CHARACTERISTICS OF MUSCLE CO-CONTRACTION DURING ISOMETRIC TRACKING

by

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(ABSTRACT)

The purpose of this research was to study the relationship between muscle coordination and the performance of a simple manual tracking task. The study employed an isometric, zero order, pursuit tracking task with a laterally translating, periodic sine wave forcing function. The speed of the target was varied by altering the Frequency (3 levels) of the simple sine wave. The control/response ratio for each trial was manipulated by requiring a percentage of each subject's flexion and extension maximum voluntary contraction effort (MVC, 5 levels) to track the target. Multiple electromyograms (EMGs) of the biceps and triceps muscle groups were taken to observe flexor and extensor activity during the tracking task. Muscle modeling techniques were used to quantify the force contributions from the biceps and triceps to the observed tracking force.

It was hypothesized that significant levels of co-active muscle effort would be present during the tracking task and that this co-contraction would have a unique characteristic function about the tracking conditions which yielded optimal tracking performance. The dependent measures investigated were the absolute tracking error as a proportion of the required tracking force (proportional error, PE), the absolute antagonist muscle force (AAF), and the ratio of antagonist to agonist force (co-contraction ratio, CR). Each muscle group's maximum muscle force (MMF) required to track each condition was also determined. The experimental design was a 3 by 5 by 2 mixed factor, repeated measures ANOVA with Gender (5 male, 5 female) as the blocking variable.

The ANOVA results revealed that both target Frequency and tracking Force level had significant effects on tracking error (PE). Orthogonal polynomial contrasts showed that the Frequency effect was characteristically linear while the Force effect was quadratic in nature. A polynomial regression function was used to predict PE from the Force and Frequency conditions. This model accounted for over 96% of the variance in the PE cell means. Further analysis revealed the optimal Force level for isometric tracking to be approximately 61% MVC.

Analysis of the force contributions from each muscle group revealed quadratic relationships for the actual muscle force (%MMF) of the biceps during flexion and of the triceps during extension. These results show that optimal tracking performance during flexion occurs at approximately 66% of the biceps MMF and 65% of the triceps MMF during extension. Actual MMF values were consistently larger than net force values indicating that due to the presence of co-contraction, the measured force output at the wrist underestimated the actual muscle forces involved in tracking .

Neither Force or Frequency had significant effects on absolute co-activity (AAF) showing that antagonist activity remained largely constant over the tracking conditions. However, co-activity was higher for the extension phase than for the flexion phase of the task.

Both Force and Frequency had significant effects on the co-contraction ratio (CR). However, no characteristic function of co-activity was found to explain the optimal tracking performance at median levels of flexion and extension force. CR increased with increasing target speed (Frequency) while it decreased with higher tracking Force levels. Since antagonist activity (AAF) remained almost constant, these results for CR must be due to changes in the level of agonist activity needed to perform the tracking task.

Higher co-contraction was also found during decreasing force production (release) than for increasing force production (exertion). Since there was no significant difference in tracking error for these parts of the task, co-activity may serve to facilitate tracking performance by controlling the rate of force release.

Dedicated to my father and mentor, Dr. Thomas H. Rockwell

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1. INTRODUCTION

1.1 Rationale

The primary objective of the human factors engineer is to design the task environment with an understanding of the capabilities and limitations of the human. To this end, there has been a great deal of effort put into the design and development of manual controls for human-machine systems. Recent interest in tele-operated and fly-by-wire control systems has generated new questions concerning the proper design of system control gain. Control gain (or, reciprocally, control/response (C/R) ratio) refers to a system's sensitivity to a given control input. Although it is generally believed that optimal C/R ratios exists for the design of control systems (Frost, 1972), there is little quantitative research available on which to base specific control design decisions. In fact, the control gain parameters are often found by trial and error (Sanders and McCormick, 1987).

The properly designed C/R ratio is often considered to be a trade-off between controlled element travel time and adjustment time (Sanders and McCormick, 1987). Research involving subjective assessment of control systems has shown that a preferred gain does exists for manual control (Hess, 1973; McRuer and Jex, 1967) but that this gain is not characterized by input force or error metrics.

In an attempt to develop quantitative metrics from which to specify the C/R ratio, Berkowitz (1990) conducted a study which examined simple sine wave tracking using an isometric control. He showed that an optimal, isometric control/response ratio exists for a one-dimensional, zero-order, pursuit tracking task. By developing a gain sensitive error metric, Berkowitz found that the optimal force level required for the tracking task was approximately 65% of the subjects maximum extension voluntary contraction (MVC_e) for elbow extension. Subjective workload assessment techniques confirmed his results.

By finding an optimal C/R ratio in terms of input force effort, Berkowitz (1990) revealed that there may be specific motor strategies involved in the manual control of systems. Specifically, there may be an underlying coordinated muscular strategy which promotes optimal tracking performance at median levels of input force.

It is possible that the muscles involved in the isometric tracking task contract simultaneously in such a way that tracking performance is best at median levels of activation. The simultaneous contraction of opposing muscle groups is known as co-activation or agonist/antagonist co-contraction. Co-contraction has been shown to exist in dynamic exertions (Marsden, Obeso, and Rothwell 1983; Redfern, 1988; Smith, 1981; Woldstad, 1989) and in isometric efforts (DeLuca and Mambrito, 1987; Ghez and Gordon, 1987; Redfern, 1988; Tyler and Hutton, 1986; Woldstad, Chaffin, and Langolf, 1988). Using electromyography (EMG), Woldstad et al. (1988) found significant levels of co-contraction during an isometric sinusoidal replication task. They suggested that co-contraction during the isometric exertions may serve to maintain joint stability during the exertion. The level and timing of co-contraction during an isometric tracking task may also help facilitate tracking performance.

An understanding of the coordinated muscle activity in isometric tracking may allow designers to provide suitable guidelines for the selection of proper control characteristics for human-machine systems. This understanding might be particularly useful for the design of isometric fly-by-wire and tele-operated controls as well as controls used in a constrained operating environment. Additionally, designers may more fully understand how the human neuromuscular system carries manual control tasks.

1.2 Experimental Approach and Objectives

The purpose of this research was to study the relationship between muscle coordination and the performance of a simple manual tracking task. The experimental methods were a combination of the techniques used by Berkowitz (1990) to study preferred force levels during isometric tracking and Woldstad et al. (1988) to study muscle co-contraction during isometric elbow exertions. This experiment studied the isometric exertions of subjects performing several tracking tasks using stationary wrist cuff equipped with a force sensing transducer. The task was the digital equivalent of an isometric, zero order, pursuit tracking task with a laterally translating, periodic sine wave forcing function. Force measurement and electromyography (EMG) were used to investigate the effects of C/R ratio and target velocity on tracking error and muscle co-activation during the tracking task. The speed of the target was varied by altering the Frequency of the simple sine wave. The control/response ratio for each trial was manipulated by requiring a percentage of each subject's flexion and extension maximum voluntary contraction (MVC) effort to track the target. Muscle modeling techniques were used to determine the agonist and antagonist contributions to the tracking force output during the tracking task. The following hypotheses were examined:

H_1 Significant levels of muscular co-contraction are present in the performance of closed-loop tracking tasks.

H_2 Co-contraction is a second order, quadratic function of required tracking force revealing an optimal level of agonist/antagonist muscle activity resulting in optimal tracking performance.

H_3 Co-contraction is higher for isometric extension than for isometric flexion of the elbow.

2. BACKGROUND

2.1 Overview

The following sections discuss topics relevant to the study of neuromuscular behavior during manual control tasks. Although the study of manual control is founded in the areas of kinesiology, physiology, experimental psychology, human factors engineering, and control theory engineering, only those areas relevant to the current research will be discussed in detail.

First, an overview of manual control and human movement is presented. This includes a brief discussion of manual control modeling and previous attempts to describe the neuromuscular aspects of manual control. The second section elaborates on recent work regarding the setting of the control/response ratio for tracking tasks. This is followed by a general discussion of the production of isometric force and isometric force variability in human exertions. The fourth section addresses the theories behind the existence of co-contraction during manual tasks. Finally, a discussion on electromyography (EMG) and human force production is presented. This includes the use of EMG in modeling the relation between muscle force and the neuromuscular system. The background section is concluded by a summary of the literature relevant to the experiment.

2.2 Human Movement and Manual Control

2.2.1 Human Movement

Control of human movement can be divided into three categories: reflexive, open-loop (skill based), and closed-loop (tracking). Reflexive movements are those in which no higher-order cognitive processing is involved. Since they are initiated and carried out completely at the spinal level, there is no voluntary control over these actions. One example of a reflexive movement is the stretch reflex.

Open-loop tasks, often called ballistic or skill-based tasks, are entirely preprogrammed within the brain before execution (Poulton, 1981; Wickens, 1984). Throwing a ball or swinging a bat are examples of ballistic movements. Once selected from memory and initiated, these movement are no longer under the control of the continuous sensory input and action selection process. These movements are considered in the realm of manual control because they are voluntary and because the programs can be modified slightly between executions in order to meet specific movement objectives.

