest possible control for EMG signals.

2. Material and methods

2.1. Participants and apparatus

Two groups of young healthy volunteers with no history of motor, neurological or perceptual deficits performed identical visuo-EMG pointing tasks, each one with a different electrode site. Participants gave informed consent according to the ethics regulation of the Institut universitaire de gériatrie de Montréal.

For the *hand group* ($n = 19$, age = 24.3 years, $\sigma = 3.9$, 13 males, 2 left-handed), the electrode montage was placed on the palm of the dominant hand, on the thenar eminence. It captured EMG activity from the following muscles: abductor pollicis brevis, flexor pollicis brevis, opponens pollicis, adductor pollicis transversalis (and marginally, from other hand muscles, such the first and second lumbricali and opponens quinti digiti).

For the *forehead group* ($n = 19$, age = 24.5, $\sigma = 3.8$, 13 males, 3 left-handed), the electrode montage was placed above the right eyebrow, approximately 3 cm from the median line. It captured EMG activity from the frontalis, orbito-ocularis, fruncidor, levator parpebrae, corrugator, masseter (and marginally from the zygomaticus, quadratus labii superioris and left auricularis).

The signal was captured by means of a montage composed of three dry silver electrodes (two differential and one common electrode, spaced 2 cm apart from each other) and a preamplifier (Neurodyne AE-104). The signal was filtered in the band 25–450 Hz, amplified and converted in RMS (root mean square) by an amplifier (Neurodyne System/3). The RMS signal was sampled at 1 KHz by a computer, smoothed to reduce variability for accurate control and displayed on a monitor as a feedback bar (Fig. 1a; see legend for details). The refresh rate of the display was 20 Hz and the total temporal lag between raw EMG signal and display was 55 ms, $\sigma = 8$ ms (see Section 4 for the potential effects of the lag).

2.2. Calibration

After a training phase (see next section for details), the gain of the feedback was calibrated in order to display only a range of amplitude $[L, H]$. H was initialized to the maximum amplitude that could be attained, and decreased until it could be reached in three consecutive attempts. The objective was that H was as high as possible, but could be attained at any time during the task. L was then determined as the average rest amplitude plus one standard deviation. After calibration, the feedback bar represented the relative amplitude in the range 0–100% instead of the voltage in the range 0–750 mV. It is noteworthy that the participant was still capable of producing levels above H, which was not an absolute maximum, and below L, which was one standard deviation above the minimum (see Fig. 1b).

The pointing task was as follows. The range $[L, H]$ was divided in eight intervals of identical size with a width of 12.5% of the total range $[L, H]$. Each interval represented a possible target. The "amplitude" A was determined by the position of the interval center and the "diameter" W by the width (see Fig. 1c). By changing the number of targets, it is possible to obtain different combinations of A/W ratios. Specifically, with eight targets, the width of each target was 12.5% of the range $[L, H]$ and the distances *from* L were $6.25\%, 18.75\%, \ldots, 93.75\%$. In order to normalize Id, we considered that the minimal level L was at distance $W/2$ from the rest level (see Fig. 1c for details). By doing so, the ratios $2A/W$ for the eight targets become $2, 4, 6, \ldots, 16$, and Id is in the range $[1, 4]$. This normalization is licit here, because we only look for linear correlations between Id and the performance indicators.