Open-loop control becomes closed-loop when the output (i.e., the difference between the actual and desired response) is used as a continuous input to modify the movement response (Sheridan and Ferrell, 1974). This is analogous to servo-control in which the current output is completely dependent on the result (error) of the last response. Closed loop tasks utilize a continuous feedback loop to insure that the task objective is met. This continuous correction in movement is often termed "current control" (Woodworth, 1899). An example of closed-loop control is an aimed hand movement. While, this movement may be initiated by a preprogrammed or ballistic cue, it soon comes under continuous corrective control as it approaches the target to ensure movement accuracy. Some form of feedback is used to facilitate this corrective control. This feedback is usually perceived in the auditory, visual, kinesthetic, or proprioreceptive sensory modalities. Figure 2.1 shows a qualitative model of closed-loop movement (Sheridan and Ferrell, 1977). The figure shows how the search and movement programs are continually updated using the input from sensory feedback in coordination with the higher order perception and cognition centers of the brain. The timing, accuracy, and appropriateness of the response are all dependent on the interactions of the components within the system. When closed-loop movements involve the control of an operator-machine system, the result is often termed manual control.

Human manual control pertains to the use of actuators to govern the state of a system (Frost, 1972). The goal in a manual control task is to use current system state information to minimize the difference between the system's actual and desired states by manipulating the control(s). The human performing a manual control task acts much like a servo-mechanism by directing output responses based solely on input information (Sheridan and Ferrell, 1974).

There are many areas involved in the human aspects of manual control. These areas include perception, cognition, learning, practice, and the motor execution of manual tasks (Figure 2.1). The capabilities and limitations of voluntary actions are governed by the many coordinated human elements involved such as sensory input, short and long term memory, response

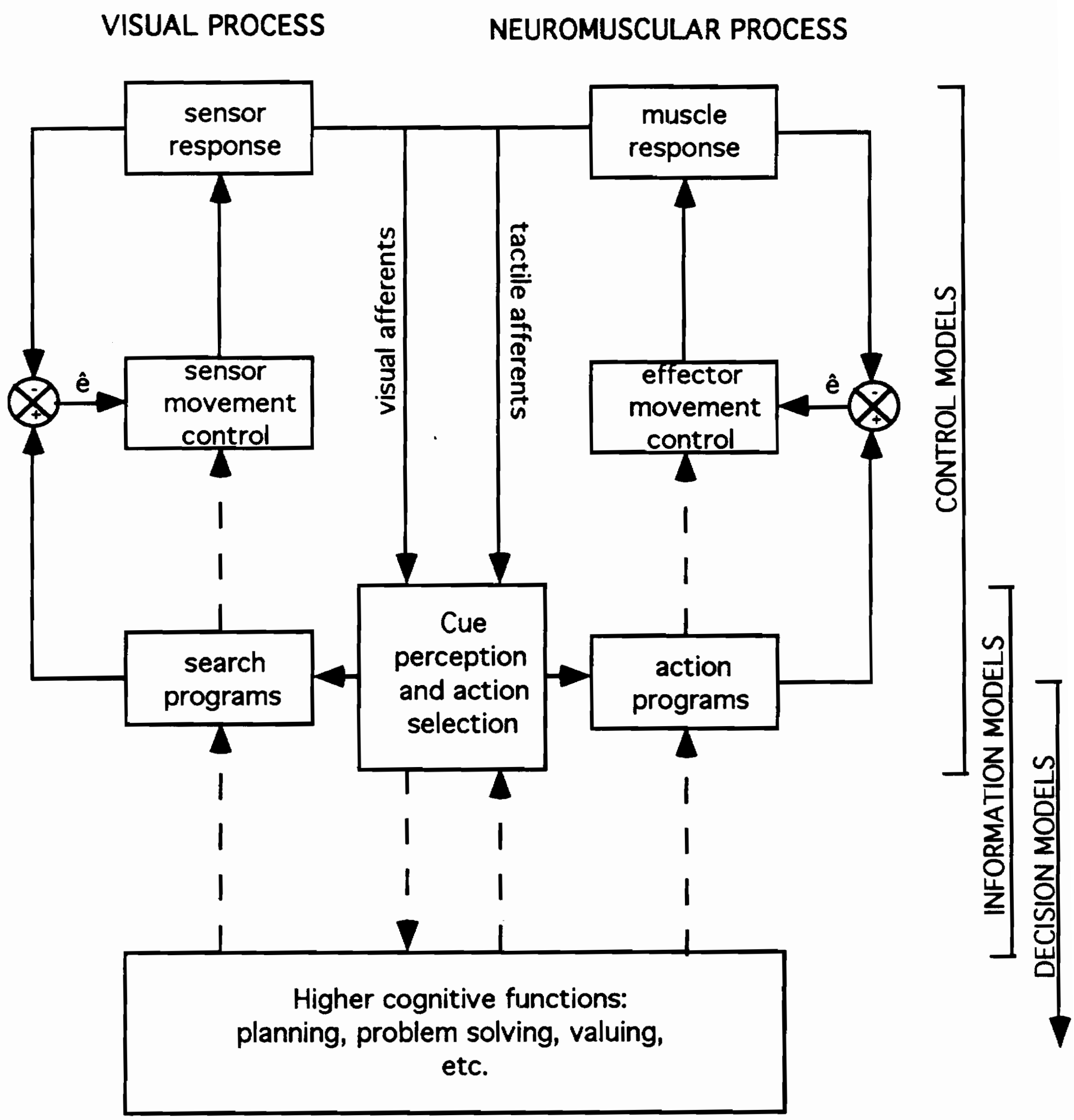


Figure 2.1 Model of human movement (adapted from Sheridan and Ferrell, 1974).

selection, and neuromuscular execution of the desired movement (Poulton, 1981).

2.2.2 Manual Control Modeling

Scientific interest and engineering design applications have led to modeling techniques that describe human manual control performance under various control and system dynamics. Manual control modeling is a commonly used technique which employs engineering control theory to describe the relationship between the actual state (output) and the desired state (input) of a system by using transfer functions. Linear transfer functions are most often used. These functions describe the difference between system states in terms of gain (amplitude) and phase (time lag or lead) relationships between the fundamental components of the system input and human control output. Transfer functions are commonly depicted in the frequency domain rather than the time domain using Laplace transformations. Analysis of systems in the frequency domain allow the design of stable systems with low human error.

Figure 2.2 is a block diagram of a compensatory tracking task. Note that each element has its own transfer function (capital letters) relating the corresponding input and output functions (small letters). For example, the operator experiences a target moving with a predetermined forcing function, i(t), compares it to the controlled element output position, o(t), to obtain an error measure, e(t). The human operator then analyzes the error and initiates a control response, f(t). The controlled element responds with $o(t+\Delta t)$ and the process repeats until the desired system response is obtained (Sheridan and

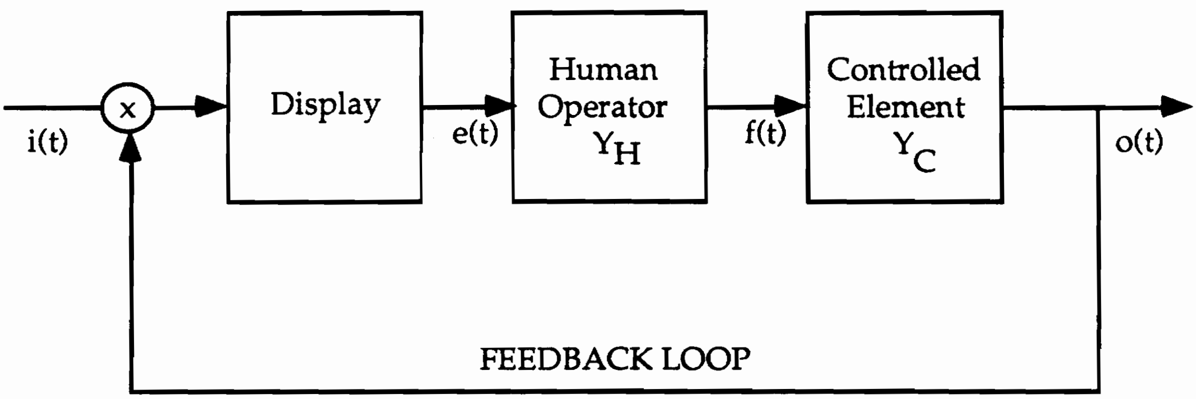


Figure 2.2 Block diagram of compensatory tracking showing transfer elements (adapted from Wickens, 1984).

Ferrell, 1974; Wickens 1984). The human operator acts as a servo, constantly using system feedback to analyze and update control commands. Although Tustin (1947 as cited in McRuer and Jex, 1967) was one of the first researchers to model the human as a servo-controller, others have used his work to develop the models used in human-control modeling today.

<u>2.2.2.1 Crossover Model</u>. The system model shown in Figure 2.2 allows for the isolation and description of human tracking behavior. Initial efforts described the human using quasi-linear models, with the realization that the operator only approximates first-order, linear behavior. The most widely known of these early models was the crossover model (McRuer and Jex, 1967; McRuer and Krendel, 1959). The function of the operator (human describing function) was a first order system with gain (K) and effective time delay (τ). In the Laplace domain, the operator describing function based on the crossover model is:

$$Y_H = K\frac{(T_L s + 1)e^{-\tau s}}{(T_I s + 1)(T_N s + 1)} \qquad (2.1),$$

where

K = pilot gain (ratio of output velocity to perceived error: Wickens, 1984)

τ = effective reaction time delay

T_L, T_I = lag and lead time equalization coefficients

T_N = neuromuscular lag coefficient

The equalization coefficients, T_L and T_I, make up for the inherent time delays due to operator reaction time and neuromuscular processes. Thus, the operator unknowingly acts as a well designed servo-controller constantly striving to minimize system error through adaptation to the task (Sheridan and Ferrell, 1974). Specifically, the human adapts to make the system act as a first order (rate) control system with first order lag (Wickens, 1984).

The human describing function, Y_H, when evaluated in the frequency domain, shows the frequency at which the system becomes unstable. System unstability is defined as the point(s) at which the system gain is greater than one while the system phase lag is greater than 180 degrees. At this crossover frequency, any corrective movements intended to reduce system error will actually result in greater discrepancy between the actual and desired system states. The model reveals this point of crossover from system stability to instability. This is why the model is called the crossover model. Researchers have found this model to be successful in predicting human tracking behavior (Wickens, 1984).

2.2.2.2 Optimal Control Model. The most important limitation of the crossover model is that it does not account for the adaptability and flexibility of the real operator. Actual system operators constantly modify their control strategies based on memory and knowledge of previous success and failure (Sutton, 1990). Also of importance is the portion of tracking error not predicted by the crossover model – commonly known as remnant. Remnant error is due to elements such as high frequency muscle tremor and quick corrections of the control (Frost, 1972; Poulton, 1981). Thus, the major shortcoming of the crossover model is that it does not account for these nonlinear elements in human tracking behavior.

An improved model was developed to better account for human adaptability and nonlinear remnant components of the human in the tracking loop. The theory behind the model is that the operator seeks to optimize some performance criterion given (Sheridan et al., 1974):

- the process being controlled,
- the accuracy with which the operator can determine the state of the system, and
- the energy or time available for system control.

The model attempts to determine how the human operator *should* perform based on the task requirements and inherent psychophysical limitations. The most widely known of these models is called the optimal control model (Baron, Kleinman, and Levison, 1970; Kleinman, Baron, and Levison, 1970). The operator's goal is to minimize the quadratic cost functional:

$$J = \int (\mathrm{Au}^2 + \mathrm{Be}^2)\,dt \qquad (2.2),$$

where the quantity in parentheses is the weighted combination of squared error (e^2) and squared control effort (u^2). The weights (A, B) can be determined objectively or subjectively. This combination is optimized such that there is a trade-off between allowable system error and the control effort required to maintain that error.

The human control characteristic is also subject to time delays and disturbance noise at the observation or motor levels of system control. Observation noise produces a noisy image of the system output to the operator. This causes some error in system state evaluation and the corresponding response selection. Similarly, motor noise induces execution error at the neuromuscular level. Execution error can be called impulse variability (Section 2.4.3).

Both the crossover and optimal control models made significant advances in describing human tracking behavior. The optimal control model, which addresses the changes in human strategy and adaptive behavior, has proved to be a more accurate predictor of manual control behavior (Wickens, 1984).

2.2.3. Neuromuscular Modeling in Manual Control

Attempts to develop more detailed human models using the control theoretic approach resulted in transfer functions describing the central nervous and neuromuscular sub-systems of the human operator. Of particular interest to the present research are the few models that pertain to the neuromuscular system during tracking performance (Hess, 1980; Jex, 1971; Magdaleno and McRuer, 1971). These models use transfer function representations of the known physiologic relationships in the neuromuscular system.

The majority of these models separate the motor process into the alpha (α) motor neuron and the spindle afferent networks which combine ($\alpha + \gamma$ efferents) to regulate muscle force (Hess, 1980; Jex, 1971; Magdaleno and McRuer, 1971). However, more complete models include proprioreceptive, position feedback from the joint centers (Jex, 1971; Magdaleno and McRuer, 1971), and force feedback from the Golgi tendon inhibitory network (Jex, 1971).

One of the most advanced of these models was developed by Jex (1971). His model uses a very detailed neuromuscular servo interacting with the manipulator dynamics. Of particular interest is the use of second-order components to represent the muscle's contractile elasticity and damping components (Figure 2.3). The most important pitfall of this model, as it pertains to the current research, is that it describes the net activation of the agonist/antagonist muscle pairs rather than the individual contributions of each muscle within the pair. Since the observed net force is the combined output of the forces from the agonist/antagonist pair, it is impossible to understand, or even describe, the active coordination of these muscle pairs in facilitating manual performance.

In summary, the control theory approach to manual control modeling represents the human operator as a "black box"; *describing* the behavior of the human operator but not *explaining* the mechanisms behind that behavior. Even though these models can provide some measures of system stability, they are largely descriptive and oversimplified in their assumptions of the human as a linear controller. The human can adapt, change strategy, and use memory to predict future system states from the past. These all influence the nonlinear behavior of the operator involved in a closed-loop manual task. In short, manual control modeling is a useful tool but of limited use due to the complexity of the human element of systems. More explanatory models of the human are needed to determine the mechanisms used while performing manual control tasks. The purpose of this research is

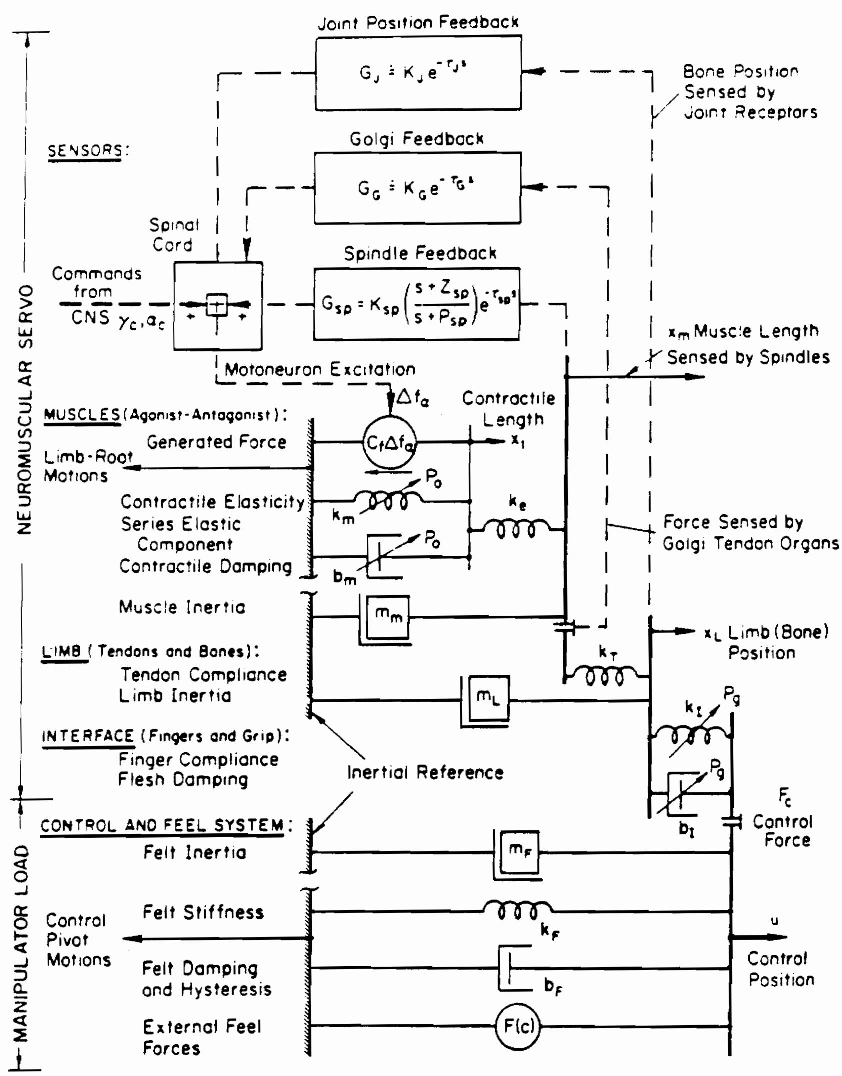


Figure 2.3 Second order differential control model of the neuromuscular control system (Jex, 1971).

to begin to understand and quantify the neuromuscular nature of humans performing manual control tasks.

2.3 Tracking with Isometric Controls

2.3.1 Tracking

Tracking is often used to study human performance in a manual control system. In tracking, the operator attempts to continuously follow a moving target by manipulating a control. The operator uses the position of the target and of the cursor to update the appropriate response; the goal being to minimize the error between the controlled element (cursor) position and the target position. In laboratory environments, the tracking task is usually performed using a control mock-up and a display which contains the target and cursor representations. Specific control and display types are used to represent the system of interest and evaluate the changes in human behavior under varying target trajectories and control parameters.

Tracking involves not only the human characteristics of voluntary movement (Figure 2.1) but also the characteristics of the control input dynamics, and system response (display) dynamics (Figure 2.2). Some of the factors that influence human behavior in tracking tasks are (Frost, 1972):

- the display type (compensatory or pursuit)
- display compensation technique (quickening, preview, etc.)
- control type (isotonic, isometric or real control)
- system response dynamics (position, velocity, acceleration, or jerk)
- system response lag order
- target forcing function (periodic, random)
- control/response ratio

A discussion of all of these aspects of tracking is beyond the scope of this research. They are mentioned here as the background characteristics of tracking. The tracking task for this research will be similar to that used by Berkowitz (1990): a zero-order, one-dimensional, pursuit tracking task with zero lag using an isometric controller. The task will employ a laterally translating, periodic sine wave forcing function. This task allows the study of neuromuscular coordination of muscle activity by representing a simple form of a closed-loop task (McRuer and Jex, 1967; Poulton, 1981) and by eliminating the effects of limb movement dynamics by the use of an isometric control. The reasons for using these tracking task characteristics will be discussed in greater detail in the following sections.