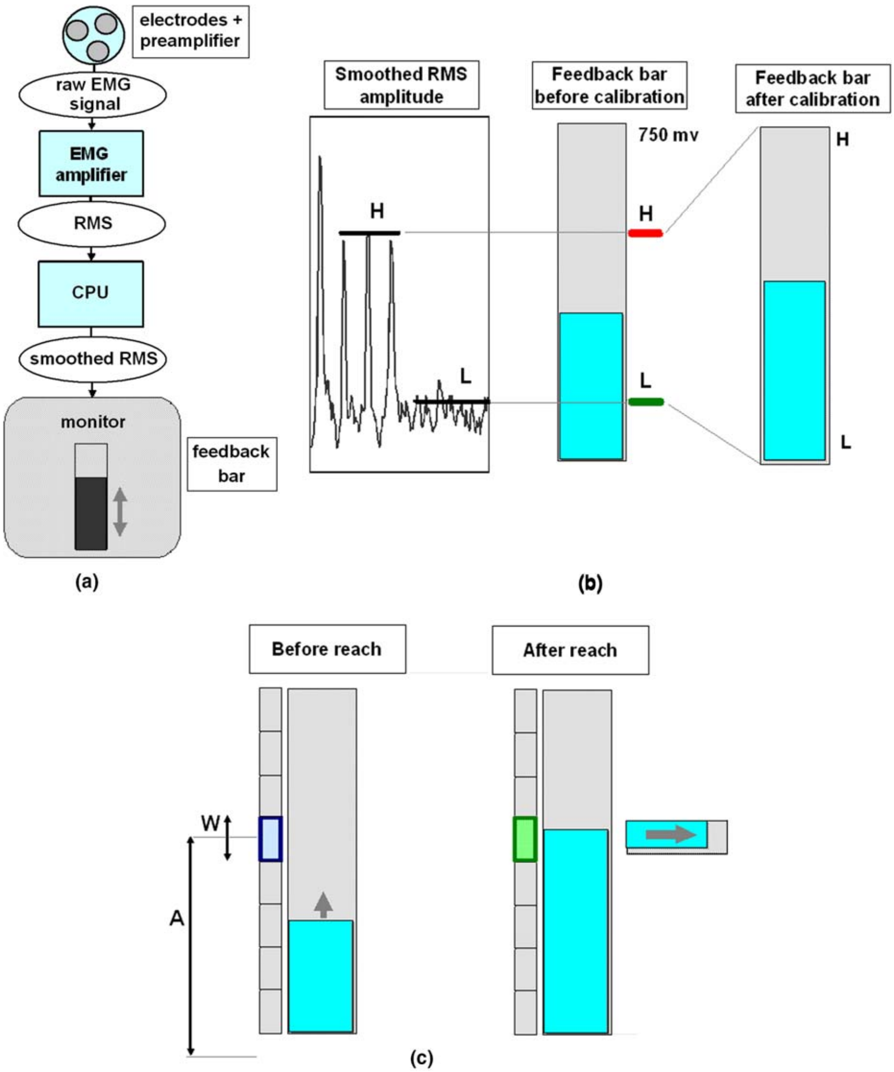

Fig. 1. Experimental setup. (a) Signal capture. Raw EMG signal filtered in the band 25–450 Hz, converted in RMS, sampled at 1 KHz and smoothed by averaging on bins of 50 ms. Feedback bar: width 2 cm, maximal height 20 cm, distance from the eyes 60 ± 10 cm, refresh rate 20 Hz. (b) Feedback calibration. Before calibration, the bar represents the voltage of the smoothed RMS signal in the range 0–750 ms. After calibration, the bar represents the signal normalized in the range L, H. (c) Pointing task. Feedback bar represented before- and after the reach in the sustained modality. A, W: amplitude and width of the target. For normalization, we considered that L was at distance $W/2$ from the rest level, i.e., for target k, $A = W/2 + (k - 1)W + W/2 = kW$. Horizontal progress bar: time elapsed in the target.

2.3. Procedure

In the initial training phase, the participant was instructed to "find a way to control the feedback bar by contracting his or her muscles". During this phase, the feedback displayed the voltage of the smoothed RMS signal. The training phase terminated when the participant and the experimenter agreed that minimal control was obtained (average duration 110 s, $\sigma = 65$ s).

Each trial proceeded as follows. The participant maintained the amplitude below level L for at least 500 ms. One second later a target was displayed, in the form of a blue rectangle on the left of the feedback bar. The participant was instructed to reach the target as quickly as possible. At the end of the trial, feedback was provided. The rectangle became green in the case of a successful trial, and orange in the case of a failed trial (see Fig. 1c).

The pointing task was performed in two modalities. In the *impulsion modality*, the target was reached by means of a single impulse. The trial was considered successful when the peak amplitude was within the target interval. In the *sustained modality*, the amplitude was stabilized in the target for 1 s. A horizontal progress bar indicated the time elapsed when on the target (see Fig. 1c). The bar reached the right extremity after 1 s, indicating a successful trial. When the amplitude went out of the target, the progress bar was reinitialized. If the EMG amplitude could not be stabilized in the target area for one uninterrupted second within 10 s of the start of the trial (i.e., after the presentation of the target), the trial was considered a failure.

Participants executed the pointing task in four conditions, in the following order: sustained low precision (4 targets), sustained high precision (8 targets), impulsion low precision, and impulsion high precision. Only the high precision condition is analyzed here.[1]

For each condition, participants performed four blocks of 32 trials. The first block was considered as training, therefore only the blocks 2, 3, and 4 were analyzed. In each block, the targets appeared at random positions with a uniform probability. The amplitude was recalibrated at the beginning of each block, in order to avoid the drifts in minimal and maximal amplitudes due to fatigue and/or changes in the skin conductivity. Recalibration was equally possible within a block (e.g., when the electrode montage was accidentally displaced) but this possibility was only used marginally (20 times in 640 blocks). A 2-min pause was taken between each block in order to rest the muscles. The total duration of the test was between 1,5 and 2 h.

2.4. Data capture and analysis

In the impulsion modality, the trial was successful when the peak amplitude was within the target interval. Note that the term "amplitude" refers to the signal that was displayed, i.e., the smoothed signal sampled at 20 Hz. Local maxima were identified as the samples preceding an amplitude decrease of at least 2 samples (i.e., 100 ms). This simple method was adequate because the smoothing eliminated all the spikes that are present in the raw EMG signal. Preliminary tests showed that in the impulsion modality, corrective commands take the form of a decrease in the rate of raise of the amplitude, but not of a decrease of the amplitude itself (because of the smoothing). Therefore, the *reach time* (R) was determined as the time between the presentation of the target and the first peak. The

[1] The low precision condition was introduced in order to compare the effect of precision demands on performance in young vs. elderly participants (data not presented here). Using both precision conditions would have caused unequal numbers of trials per value of *Id*, because some values of *Id* are only possible in the high precision condition.

dependent variables analyzed were the rate of failure (F) and the reach time, considered as the execution time for the trial.

In the sustained modality, the trial was a failure if the time ran out (amplitude not stabilized within 10 s) and successful otherwise. The *reach time* (R) was determined as the time between the presentation of the target and the moment when the amplitude entered within the target. However, in this modality, the total completion time including stabilization may lead to markedly different results (see Hourcade, Bederson, Druin, & Guimbretière, 2004). We therefore used as an additional dependent variable the *complete stabilization time* (S), i.e., the time between the presentation of the target and the end of the 1-s stabilization period. The difference between the complete stabilization time and the reach time is an estimate of the duration of the stabilization phase. The variables analyzed therefore, were the rate of failure (F), the reach time and the complete stabilization time (in successful trials only, since S is not significant in failed trials: it is the arbitrary 10 s time out).

Two types of analyses were conducted. First, the validity of Fitts' law was examined. Correlations were calculated between Fitts' index of difficulty ($Id = \log_2(2A/W)$, A: amplitude of the center of the target, W: width of the target) and the dependent variables averaged across participants and trials (impulsion modality: F, R; sustained modality: F, R, S). Correlations with a level of confidence of $p < .05$ (unilateral) were considered significant. The correlations were calculated separately for each Electrode site × Modality combination, because the information transmission coefficients may differ according to the effector (forehead vs. hand muscles) and the task (reach the target vs. stop in the target). Second, the joint effects of electrode site and target amplitude were examined. Thus, each dependent variable was analyzed separately (using ANOVAs) according to the factors of electrode site (hand vs. forehead) and amplitude (eight levels).