2.3.2 Periodic Sine Wave Tracking

Manual control requiring the operator's response to a periodic, nearly sinusoidal, forcing function is common in many vehicle control tasks (McRuer and Jex, 1967). In periodic tracking, the operator can pick-up the "rhythm" of the task after only a few oscillations of the forcing function. After some practice, the operator can synchronously track the wave and the task becomes almost completely preprogrammed or open-loop (Frost, 1972; McRuer and Jex, 1967). The tracking task begins as closed-loop, requiring visual feedback, and may eventually becomes open-loop, with no need for that feedback. From 0 to 2 Hz the operator can adequately track the wave but with increasing time lag. The possibility of preprogramming during this frequency range exists if the time available for tracking is of sufficient length. From 2 to 5 Hz the task becomes completely open-loop as the visual process is no longer efficient and the limit of the neuromuscular response rate is reached (McRuer and Jex, 1967). Here both phase and frequency drift occurs. To observe closed-loop tracking behavior, the periodic function must be less than 2 Hz and the duration of tracking should be kept as small as possible.

2.3.3 Isometric Tracking

The isometric control has been of interest to designers since it operates on force input only, thus requiring no control displacement and no mechanical linkage. Isometric controls can be used in fly-by-wire systems, tele-operated control systems, and systems where controls are used in a constrained operating environment. The control/response ratio (C/R) of these systems can be changed easily through the use of computers; making the control more adaptive to individual human characteristics. Unfortunately, there are no specific guidelines to help designers select appropriate C/R ratios for human-machine systems (Berkowitz, 1990; Schopper, 1987). In fact, the control parameters are often found by trial and error (Sanders and McCormick, 1987).

Schopper (1987) studied the effects of changing force level on tracking accuracy during a two dimensional pursuit tracking task with linear and circular elements. The linear part of the track included reference force levels of 22.25 N, 44.5 N and 66.75 N. He measured tracking error as lag or lead in centimeters at each reference force level. Schopper's results showed force to be a significant determinant of tracking performance with higher force levels causing larger tracking errors. However, the error at the highest force level (66.75 N) was not included in the linear tracking analysis as it represented a change in direction of tracking. Also, females were not always capable of reaching the 66.75 N reference point. Although Schopper (1987) investigated tracking error at different force levels, he tested only one control response ratio. Also, the discrete force levels chosen for the study represent varying degrees of difficulty for different operators.

Berkowitz (1990) attempted to overcome some of Schopper's (1987) limitations by studying tracking performance as a function of C/R ratio and target speed. The task used was a zero-order, one-dimensional, pursuit tracking task with zero lag using an isometric joystick controller. The flexors and extensors of the elbow were isolated for the task by fixing the subject's posture at 90° shoulder abduction, 0° shoulder flexion, and 90° included elbow angle with the hand in the semi-supinated (thumb-up) position. A laterally translating sinusoidal forcing function was used to vary target speed. The C/R ratio was controlled by varying the force required to track the target as a percentage of the maximum voluntary contraction force produced during isometric elbow extension for each subject (%MVC_e). Because subjects usually cannot exert as much force in elbow extension as they can in elbow flexion (with elbow at an included elbow angle of 90°), extension force was used as the limit of the required tracking force. Berkowitz (1990) used five levels of required, maximum tracking force (10, 25, 50, 75, and 100% MVC_e) and five levels of forcing frequency (0.2, 0.4, 0.6, 0.8, and 1.0 Hz.).

Berkowitz (1990) hypothesized that tracking performance would degrade as both the target speed and the force level required to track the target increased. The performance metric was average absolute tracking error (in Newtons) as a proportion of the tracking force required for the trial condition. This proportional tracking error (PE) was expressed as:

$$PE = \frac{\sum \frac{\text{ABS(Forcing Function (N) - Actual Track (N))}}{\text{Max. Required Force Level for Trial (N)}}}{\text{Number of Observations}} \qquad (2.3).$$

This measure represents the average absolute difference between the position of the target (desired position) and the subject's actual tracking position as a function of required control effort.

Berkowitz's results showed significant effects of both target speed (frequency) and tracking force for PE at the $\alpha = 0.05$ level. Proportional tracking error was an increasing linear function of tracking frequency, and a convex quadratic function of required tracking force. The quadratic nature of the force effect was consistent between target frequency conditions as the frequency by force interaction effect for PE was not significant.

Although his methodology was sound, Berkowitz's (1990) procedure introduces some questions. The periodic forcing frequencies were well within the closed-loop capability of the operator (0.2-1.0 Hz). However, the extensive length of the trials (20 second warm-up and an adjacent 20 second trial) may have allowed for complete preprogramming of the task response (through practice) by the time data was collected (Frost, 1972; McRuer and Jex, 1967). Therefore, the nature of the task may have changed from the desired closed-loop pursuit task to a completely open-loop task; measuring only the ability of subjects to synchronously track a target using open-loop mechanisms. The increased length of the experimental trials could also have introduced fatigue effects into the data. This could significantly effect the observed tracking error, particularly during the trial conditions requiring high force levels.

Berkowitz (1990) scaled the C/R ratio as a percent of each subject's extension MVC in an attempt to normalize each subject's performance with respect to individual strength capability. This was an excellent way to ensure that each subject was performing at the same level of difficulty relative to his/her maximum strength, thus reducing subject selection bias. However, using required tracking force levels derived only from each subject's maximum elbow *extension* strength leads to the possibility of subject bias during the *flexion* part of the sine-wave track. This bias is caused by requiring different levels of relative effort from each subject during flexion tracking.

2.3.4 Optimal Control Response Gain

Berkowitz (1990) found that tracking error exhibited a convex quadratic function across required force levels (Figure 2.4). The linear characteristics of frequency and the quadratic characteristics of force were used to model subject error performance (PE). PE was described using a polynomial regression equation of the form:

$$PE = \beta_1 + \beta_2 \text{Frequency} + \beta_3 \text{Force} + \beta_4 \text{Force}^2 \qquad (2.4).$$

This model accounted for over 95% of the variance in tracking error, and predicts a zone of optimal control at median force levels (Figure 2.4). By partially differentiating Equation 2.4 with respect to force, Berkowitz found that the optimal force level for isometric tracking was found at approximately

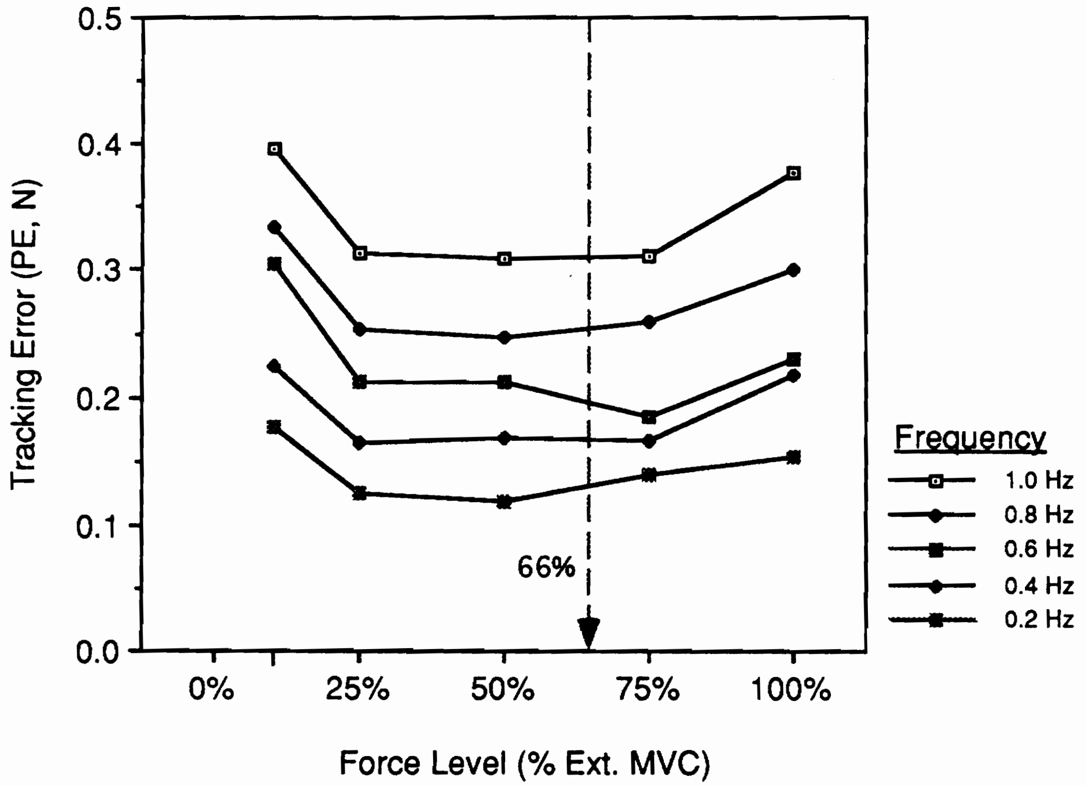


Figure 2.4 Two-way interaction of target frequency and required tracking force showing the zone of optimal control at approximately 66% MVC_e (adapted from Berkowitz, 1990).

66% MVC_e. Berkowitz's empirical finding of the U-shaped function of control force for the isometric task was reinforced by the subjective rating of a subset of the trial conditions by each subject. The subjective results found a zone of optimal control at approximately the 50% MVC_e. Other researchers have found a subjective zone of optimal control as a function of control gain (Hess, 1973; McRuer and Jex, 1967).

Hess (1973) showed the existence of a subjective optimal zone of control gain for an isometric control task. He used a compensatory tracking task using four levels for the control/response ratio. Since the size of the tracking display was not given, it is not possible to convert these C/R ratios to required force levels. However, a U-shaped curve was generated using the subjective ratings of operators performing the tracking task, indicating a zone of optimal control gain.