3. Results

3.1. Fitts' index of difficulty vs. performance indicators

The correlation coefficients between Fitts' index of difficulty and the performance indicators (impulsion modality: reach time R, rate of failure F; sustained modality: reach time R, complete stabilization time S, rate of failure F) are presented in Table 1. The only correlations that attained significance are those for reach time R in the sustained modality (forehead: $r = .95$; thumb: $r = .69$). These results are consistent with the data depicted in Figs. 2 and 3, which show the performance indicators as a function of Id in the two pointing modalities.

Table 1
Correlations of performance measures (Reach time – R, complete stabilization time – S, and failure rate – F) with Fitts' index of difficulty Id, as a function of response modality and electrode site

	Impulsion modality		Sustained modality	
	Hand	Forehead	Hand	Forehead
R	−.54	−.30	.69*	.95*
S	–	–	−.32	−.12
F	−.05	−.07	.05	.36

* $p < .05$, unilateral test (df = 6).

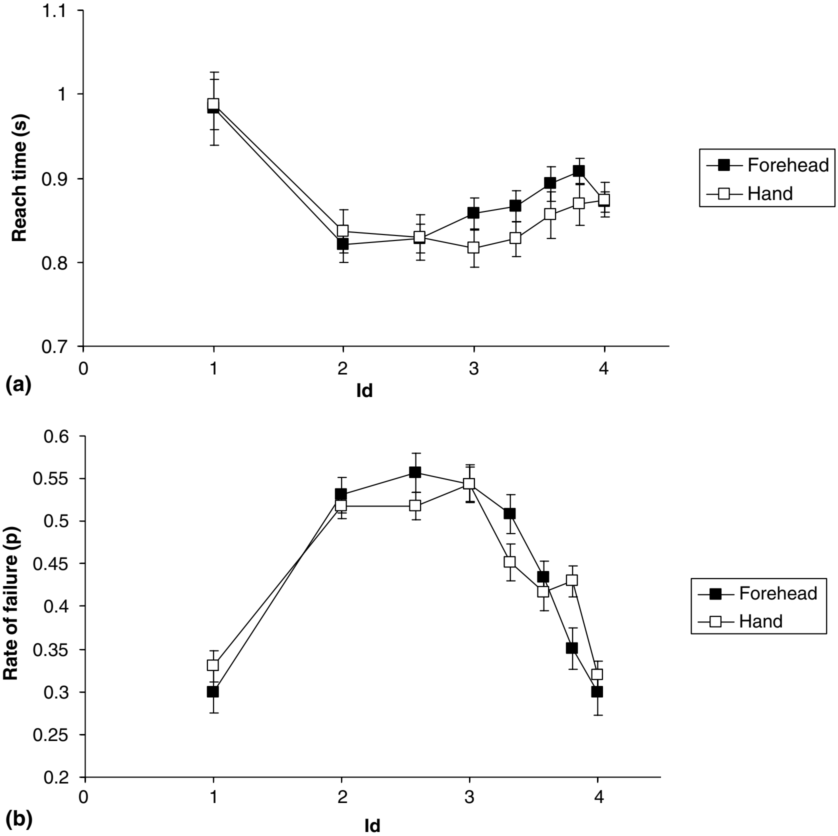

Fig. 2. Mean performance indicators and standard errors of the means vs. Fitts' index of difficulty in the impulsion modality. (a) Reach time *R*. (b) Rate of failure *F*. Variables averaged across trials and participants. The electrode sites (forehead and hand) are represented separately (see legend on figures).

In the impulsion modality, it can be observed that the reach time *R* followed similar patterns for the two electrode sites. The longest reach time was obtained with lowest amplitude ($Id = 1$). Reach time then dropped to about its lowest level at the next amplitude ($Id = 2$) and then rose slightly with increasing amplitudes afterwards. It is noteworthy that for all amplitudes the reach time is long enough to allow for visually-guided adjustments of amplitude (forehead: average 869 ms; thumb: average 863 ms). The rate of failure is high (forehead and thumb: average 45%) and follows a convex pattern as a function of *Id*, i.e., performance is better for extreme amplitudes than for intermediate ones. It is worth noting that this is not due to a boundary effect. As seen before, overshoots for the maximal amplitude, and undershoots for the minimal amplitude, are equally possible.

In the sustained modality, reach time *R* increased with the index of difficulty for both electrode sites. Although the average reach time was comparable to that of the impulsion modality (forehead and thumb: average = 932 ms), the different patterns as a function of *Id* may reflect the difference between modalities in the final state (not stabilized vs. stabilized). The complete stabilization time *S* (forehead and thumb: average = 2687 ms) followed

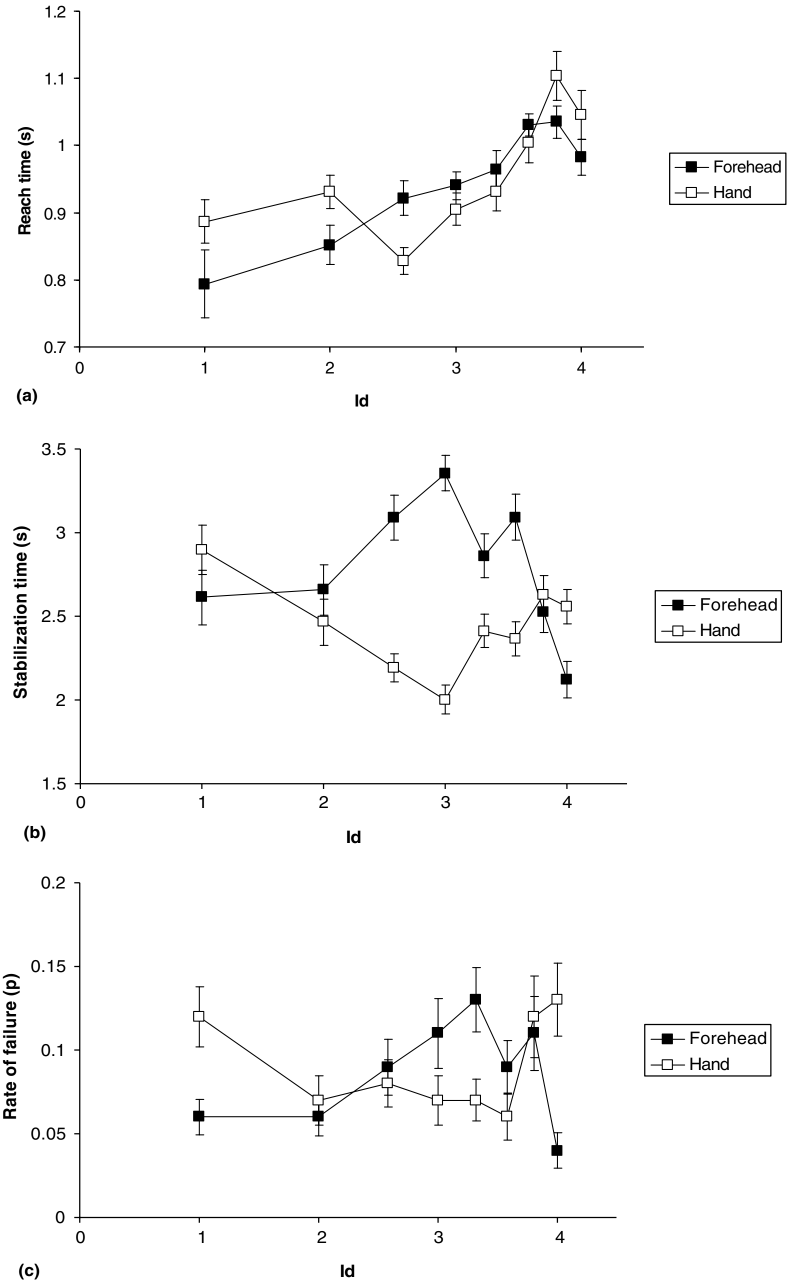

Fig. 3. Mean performance indicators and standard errors of the means vs. Fitts' index of difficulty in the sustained modality. (a) Reach time *R*. (b) Complete stabilization time *S*. (c) Rate of failure *F*. These variables are averaged across participants and trials. The electrode sites (forehead and hand) are represented separately (see legend on figures).

different patterns according to electrode site. With the forehead, it is convex, i.e., performance is better for extreme amplitudes than for intermediate ones. Conversely, with the thumb, it is concave, i.e., better performance is obtained for the middle amplitudes than for those at the extremes. The rate of failure F follows the same patterns as the complete stabilization time S, i.e., convex for the forehead and concave for the thumb. As expected, the rate of failure F (forehead and thumb: average 8%) is significantly lower in the sustained than the impulsion modality.

3.2. Effects of electrode site and amplitude in the impulsion modality

For R, the main effect of electrode site, $F(1,36) < 1$, and the Electrode × Amplitude interaction, $F(7,252) < 1$, were both non-significant. However, a significant main effect of amplitude was found, $F(7,252) = 6.5$, $p < .001$. This amplitude effect was non-monotonic and corresponds to a decrease of R between Id's of 1 and 2, $t(37) = 4.6$; $p < .001$, followed by a slight rise with increasing Id (see Fig. 2a). The largest pairwise difference is between levels 2 and 7, which is significant, $t(37) = 2.1$; $p < .05$.

The analysis of F revealed a significant main effect of amplitude, $F(7,252) = 14.1$, $p < .001$, but no significant main effect of electrode site, $F(1,36) < 1$, and no significant interaction, $F(7,252) < 1$. The main effect of amplitude was non-monotonic and characterized by a convex, bow shape (see Fig. 2b). This non-monotonicity was demonstrated by the fact that F increased significantly between amplitudes 1 and 2, $t(37) = 6.3$; $p < .001$, whereas it decreased significantly between amplitudes 7 and 8, $t(37) = 2.3$; $p < .05$.

In summary, electrode site had no significant effect on reach time or rate of failure. Also, there was no significant Electrode site × Amplitude interaction on any of the dependent variables. Amplitude had a significant effect on both reach time and rate of failure, but these effects were non-monotonic.

3.3. Effects of electrode site and amplitude in the sustained modality

The analysis of R showed no significant main effect of electrode site, $F(1,36) < 1$, and no significant Electrode site × Amplitude interaction, $F(7,252) = 1.5$, n.s., but the main effect of amplitude was highly significant, $F(7,252) = 7.9$, $p < .001$. This amplitude effect corresponds to a regular increase of R with Id (and amplitude) which is depicted in Fig. 3a.