McRuer and Jex (1967) present a review of previous research finding U-shaped functions for pilot ratings of handling quality over varying system control/response characteristics (Figure 2.5). Although there were few details on the nature of the studies reviewed, the research used single axis compensatory tracking tasks with zero- and first-order control gains. McRuer and Jex (1967) state "The subjective rating trends associated with the controlled element gain are the best known, for numerous experiments have shown that an 'optimum' gain exists...which is not revealed by either the error or force criteria..."(p. 246).

The fact that the actual error scores did not indicate a level of optimal gain is not in conflict with the empirical findings of Berkowitz (1990). More

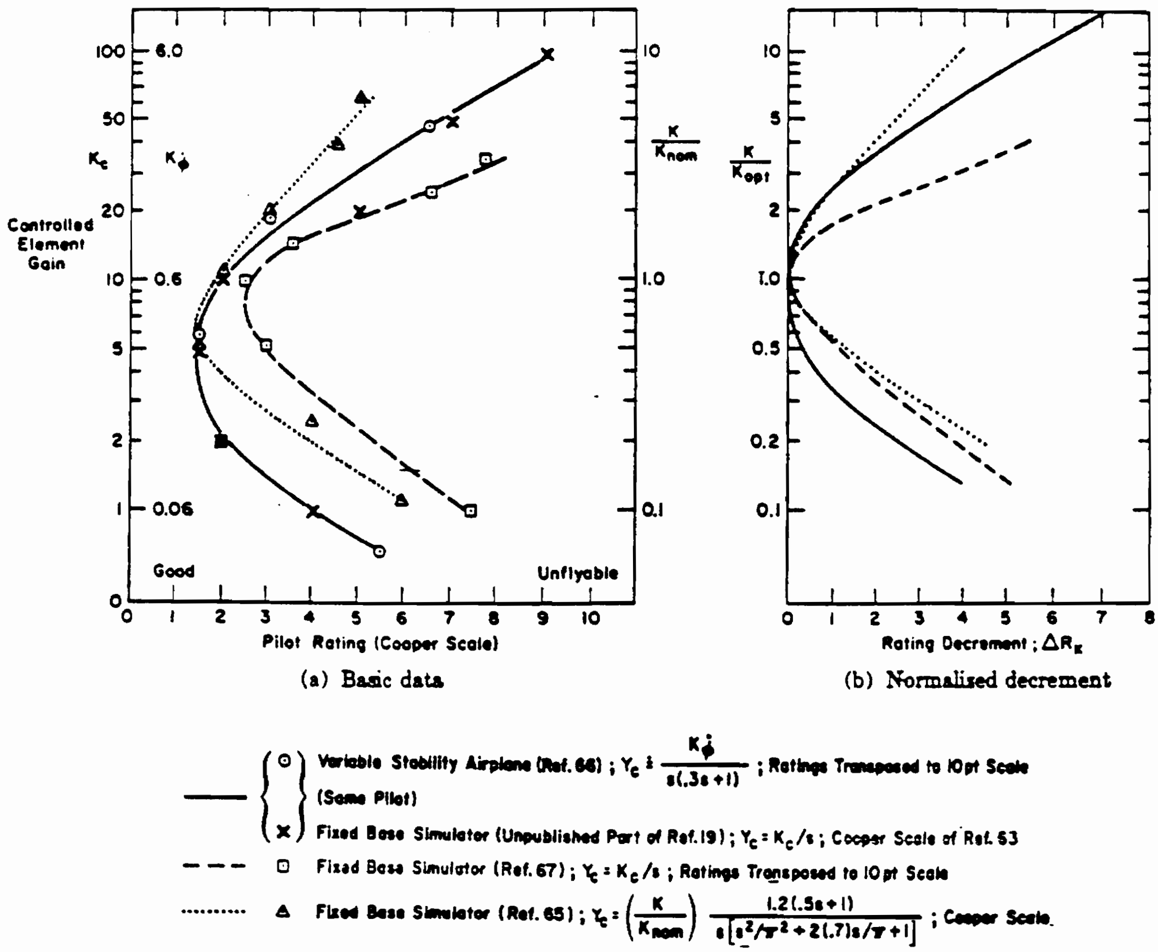


Figure 2.5 Subjective optimal control gain found for different control tasks (McRuer and Jex, 1967).

traditional error measures such as root-mean-square (RMS) error, absolute error, or average error (Poulton, 1974) express overall error rather than the proportional error scaled to the C/R ratio being used (Berkowitz, 1990). The choice of error metric is critical to the interpretation of the tracking performance results. The more traditional error measures do not take into account differences in the gain characteristics of the task. For example, a 2 N input error using a sensitive control would be greater than the same 2 N error using a insensitive control. Therefore, error as a proportion of the C/R ratio (PE) removes the effect of the control gain and is a representative measure of tracking error.

2.4 Production and Regulation of Muscle Force

2.4.1 Muscles of the Elbow

The upper limb of the human is comprised of three major skeletal bones. The radius and ulna of the lower arm connect with the humerus of the upper arm at the elbow joint. The humerus extends from the elbow to the shoulder which connects to the torso.

The elbow is a hinge joint allowing for flexion and extension of the upper and lower segments of the arm. Flexion serves to reduce the included angle at the elbow and extension increases this angle. In addition, supination and pronation of the forearm is performed by rotating the radius across the ulna.

Four groups of muscles control movement at the elbow: the flexors, extensors, pronators, and supinators. This research is concerned with the flexion and extension exertions of the elbow joint. These flexors and extensors work antagonistically, rather than synergistically, in the production of elbow movement.

The anterior compartment of the arm holds the primary muscles contributing to the flexor group. The primary flexor muscles include the biceps brachii, brachialis, and the brachioradialis of the forearm. The biceps brachii, as the primary flexor, has two heads originating at the scapula and a common insertion point at the elbow (Leeson and Leeson, 1989). Both of the biceps heads are superficial. The two biceps heads are controlled by the central nervous system through a branch of the musculocutaneous nerve originating at the fifth and sixth cervical vertebrae. The biceps also contribute to forearm supination and shoulder flexion.

The posterior compartment of the upper arm contains the muscles that contribute to the extensor group. The primary elbow extensor muscles are the triceps brachii and the anconeous muscles. The triceps brachii, as the primary extensor, has three heads: the long head, the lateral head, and the medial head. The triceps long head originates at the scapula while the lateral and medial heads originate on the medial portion of the humerus. The common insertion point for the triceps is the upper surface of the olecranon of the ulna (Leeson and Leeson, 1989). The triceps are controlled by branches of the radial nerve originating at the seventh and eighth cervical vertebrae.

This research concerns the elbow flexor and extensor activity associated with an isometric tracking task. Externally measured torque is the net result of the activity from the several flexor and extensor muscles. For the purposes of this research, flexion and extension of the elbow is assumed to be represented proportionately by the activity of the primary flexor and extensor muscles: the biceps and triceps brachii, respectively. The muscles are assumed to be the *muscle equivalents* of the entire flexor and extensor groups (Bouisset, Lestienne, and Maton, 1977; Bozec, Maton, and Cnockaert, 1980; Woldstad, 1989). The definition of an *equivalent muscle group* states that although other muscles may contribute to the function of the group, activity in these muscles is proportional to the activity of the predominate muscle within the group (Woldstad, 1989). In other words, flexor and extensor force can be represented by the observed activity in the biceps and triceps muscles.

To study the coordinated agonist/antagonist muscle strategies involved in lateral, isometric tracking, the flexor and extensor muscles of the elbow need to be isolated. This isolation minimizes the contribution of the other muscles in the body to measured force during the tracking task. The flexor and extensor muscles of the elbow can be isolated with a posture of:

- 90° shoulder abduction
- 0° shoulder flexion in the frontal plane
- 90° included elbow angle
- forearm in a semi-supinated position (thumb up).

This posture places the upper arm parallel to the floor with the elbow at a right angle. By using this posture, lateral tracking can only be performed by flexion and extension of the elbow. Again, by the assumption of equivalent muscle groups, the net output force can be represented solely by the activity of the biceps and triceps muscles.

2.4.2 Production of Muscle Force

Muscle contraction is an mechanical process brought about by a chain reaction of electrochemical events originating in the higher brain centers.

When muscle contraction is desired, the brain initiates an electrical impulse which travels through a series of synaptically connected nerve fibers (motoneuron) to the muscle site. The signal then travels through the innervated muscle fibers by means of a chemical reaction. This reaction depolarizes the muscle fibers resulting in muscle contraction. The motoneuron and the muscle fibers that it innervates is called a motor unit (Astrand and Rodahl, 1986). Large motor units, used for coarse control, have more innervated fibers than small motor units which are used for fine control.

When a motor unit is activated, all of the innervated fibers contract. This is known as the "all or nothing response". The higher brain centers regulate muscle force by controlling the number of motor units firing (recruitment), the rate at which these motor units fire (rate coding), and the degree of syncronization between the motor units contracting (Bouisset, 1973). During force production, motor unit recruitment usually follows the "size principle", with smaller units being turned on first followed by the larger motor units (Astrand and Rodahl, 1986). As force production is increased, larger motor units are activated while the active motor units increase in firing frequency, resulting in greater force production (Astrand and Rodahl, 1986).

The contractile characteristics of muscles fibers change as the length of the muscle (length/tension curve) and the velocity with which the limb is moving change. When muscles contract isometrically, there is no change in muscle length. When a muscle is contracted statically, its mechanical properties remain constant. Further, the inertial effects of moving limbs do not effect the contraction.

The coordinated contraction of agonist and antagonist muscles are regulated by several mechanisms. Figure 2.6 shows one model of how the central and spinal level processes influence agonist, antagonist and co-active muscle activity (Basmajian and DeLuca, 1985). In voluntary, goal-directed movements, the primary influence on muscle activity comes from the central nervous system which drives the agonist and antagonist motoneuron pools. However, there are two spinal-level mechanisms that influence muscle activation (Figure 2.6). These mechanisms are the muscle spindle and the Golgi tendon organs located within the muscle and tendon tissue, respectively.