For S, the main effects of electrode site, $F(1,36) = 1.7$, n.s., and of amplitude, $F(7,252) = 1.3$; n.s., were both non-significant. However, the Electrode × Amplitude interaction was highly significant, $F(7,252) = 7.2$, $p < .001$. Simple effects of this interaction revealed significant effects of amplitude for both the hand, $F(7,126) = 3.1$, $p < .005$, and forehead, $F(7,126) = 5.1$, $p < .001$, electrode placements. These amplitude effects were non-monotonic, bow-shaped, and in opposite directions for the hand and forehead (see Fig. 3b). The non-monotonicity of the amplitude effect for the hand was demonstrated by a marked reduction of S between amplitudes 1 and 4, $t(18) = 3.5$, $p < .005$, together with a significant increase of S from amplitude 4 to 8, $t(18) = 2.7$; $p < .05$. For the forehead, the pattern was reversed, with a significant increase of S between amplitudes 1 and 4, $t(18) = 3.2$, $p < .01$, followed by a significant decrease between amplitudes 4 and 8, $t(18) = 5.6$, $p < .001$.

For F, the main effects of electrode site, $F(1,36) < 1$, and of amplitude, $F(7,252) < 1$, as well as the Electrode site × Amplitude interaction, $F(7,252) = 1.8$, failed to reach significance.

In summary, electrode site had no significant effect on reach time and rate of failure. Amplitude had an effect on the reach time, which increased regularly with increasing Id

(see previous section) amplitude affected complete stabilization time S in a non-monotonic, bow-shaped manner that was in opposite directions for the forehead and thumb electrodes. Finally, no significant effect was observed on the rate of failure.

4. Discussion

The goal of the present research was to determine whether myocontrol abides by the same laws as motor control. More specifically, we wanted to determine whether visuo-EMG pointing tasks are ruled by speed–accuracy trade-offs in two pointing modalities, the impulsion and sustained modalities.

The present results demonstrate different speed–accuracy trade-offs across the two pointing modalities. Thus, the impulsion modality is a high-speed and low-accuracy task (average reach time across electrode sites = 866 ms, average rate of failure = 45%), whereas the sustained modality is a low-speed and high-accuracy task (average execution time = 2687 ms, average rate of failure = 8%). In relation to the laws that govern myocontrol, the anticipation of the final state of the effector (steady EMG amplitude in the sustained modality, not stabilized EMG amplitude in the impulsion modality) seemed to affect the reach phase. Thus, while reach times are relatively similar in both modalities (sustained: 932 ms, impulsion: 866 ms), they nevertheless seem to obey different laws (see Figs. 2 and 3). Whereas in the impulsion modality, the reach time followed a non-monotonic pattern as a function of amplitude, in the sustained modality, the reach time increased markedly with amplitude. This suggests that the necessity of stabilizing the EMG signal elicited particular control and execution strategies for reaching the target that differ from those in the impulsion modality.

There is also evidence of speed–accuracy trade-offs *within* each modality. Indeed, in the sustained modality, the relation between reach time and amplitude is well described by Fitts' law, which characterizes a trade-off between precision demand and execution time. In the impulsion modality, rate of failure and reach time, as functions of amplitude, follow opposite patterns (Fig. 2), convex for the rate of failure and concave for the reach time. In other terms, low rates of failure correspond to slow reach times (extreme amplitudes) and vice versa (middle amplitudes). This suggests a trade-off between execution time and resulting accuracy rather than with precision demand.[2]

4.1. Fitts' law is not verified in the impulsion modality

The lack of support for Fitts' law in the impulsion modality was unexpected, because this is clearly a pointing task, similar to reaching a target without end-point stabilization (like alternate lines drawing, or alternate peg-in-the-hole tasks). Both the instructions and the feedback provided after each trial emphasized that the objective was to attain the target. Therefore it is reasonable to assume that participants tried to work at maximal accuracy *and* maximal speed, as is required for the application of Fitts' law.

[2] Both types of accuracy, precision demand and observed accuracy could have been represented by means of a unique indicator, namely the *effective Id* calculated from the dispersion of end points (Schmidt, Zelaznik, Hawkins, Frank, & Quinn, 1979). However, the difference in accuracy between the impulsion and sustained modalities would have complicated the comparison of the results.

The unexpected concave pattern presented by reach time as a function of *Id* was mirrored by the convex pattern of the rate of failure. In other terms, extreme amplitudes were more slowly attained than intermediate ones, but they were also attained more accurately. Whereas for high amplitudes this may be compatible with a decrease in variability for movements of maximal amplitude (e.g., Sherwood & Schmidt, 1980), there is no such explanation for the minimal amplitude. A bias in the experimental protocol is unlikely, because overshoots and undershoots are just as possible for minimal and maximal amplitude targets as they are for intermediate ones (see Fig. 1b and c). The convex pattern of the rate of failure cannot be explained by the visual aspect of the task either. Absolute judgment of sizes in central visual field presents almost no distortion, i.e., the psychophysical mapping (the correspondence between physical size and subjective magnitude) is almost linear (e.g., Spence, 1990).

The most likely explanation is that the high rate of failure is caused by a poor precision in the production of EMG amplitude, and that the relatively higher accuracy obtained for extreme amplitudes results from the application of specific muscular synergies (and patterns of activation of the recruited motor units) for the smallest and the largest levels of amplitude. Each muscular synergy may correspond to different information transmission coefficients, which may contribute to the present exception to Fitts' law. Such exceptions have been observed in motor control when different effectors are used for the same task (Danion et al., 1999; Smits-Engelsman et al., 2002).

Indeed, the high inaccuracy complicates the interpretation of the results in terms of information transmission capacity (although the case of inaccurate transmission was considered by Shannon in his seminal paper (Shannon, 1948, p. 20)). However, high inaccuracy clearly indicates that beyond some point, participants were unable to trade speed for additional accuracy, i.e., slower movements or corrections could not increase accuracy. In any case, the present findings are by no means a "violation" to Fitts' law. They only indicate that some conditions of application of Fitts' law are not verified for pointing tasks with EMG amplitude in the impulsion modality, namely broad speed–accuracy trade-offs and the use of the same effectors for all the targets.