The muscle spindle is an organ within the muscle that monitors and regulates muscle length by direct feedback through the spine. The efferent signals from the central drives set the sensitivity of the muscle spindles to the muscle stretch. When this organ senses an unexpected change in muscle length, it commands the agonist muscle to compensate by contracting harder while simultaneously inhibiting the activity of the antagonist (Basmajian and DeLuca, 1985). This is commonly called the "stretch reflex".

The other spinal level process which affects muscle contraction is the inhibitory Golgi organ located at the muscle tendon. When this organ senses a force overload applied to the muscle, it sends inhibitory signals to the agonist to stop the contraction. Simultaneous, excitatory (contraction) signals

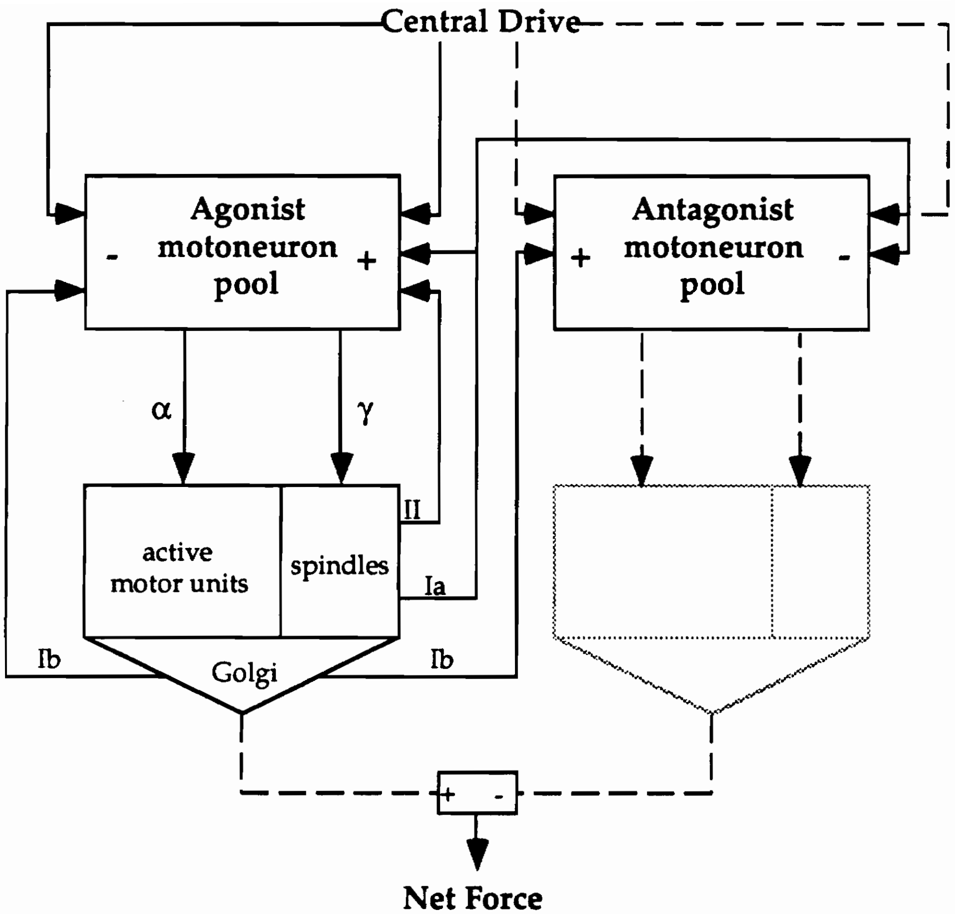


Figure 2.6 Diagram of the interaction of muscle control mechanisms (Basmajian and DeLuca, 1985).

are sent to the antagonist to facilitate rapid relief of the agonist muscle (Basmajian and DeLuca, 1985).

Many internal and external perturbations influence muscle contraction and the outcome of voluntary exertions. This is true even with isometric contractions where no movement is involved. The variability in the control of muscular response contributes to the error in performing precision manual tasks such as tracking.

2.4.3 Control of Force Production

While studying human motor control, the neuromuscular strategies used in the control of movement are of interest, particularly the nature of inaccuracies (errors) in movement. Relevant to the current research are the concepts of the speed-accuracy trade-off, force variability, and timing variability during human movement. The research reviewed below primarily involves the study of rapid, aiming movements and must be applied to the study of closed-loop tracking with caution. However, many of the general principles presented here may be applicable to the study of tracking behavior.

2.4.3.1 Speed Accuracy Trade-off. The study of the speed-accuracy trade-off goes back to the work of Woodworth (1899) and his ideas of ballistic and current control phases of voluntary movement. Paul Fitts (1954) quantified the nature of this trade-off in terms of required accuracy (W) and distance (D) to be moved. His equation is known as Fitts' law:

$$MT = A + B * \log_2(2D/W) \quad (2.5),$$

where MT is the movement time and A and B are coefficients determined by the ease of control use. The logarithmic portion, $\log_2(2D/W)$, of the equation is called the index of difficulty (ID). Fitts' law indicates a logarithmic trade-off between the speed of a movement and its accuracy. Since its inception, Fitts' law has been found to be very robust in determining the speed-accuracy trade-off associated with the use many systems.

Fitts explained his finding in terms of information theory, stating that the ID represented the uncertainty associated with a controlled movement. However, other researchers that followed disagreed with this rather "non-intuitive" explanation for the trade-off (Crossman and Goodeve, 1963/1983; Meyer, Abrams, Kornblum, Wright, and Smith, 1988; Schmidt, Zelaznik, Hawkins, Frank, and Quinn, 1979). These researchers looked for physiological or neuromuscular reasons for the speed-accuracy trade-off.

Crossman and Goodeve (1963/1983) developed the iterative corrections model which stated that the speed-accuracy trade-off was due to the increasing number of discrete submovements (all having equal time) needed to achieve a required accuracy. This followed from the early work of Craik (Poulton, 1981) who believed that tracking behavior was actually a string of discrete movements. Craiks' ratio rule stated that for simple ballistic movements, the movement error would be approximately 10% of the movement distance. Therefore, more precise movements would require more submovements and, therefore, more time to complete. A much more refined model using the idea of submovements was the stochastic optimized submovement model developed by Meyer et al. (1988). Meyer, Smith, Kornblum, Abrams, and Wright (1989) provide an excellent review of research involving the speed-accuracy trade-off.

Both Fitts' Law and the submovement paradigms are based on the idea that variability in movement is due to within-movement error detection and correction. However, some researchers believe that the speed-accuracy trade-off seen in human movement is due to errors made in the initiation of movement. These errors are brought about by inherent noise in the motor system leading to errors in the execution of motor programs producing movement. This motor-output variability is dependent on the parameters (such as movement distance, D, and movement time, MT) used to select and initiate a preprogrammed response. An important finding which supported this theory came with the impulse variability model developed by Schmidt et al. (1979) in studying rapid static and dynamic exertions. This model states that the variability in the accuracy of ballistic movements is proportional to the velocity of the movement:

$$\sigma_{impulse} \propto D/MT = \text{velocity} \qquad (2.6),$$

so that longer-faster movements are more variable than slow-short movements.

A question arises as to which of the two theories of speed-accuracy trade-off is correct: Fitts' logarithmic trade-off, or Schmidt et al.'s (1979) linear trade-off. Meyer, Smith, and Wright (1982) suggest that although the impulse variability model (Schmidt, 1979) is intuitively appealing, its development was seriously flawed. Moreover, they showed that the linear speed-accuracy trade-off does not hold for conditions similar to that under which it was developed. Meyer et al.'s (1982) principal objection was that the model was developed solely on the initial ballistic phase of the movement and therefore neglected current control and the effects of antagonist muscles known to decelerate the motion towards its completion. Also, Schmidt et al. (1979) had made gross assumptions in manipulating the model's random variables. Meyer et al. (1982) suggested that the Fitts' law and impulse variability models, although differing in their task requirements, were really evaluating the same motor process. In an attempt to tie the two movement paradigms together, Meyer et al. (1982) offered the symmetric impulse variability model. This model was able to unify the linear and logarithmic speed-accuracy trade-off theories.

2.4.3.2 Force Variability The impulse variability model (Schmidt et al., 1979) further states that the sources of exertion variability (σ) are in the magnitude and duration of the force impulse used to accomplish the exertion goal, and that these parameters follow Weber's Law. More specifically:

$$\sigma_{force} \propto \text{force} \qquad (2.7),$$

$$\sigma_{duration} \propto \text{duration} \qquad (2.8),$$

With respect to isometric tracking, this means that if more force is required track, proportionately more variability (error) should be seen in tracking performance.

The idea that the control of force was determined by the amount of force exerted was not new. Early work by Bahrick, Bennett, and Fitts (1955) proposed the impulse ratio model which states:

$$\text{Error} \propto k\left(\frac{\delta F}{F(\delta F)}\right)^{-1} \qquad (2.9),$$

where F is the target force required, δF is the change in force (from some preload) needed to reach F, and k is a scaling coefficient. This means that errors in positioning should increase as the proportion of change in force to target force decreases. Therefore, a 1 cm change in position will be more accurate if the force required is from 9 to 10 N versus from 19 to 20 N. Recent studies concerning isometric force variability (Carlton and Newell, 1988; Newell and Carlton, 1988; Newell and Carlton, 1985; Sherwood and Schmidt, 1980) show that although force variability may be a function of force, it is not necessarily proportional to force.