4.2. Fitts' law is verified for the reach phase in the sustained modality

In the sustained modality, the reach time (*R*) showed a strong correlation with Fitts' index of difficulty *Id* ($r = .95$). According to the interpretation standards that are generally applied in the literature, it should be concluded that Fitts' law is verified in the sustained modality. In this respect however, it should be emphasized that high coefficients of linear correlation such as observed here fail to clearly discriminate between Fitts' law and other monotonically increasing functions, like the linear law proposed law proposed by Schmidt et al. (1979), or the power law proposed by Gan and Hoffmann (1988).

When the stabilization phase was considered, the execution time represented by the dependent variable *S* as a function of *Id* presented non-monotonic patterns for both electrode sites, which is incompatible with the predictions of Fitts' law (see Fig. 3b). Such a dissociation between the reach time and the total completion time has been observed before in pointing tasks performed by young children, presumably because they had difficulty in executing the final phase of the movement (Hourcade et al., 2004; see also Meyer, Schmidt, Kornblum, Abrams, & Wright, 1990). In the present case, the non-monotonic patterns originated in the stabilization phase, because they were absent from the reach time, which increased almost linearly with *Id* (see Fig. 3a). The convex pattern found for the forehead electrode site suggests that it was more difficult to stabilize the signal at middle than at extreme amplitudes. Conversely, the concave pattern found for the hand electrode site suggests that stabilization at middle amplitudes was easier.

These findings support the idea that the difficulty of stabilizing the EMG signal as a function of amplitude depends on the muscular synergies. It has been proposed that for maintaining a steady amplitude, the recruitment of motor units changes in time (Light et al., 2002). This dynamic recruitment seems to be demanding regardless of the muscles used.

4.3. Common mechanisms and limiting factors in myocontrol

Apart from the stabilization time in the sustained modality, the present results did not show any significant effect of electrode site. This is all the more significant in that the forehead and the hand are controlled through distinctive neural pathways, i.e., trigeminal nerve vs. pyramidal pathway. This general absence of an electrode site effect suggests common mechanisms and/or limiting factors for myocontrol. Controlling myoelectric amplitude is not a familiar task. In normal circumstances, myoelectric signals are used to control movements, but they are not controlled directly. From an evolutionary viewpoint, the motor system is presumably adapted for the control of effectors like limbs, eyes, or head. The corresponding variables (dynamic, kinematic and geometric) present low frequencies as well as biomechanical constraints. Conversely, myoelectric signals present relatively high frequency components (e.g., 500 Hz) and may mostly be constrained by spinal sensorimotor circuits.

Also, the control of myoelectric amplitude is quite specific. Whereas proprioceptive signals convey information on the geometry and dynamics of muscles and joints, they have only indirect relationships with the overall EMG amplitude. This amplitude results from the spatial location, firing rate and synchronicity of motor units' discharge (Thorsen et al., 2001). The recruitment of these units is partially under voluntary control. By controlling the contraction (e.g., isometric, isotonic) and position of synergetic muscles, motor units from different muscles are recruited. Also, it is possible to recruit sets of motor units pertaining to the same muscle whose activity depends upon specific combinations of forces and movements, e.g., pronation/supination vs. endo-/exo-rotation (Ter Haar Romeny et al., 1984). In terms of information, the limited control on the sources of the myoelectric amplitude (recruitment, firing rate and synchronicity) may limit the information transmission capacity of the motor system in EMG pointing tasks.

Regardless of the application in which EMG amplitude is used as a control signal, e.g., myo-prostheses, computer or domestic interfaces, the myoelectric signal has to be filtered and smoothed in order to eliminate high frequency components that are almost uncontrollable. Although generally beneficial, a potentially negative impact of smoothing is that it introduces a temporal lag between the raw myoelectric activity and the response of the processed signal. Indeed, it may be suggested that complementary studies will be required in order to assess the combined effect of slower variation and lag on performance. However, it has been shown that Fitts' law remains valid with lags up to 260 ms (with head-mounted pointers; So, Chung, & Goonetilleke, 1999). Thus in the present case, it is reasonable to assume that the lag (55 ms) did not affect the relationship between performance and precision demand.

5. Conclusions

The present findings highlighted important differences between myo- and motor control. In myocontrol, performance in pointing tasks is relatively independent from the electrode placement, whereas in motor control it may depend markedly on the effector (e.g., foot vs. hand, Hoffmann, 1991). Also the speed–accuracy trade-offs encountered in the present study were different from those observed in motor control. In the impulsion pointing modality, different muscular synergies appear to be chosen *a priori* in order to attain the desired amplitude. After that, the target amplitude is reached in a constant time. Conversely, in the sustained pointing modality, the reach time depends on the amplitude in a way that suggests the presence of corrective commands. The stabilization of EMG amplitude is also a specific feature of myocontrol. Whereas it is relatively easy to stabilize a body part at the end-point of a movement, EMG amplitude is difficult to stabilize, as indicated by the large stabilization times found in the present study. Furthermore, the difficulty depends markedly on the combination of amplitude and electrode placement. Here, it was easier to stabilize extreme amplitudes with the electrode placed on the hand, and middle amplitudes with the electrode placed on the forehead.

Different factors may contribute to the specificity of myocontrol. From an evolutionary viewpoint, the motor system is not adapted to control the specific sources of the EMG amplitude. The relationship between proprioceptive information and myoelectric output is not integrated in the motor system, either in the form of spinal reflexes or automatisms, i.e., highly practiced movements that can be executed with minimal attention demands (Shiffrin & Schneider, 1977). Also, it seems that there are many degrees of freedom for acting upon the EMG amplitude such as intra- and inter- muscular synergies of motor units, patterns of discharge and synchronicity, and that the participants take advantage of this possibility.