Newell and Carlton (1988), in an evaluation of the impulse ratio and impulse variability models, studied the effects of force, preload, and rate of force production on the force variability of isometric exertions. Subjects were required to rapidly exert force against a isometric handle to match a criterion force level presented on an oscilloscope. Knowledge of results was given only after each trial was complete in order to eliminate the use of concurrent visual feedback during the exertion. Interpretation of these results towards the understanding of tracking is therefore limited.

Newell and Carlton's (1988) results show that the change in force level – not criterion force – had the greatest impact on force variability and that time to peak force (rate of force production) and preload level also showed significant effects. Increases in preload, change in force level from preload, and decreases in time to criterion force all increased force variability. The effects of time to peak force supported their earlier work (Newell and Carlton, 1985). Together, these results indicate that neither the impulse ratio nor the impulse variability model is adequate in predicting isometric force variability. Newell and Carlton presented a new equation for predicting force variability:

$$\sigma(F) \propto \frac{\delta F}{T^{1/2}} \qquad (2.10)$$

where σ(F) is the standard deviation in force and T is the time to reach the criterion force. With respect to tracking, the work of Newell and Carlton (1988) indicates that increases of change in required force and speed of tracking will increase the variability about the observed force. This result was shown directly by Berkowitz (1990). However, Kim, Carlton, and Newell (1990) found that the time to criterion force effect is not as pronounced as predicted in Equation 2.8.

The idea that more error is associated with faster, more forceful movements has been supported empirically (Berkowitz, 1990; Schopper, 1987; Schmidt et al., 1979). However, the presence of an optimal zone for control gain (Berkowitz, 1990) indicates that the proportional effect of required force on force variability is not uniform across the entire range of individual force production. More specifically, Berkowitz (1990) indicated that the proportional variability in force was increased at lower and higher force levels. In contrast, Sherwood and Schmidt (1980) showed that proportional variability in force during isometric exertions increases at low to median levels as predicted by impulse variability but decreases as the criterion force approaches maximum voluntary contraction. Newell and Carlton (1985) also found that force variability was proportionately decreasing at near maximal force values. This is in contrast to the results of Berkowitz (1990) but it must be kept in mind that the difference may be due to nature of the task (ballistic vs closed-loop tracking). It is also very important to note that Berkowitz scaled all force values to the individual's maximum capabilities before comparisons were made. This normalization allows subject data to be directly compared without the additional problem of within-subject variability. Although this was done in the Newell and Carlton (1985) study, it was not done for the research of Sherwood and Schmidt (1980) who indicated a greater disproportionality at near maximal force levels.

Newell and Carlton (1985) found that the attained maximum force was a function of the time to peak force for an isometric rate matching task. Maximum force was greater when more time was allowed to achieve that force. With regards to the tracking study, this indicates that the speed of tracking may influence the error at near maximal values of the track.

2.4.3.3 Timing Variability. Several researchers have also investigated the effects of movement parameters on the variable error in exertion timing (VE_t). Newell and Carlton (1988, isometric exertions) and Schmidt (1988, dynamic movements) report increase in timing error with increasing movement time to reach a target. Carlton and Newell (1988) showed an increase in timing error with increasing preload and decreasing velocity for a simple dynamic movement. Schmidt (1988) ascribes these results to the human limitations in estimating longer periods of time. Schmidt further suggests that variability in timing error is proportional to movement time:

$$VE_t \propto MT \tag{2.11}$$

Relating this to tracking performance, more error should be seen at slower tracking speeds as the ability of the subject to evaluate and predict target trajectory is decreased. However, this cannot be the case over all tracking speeds due to the psychological and physiological refractory limitation of the human (Poulton, 1981).

2.5 Co-contraction

Although the parameters used in describing force and timing variability help predict what can be expected in manual performance, there is still little explanation of the strategies or mechanisms that contribute to those sources of inaccuracy. For many years researchers have studied the muscular coordination involved in human movements. Much of this work has involved the understanding of synergistic and antagonistic muscle coordination in performing movement tasks. To date, the purpose of simultaneous agonist and antagonist co-contraction is a topic of wide debate.

2.5.1 Theories for the Purpose of Co-contraction

Since the early work of Sherrington (1905, as cited by Smith, 1981; Tyler and Hutton, 1985; Woldstad et al. 1988), the theory of reciprocal inhibition has been widely accepted as the means of explaining coordinated muscle activity. Reciprocal inhibition suggests that contraction of a primary mover in an agonist/antagonist muscle pair results in complete inhibition of the associated antagonist muscle. Many biomechanists have utilized the concept of reciprocal inhibition in order to simplify the modeling of complex muscle/joint structures in the human. However, much of the current research indicates that there is a significant degree of co-activity in the muscles during human movement. Researchers have found muscle co-contraction in a variety of dynamic exertions (Marsden et al., 1983; Redfern, 1988; Smith, 1981; Woldstad, 1989) as well as in isometric efforts (DeLuca and Mambrito, 1987; Ghez and Gordon, 1987; Redfern, 1988; Tyler and Hutton, 1986; Woldstad et al., 1988).

Smith (1981), in an effort to close the gap between researchers in the reciprocal and co-active camps, discussed circumstances based on previous literature when either reciprocal or coactive control could be expected. Conditions favoring reciprocal inhibition included rhythmic tasks, low velocity displacements, and exertions where movement is prevented by external resistance. Conversely, conditions favoring co-activation included high precision tasks, high velocity movements, and isometric exertions which require joint stability.

DeLuca and Mambrito (1987) also observed that the agonist/antagonist muscle pair were used in two control modalities: proportional activation and reciprocal activation, during an isometric tracking task using the muscles of the thumb. Their findings show that co-activation occurs under conditions of task uncertainty or when fast compensatory force was required (such as force reversal or deceleration). Reciprocal activation was found when both task uncertainty was present and fast compensatory actions were required.

DeLuca and Mambrito (1987) clarified that during tracking with some uncertainty in the track, force reversal was facilitated by a proportional shift in agonist/antagonist activity rather than a complete relaxation of the agonist, which would bring about the quickest force reversal. Therefore, during a sinusoidal tracking task, the co-activation of the agonist/antagonist pairs would "teeter-totter" about some co-activation level at the points of force reversal. The degree of the shift would be proportional to the uncertainty of the tracking task and possibly to the force required at the point of reversal. In the extreme case of rapid, random tracking there would be no median level of co-contraction and a complete silence of agonist activity at points of force reversal; this would be reciprocal control. Therefore, DeLuca and Mambrito (1987) show evidence that the central nervous system may control the motoneuron pools of the agonist/antagonist muscle pair as if it were one pool performing the same task.

As some researchers have found co-contraction, others have tried to determine the reason for this seemingly counter-productive activity. In studies involving isometric exertions, co-contraction has been attributed to the need for joint stability (Woldstad et al., 1988), to reverse the rate of force build-up (DeLuca and Mambrito, 1987; Ghez and Gordon, 1987), or to facilitate relaxation of the agonist when high rates of relaxation are required (Redfern, 1988).

Studies have found that the antagonist muscle is essential to the deceleration of movement as part of a triphasic pattern of muscle activation

(DeLuca and Mambrito, 1987; Ghez and Gordon, 1987; Hallett, Shahani, and Young, 1975; Marsden et al., 1983; Woldstad, 1989). The triphasic pattern, as measured using EMG, is characterized by an impulse from the agonist to start the movement, followed by an antagonist burst to decelerate the movement, and a second agonist burst to complete the movement.

2.5.2 Characteristics of the Triphasic Pattern

Hallett et al. (1975) studied rapid and smooth dynamic flexion movements about the elbow. Results indicated that the antagonist burst was centrally preprogrammed rather than currently controlled for rapid movements. Smooth (ramp) movements showed no triphasic pattern. However, Marsden et al. (1983) and Angel (1977 contend that the antagonist burst is under the control of propriorecptive influences as the antagonist burst was effected by perturbations to the limb while moving.

Studies have shown that the magnitude of the primary agonist and antagonist bursts are dependent on movement parameters such as velocity and distance much like impulse variability (see Section 2.4). Marsden et al. (1983) show a linear relationship between the magnitude of the antagonist burst and the peak velocity of the movement. The slope of this antagonist-peak velocity relation changes as a function of movement distance, with longer movements requiring less antagonist activity. Marsden et al. (1983) interpret the complexity of these interactions between antagonist activity and movement parameters as an argument against the theory of central programming of the antagonist burst.

2.6 EMG and Isometric Force Production

Electromyography (EMG) is a useful tool in measuring muscle activity. EMG can detect the electro-chemical depolarization of the muscle fibers as they contract (Section 2.4.2). Since this electrical activity is caused by the neural impulse sent from the central nervous system, EMG measures how much muscle force is being requested by the CNS. Therefore, when applied correctly, EMG data can reveal much about the way in which the central nervous system coordinates and controls muscle force production.

2.6.1 EMG and Isometric Muscle Tension

Researchers have tried to define the relationship between EMG activity and isometric muscle force. Even today, the nature of the EMG-force relation is not resolved.

Lippold (1952) was one of the first to study isometric force using integrated, surface EMG. He showed a linear relation between EMG and isometric muscle force for the gastrocnemius and soleus muscles of the ankle during plantar flexion. The correlation coefficient for these linear relations was between 0.93 and 0.99. This linear relation has been widely supported (Hof and Van den Burg, 1977; Moritani and DeVries, 1978; Woldstad, 1989; Woldstad et al., 1988; Woods and Bigland-Ritchie, 1983). However, there is much support for the finding that the force-EMG relation is quasi-linear or quadratic (Komi and Viitasalo, 1976; Lawrence and DeLuca, 1983; Zuniga and Simons, 1969) with an increase in the EMG per unit force as force increases.