The present findings are of practical importance for myoelectric interfaces. Because of the difficulty of stabilizing EMG amplitude, the entry of commands should use impulsions or very short stabilization periods. The relationships between accuracy and target amplitude may be useful for adequately choosing the number of different commands and the corresponding intervals of amplitude. Indeed, the present study focused on performance indicators, execution time and accuracy. In order to gain insights on the laws that govern the control of EMG amplitude, it would be of interest to record geometric and kinematic variables, i.e., the EMG amplitude and its variation as a function of time during pointing. Also it would be useful to conduct studies for assessing the effects of signal processing parameters on performance. Finally it is worth recalling that myocontrol has to be entirely learned. Whereas the present findings are of practical importance during the learning phase, complementary studies will be required in order to assess the effect of training on myocontrol. These studies may bring valuable insights on the potential of myoelectric systems in assistive technologies for disabled users of myo-prostheses and computer or domestic interfaces.

Acknowledgements

This work was supported by CIHR grant 109291 and VRQ grant 2200-099. We thank Daniel Ramirez Hoyo for his valuable contribution, and the anonymous reviewers for their useful comments.

References

Balliet, R., Shinn, J. B., & Bach-Y-Rita, P. (1982). Facial paralysis rehabilitation. *International Rehabilitation Medicine, 4*, 67–74.

Barreto, A. B., Scargle, S. D., & Adjouadi, M. (2000). A practical EMG-based human–computer interface for users with motor disabilities. *Journal of Rehabilitation Research and Development, 37*, 1–14.

Billon, M., Bootsma, R. J., & Mottet, D. (2000). The dynamics of human isometric pointing movements under varying accuracy requirements. *Neuroscience Letter, 286*, 49–52.

Bohan, M., Longstaff, M. G., Van Gemmert, A. W. A., Rand, M. K., & Stelmach, G. E. (2003). Differential effects of target height and width on 2D pointing movement duration and kinematics. *Motor Control, 7*, 278–289.

Bootsma, R. J., Marteniuk, R. G., MacKenzie, C. L., & Zaal, F. T. J. M. (1994). The speed–accuracy trade-off in manual prehension: Effects of movement amplitude, object size and object width on kinematic characteristics. *Experimental Brain Research, 98*, 535–541.

Bryden, P. J. (2002). Pushing the limits of task difficulty for the right and left hands in manual aiming. *Brain and Cognition, 48*, 287–291.

Cox, R. J., & Matyas, T. A. (1983). Myoelectric and force feedback in the facilitation of isometric strength training: A controlled comparison. *Psychophysiology, 20*, 35–43.

Danion, F., Duarte, M., & Grosjean, M. (1999). Fitts' law in human standing: The effect of scaling. *Neuroscience Letter, 277*, 131–133.

Doherty, E., Bloor, C., & Cockton, G. (1999). The "Cyberlink" brain–body interface as an assistive technology for persons with traumatic crain injury: Longitudinal results from a group of case studies. *Cyberpsychology and Behavior, 2*, 249–259.

Etnyre, B. R. (1998). Accuracy characteristics of throwing as a result of maximum force effort. *Perceptual and Motor Skills, 86*, 1211–1217.

Fitts, P. M. (1954). The information capacity of the human motor system in controlling the amplitude of movement. *Journal of Experimental Psychology, 47*, 381–390.

Fitts, P. M., & Peterson, J. R. (1968). Information capacity of discrete motor responses. *Journal of Experimental Psychology, 67*, 103–112.

Gan, K. C., & Hoffmann, E. R. (1988). Geometrical conditions for ballistic and visually controlled movements. *Ergonomics, 31*, 829–839.

Hallett, M., Shahani, B. T., & Young, R. R. (1974). EMG analysis of stereotyped voluntary movements in man. *Journal of Neurology, Neurosurgery and Psychiatry, 38*, 1154–1162.

Hoffmann, E. R. (1991). A comparison of hand and foot movement times. *Ergonomics, 34*, 397–406.

Hourcade, J.-P., Bederson, B. B., Druin, A., & Guimbretière, F. (2004). Differences in pointing task between preschool children and adults using mice. *ACM Transactions on Computer–Human Interaction, 11*, 357–386.

Jacobs, A., & Felton, G. S. (1969). Visual feedback of myoelectric output to facilitate muscle relaxation in normal persons and patients with neck injuries. *Archives of Physical Medicine and Rehabilitation, 50*, 34–39.

Jungling, S., Bock, O., & Girgenrath, M. (2002). Speed–accuracy trade-off of grasping movements during microgravity. *Aviation, Space and Environmental Medicine, 73*, 430–435.

Lee, K. H., Hill, E., Johnston, R., & Smiehorowski, T. (1976). Myofeedback for muscle retraining in hemiplegic patients. *Archives of Physical Medicine and Rehabilitation, 57*, 588–591.

Lee, M. Y., Wong, M. K., Tang, F. T., Chang, W. H., & Chiou, W. K. (1996). Design and assessment of an adaptive intermittent cervical traction modality with EMG biofeedback. *Journal of Biomechanical Engineering, 118*, 597–600.

Light, C. M., Chappell, P. H., Hudgins, B., & Engelbart, K. (2002). Intelligent multifunction myoelectric control of hand prostheses. *Journal of Medicine Engineering & Technology, 26*, 139–146.

McKenzie, I. S. (1995). Input devices and interaction techniques for advanced computing. In *Virtual environments and advanced interfaces design* (pp. 437–470). Oxford: Oxford University Press.

Meyer, D. E., Schmidt, J. E. K., Kornblum, S., Abrams, R. A., & Wright, C. E. (1990). Speed–accuracy trade-offs in aimed movements: Toward a theory of rapid voluntary action. In M. Jeannerod (Ed.), *Attention and performance XIII* (pp. 173–226). Hillsdale, NJ: Lawrence Erlbaum.

Mon-Williams, M., & McIntosh, R. D. (2000). A test between two hypotheses and a possible third way for the control of prehension. *Experimental Brain Research, 134*, 268–273.

Mottet, D., & Bootsma, R. J. (2001). The dynamics of rhythmical aiming in 2D task space: Relation between geometry and kinematics under examination. *Human Movement Science, 20*, 213–241.

Mottet, D., Bootsma, R. J., Guiard, Y., & Laurent, M. (1994). Fitts' law in two-dimensional task space. *Experimental Brain Research, 100*, 144–148.

Murata, A., & Iwase, H. (2001). Extending Fitts' law to a three-dimensional pointing task. *Human Movement Science, 20*, 791–805.

Plamondon, R, & Halimi, A. M. (1997). Speed/accuracy trade-offs in target-directed movements. *Behavioral and Brain Science, 20*, 279–349.

Radwin, R. G., Vanderheiden, G. C., & Lin, M. L. (1990). A method for evaluating head-controlled computer input devices using Fitts' law. *Human Factors, 32*, 423–438.

Riek, S., Tresilian, J. R., Mon-Williams, M., Coppard, V. L., & Carson, R. G. (2003). Bimanual aiming and overt attention: One law for two hands. *Experimental Brain Research, 153*, 59–75.

Ruddle, R. A., & Jones, D. M. (2001). Manual and virtual rotation of a three-dimensional object. *Journal of Experimental Psychology: Applied, 7*, 286–296.

Schmidt, R. A., Zelaznik, H., Hawkins, B., Frank, J. S., & Quinn, J. T. (1979). Motor-output variability: A theory for the accuracy of rapid motor acts. *Psychological Review, 86*, 415–451.

Sears, H. H., & Shaperman, J. (1991). Proportional myolectric hand control: An evaluation. *American Journal of Physical Medicine and Rehabilitation, 70*, 20–28.

Shannon, C. E. (1948). A mathematical theory of communication. *Bell System Technical Journal, 27* 379–423 and 623–656.

Sherwood, D. E., & Schmidt, R. A. (1980). The relationship between force and force variability in minimal and near-maximal static and dynamic contractions. *Journal of Motor Behavior, 12*, 75–89.

Shiffrin, R. M., & Schneider, W. (1977). Controlled and automatic information processing: II. Perception, learning, automatic attending and a general theory. *Psychological Review, 84*, 127–190.

Silcox, D. H., 3rd, Rooks, M. D., Vogel, R. R., & Fleming, L. L. (1993). Myoelectric prostheses. A long-term follow-up and a study of the use of alternate prostheses. *Journal of Bone and Joint Surgery, American Volume, 75*, 1781–1789.

Smits-Engelsman, B. C., Van Galen, G. P., & Duysens, J. (2002). The breakdown of Fitts' law in rapid, reciprocal aiming movements. *Experimental Brain Research, 145*, 222–230.

Smyrnis, N., Evdokimidis, I., Constantinidis, T. S., & Kastrinakis, G. (2000). Speed–accuracy trade-off in the performance of pointing movements in different directions in two-dimensional space. *Experimental Brain Research, 134*, 21–31.

So, R. H., Chung, G. K., & Goonetilleke, R. S. (1999). Target-directed head movements in a head-coupled virtual environment: Predicting the effects of lags using Fitts' law. *Human Factors, 41*, 474–486.

Spence, I. (1990). Visual psychophysics of simple graphical elements. *Journal of Experimental Psychology: Human Perception and Performance, 16*, 683–692.

Steriadis, C. E., & Constantinou, P. (2003). Designing human–computer interfaces for quadriplegic people. *ACM Transactions on Computer–Human Interaction, 10*, 87–118.

Ter Haar Romeny, B. M., Van der Gon, J. J., & Gielen, C. C. (1984). Relation between location of a motor unit in the human biceps brachii and its critical firing levels for different tasks. *Experimental Neurology, 85*, 631–650.

Thorsen, R., Spadone, R., & Ferrarin, M. (2001). A pilot study of myoelectrically controlled FES of upper extremity. *IEEE Transactions on Neural Systems and Rehabilitation Engineering, 9*, 161–168.

Trejo, L. J., Wheeler, K. R., Jorgensen, C. C., Rossipal, R., Clanton, S. T., Matthews, B., et al. (2003). Multimodal Neuroelectric Interface Development. *IEEE. Transactions on Neural Systems and Rehabilitation Engineering, 11*, 199–204.

Viviani, P., & Terzuolo, C. (1982). Trajectory determines movement dynamics. *Neuroscience, 7*, 431–437.

Webb, C. (1977). The use of myoelectric feedback in teaching facial expression to the blind. *Biofeedback and Self Regulation, 2*, 147–160.

Wieselmann-Penkner, K., Janda, M., Lorenzoni, M., & Polansky, R. (2001). A comparison of the muscular relaxation effect of TENS and EMG-biofeedback in patients with bruxism. *Journal of Oral Rehabilitation, 28*, 849–853.

William, K., & Durfee, W. K. (1994). Control of prosthetic gait. *Current Opinion in Neurobiology, 4*, 920–923.

Woodworth, R. S. (1899). The accuracy of voluntary movement. *Psychological Review, 3* (whole No. 13).

Yang, N., Zhang, M., Huang, C., & Jin, D. (2002). Motion quality evaluation of upper limb target-reaching movements. *Medical Engineering and Physics, 24*, 115–120.

Zecca, M., Micera, S., Carrozza, M. C., & Dario, P. (2002). Control of multifunctional prosthetic hands by processing the electromyographic signal. *Critical Reviews in Biomedical Engineering, 30*, 459–485.