Lawrence and DeLuca (1983) provide a comprehensive study involving muscles varying in size and fiber composition in several isometric tasks.

Their results show that the force-EMG relation is widely variable but primarily dependent on the muscle type. The relationship was found to be independent of the skill level of the group and the rate of force production. Lawrence and DeLuca (1983) offer some possible explanations for the variability observed in the force-EMG relation:

- motor recruitment and firing rate properties of the muscle
- location and distribution of fibers within the muscle (spatial bias)
- cross talk from myoelectric signals in other muscles
- agonist/antagonist muscle interaction
- viscoelastic properties of the muscle.

<u>2.6.1.1 Effects of Motor Unit Control Characteristics</u> Lawrence and DeLuca (1983) found that larger muscles which regulate force primarily by recruitment (biceps, deltoid) exhibited non-linear force-EMG relations. These muscles are often used for coarse control. Conversely, smaller muscles which regulate force primarily by firing rate (first dorsal interosseous of the thumb) exhibited a more linear force-EMG relations. These smaller muscles are often used for fine control. These results were essentially duplicated for isometric exertions of the soleus and adductor pollicis (linear), and the biceps and triceps (non-linear) by Woods and Bigland-Ritchie (1983).

Further evidence supports the linear force-EMG relation with firing rate (Basmajian and DeLuca, 1983; Lawrence and DeLuca, 1983; Milner-Brown and Stein, 1975; Woods and Bigland-Ritchie 1983). In fact, only one study reviewed (Moritani and DeVries, 1978) suggests that firing rate contributes non-linearly to near maximal isometric exertions.

Researchers have also shown that the larger muscles of the body may use recruitment up to 70%-80% of maximum voluntary contraction

(Basmajian and DeLuca, 1983). Milner-Brown and Stein (1975) and Woods and Bigland-Ritchie (1983) offer substantial support that the relation between recruitment and force is highly non-linear. This is .eflected in the findings that large muscles exhibit non-linear force-EMG relationships (Lawrence and DeLuca, 1983; Woods and Bigland-Ritchie, 1983).

2.6.1.2 Effects of Fiber Type Composition of Muscle. Woods and Bigland-Ritchie (1983) further state that the fiber type composition within a muscle greatly influences the force-EMG relation. They found a linear relationship for fast twitch muscle fibers, having higher force thresholds, and a non-linear relationship for slow twitch fibers, recruited at lower force thresholds.

2.6.1.3 Effects of Synergy and Co-contraction. The linear relation found in the first dorsal interosseous (Lawrence and DeLuca, 1983) may be greatly influenced by the changing co-active activity in the antagonist muscles of the hand as found in isometric prehension (Smith, 1981). This co-contraction leads to a net force response which is not representative of the individual force within each muscle. Therefore, the muscle forces being produced by the agonist is underestimated due to the counter-action of the antagonist muscle group(s) (Woldstad et al., 1988). Similarly, it is possible that synergistic muscles could contribute non-proportionately at differing levels of overall agonist tension resulting in a non-linear representation of single muscle EMG activity to group muscle force (Hof and van den Berg., 1977).

There remains some inconsistencies in the data available on the nature of the force-EMG relationship. In addition to the possible sources for the inconsistency presented by Lawrence and DeLuca (1983), Redfern (1988) adds that instability of the myoelectric signal, and differences in the EMG measurement system and analysis methods (Siegler, Hillstrom, Freedman, and Moskowitz 1985) contribute to variation in results. Siegler et al. (1985) also suggest that the force-EMG relationship is greatly influenced by within-subject variability, and by the spatial bias of surface electrode placement. However, even with these sources of experimental error, some useful information can be extracted from the existing literature. Taken together, these results indicate that muscles with mixed fiber type using a combination of recruitment and rate coding to regulate force should exhibit a non-linear force-EMG relation (even if only slight). The results of Woldstad (1989) indicate that although the use of a quadratic equation better predicts the force-EMG relation for the biceps and triceps muscles, the benefit of using the polynomial versus a linear equation was minimal.

2.6.2 Models Relating EMG to Isometric Muscle Tension

Along with the attempts to determine the force-EMG relationship, several researchers have been developing models to predict muscle force by using EMG data as the model input. These models go beyond the simple linear or quadratic models of Section 2.6.1 to include antagonist co-contraction, movement velocity, and length tension relationships. These models are usually of an empirical or mechanical nature. The present study is only concerned with those models predicting isometric force.

2.6.2.1 Empirical Models. Woldstad et al. (1989) developed a model to study muscle co-contraction during a sinusoidal replication task. Subjects were to reproduce a slow sinusoid motion of +/- 98 N at 0.33 Hz. They utilized a simple linear regression model containing linear components for both the agonist and antagonist muscle force. Net force was modeled using integrated EMG (iEMG) such that

$$\text{Net Force}(t) = B_1{*}iEMG_B(t-\tau) + B_2{*}iEMG_T(t-\tau) \quad B_1>0;\; B_2<0 \quad (2.12).$$

The coefficients B_1, and B_2 are found using a least squares regression against observed muscle force and $iEMG_B$, $iEMG_T$ are the average, integrated values for the biceps and triceps, respectively, at time t. The value τ represents a neuromechanical, pure time delay between muscle EMG activity and the onset of force production (Woldstad, 1989). This delay needs to be incorporated into the model as there is often a considerable time delay between the detection of EMG activity and the onset of force production. The value of τ can be determined by shifting the iEMG forward in time until the maximum least-squares fit is found. Woldstad et al.'s results showed that this model effectively predicted force output ($\underline{R}^2 = 0.95$). Further, the coefficients of the regression B_1, and B_2 were orthogonal and directly comparable as they contributed in opposite directions to the force output. With the orthogonal coefficients and the measured iEMG, measures of the relative antagonistic muscle activity as well as ratios of co-contraction were developed to determine co-active muscle behavior during the exertion. Co-contraction at the elbow was found to be greater for isometric extension than for isometric flexion.

Hof and van den Berg (1977) and Redfern (1988) have used a similar, linear component models using the soleus, triceps surea, and the gastrocnemius muscles to predict ankle torque. By using one pair of synergistic muscles, orthogonal coefficients could not be obtained either by proportionality or least squares regression. However, using known posture and length/tension effects on muscle tension, each muscle could be isolated to determine its force-iEMG relation thus orthogonal coefficients were obtained.

2.6.2.2 Mechanical Models. The second class of models used to describe muscle behavior are mechanically based models. These models use mathematical representations of the passive and active components of the muscle to develop a differential equation based on EMG input. Gottlieb and Agarwal (1971) used a transfer function of output force over input EMG to evaluate a second order differential equation describing the muscle. The model used was:

$$\frac{\text{Force}}{\text{EMG}} = \text{K} \cdot \text{G}_1(\text{S}) \cdot \text{G}_2(\text{S}) \tag{2.13}$$

where:

$$\text{G}_i(\text{S}) = \frac{1}{\text{T}_i\text{S} + 1} \tag{2.14}$$

$G_2(S)$ is the Laplace transform of the transfer function of the linearized mechanical portion of the passive muscle, and $G_1(S)$ describes the active state of the muscle. The constant K is the gain factor and T_i is the time delay of the respective component. Gottlieb and Agarwal's results show that the second-order differential equation described the isometric step and ramp tracking task involving the soleus and anterior tibial ankle muscles to a fair degree. Unfortunately, random "tweaking" of the time parameters was necessary to obtain the rather average results.

Similarly, Coggshall and Bekey (1970) used a second-order differential equation to model the force output for the triceps during an isometric task. The equation used was:

$$\frac{d^2 F_h(t)}{dt^2} + a_1 \frac{dF_h(t)}{dt} + a_2 F_h(t) = KV_r(t) \qquad (2.15)$$

where F_h is the modeled force, $V_r(t)$ is the full-wave rectified EMG, and a_1, a_2 are constants found using the method of steepest descent. Results showed that the model predicted the triceps force well except during intervals of rapid force change. It was suggested that the increased error was due to short term parameter variation, or a non-linear, differential force-EMG relationship (Coggshall and Bekey, 1970). Antagonist co-contraction from the biceps was not taken into account. Further, it has been established that isometric triceps extension is often accompanied by significant levels of biceps co-activation (Woldstad, 1989; Woldstad et al., 1988).

2.7 Summary

The preceding background review demonstrates several points which can be summarized as follows:

- Manual control models of the human neuromuscular system are largely descriptive and do not quantify the neuromuscular strategies involved in the initiation and execution of manual control movements. Moreover, these models only represent the net result of agonist/antagonist muscle activity and not the coordinated activity of these muscles.

- Isometric, pursuit, sine wave tracking, the simplest form of a closed-loop manual task, provides an opportunity to investigate coordinated muscle activity during manual control.

- Berkowitz (1990) investigated the effects of required force and target speed on the performance of an isometric, pursuit tracking task. He demonstrated, empirically and subjectively, that an optimal zone for the isometric control/response ratio exists when tracking performance was measured proportional to the C/R used. The corresponding optimal force level was found to be approximately 65% of extension MVC. Other studies (Hess, 1974; McRuer and Jex, 1967) have also demonstrated a subjective optimal zone for the control/response ratio.

- Variability in the control of force production increases with increased movement velocity, increased force, and increased rate of force production.

- Woldstad et al. (1988) showed that co-contraction exist for an isometric sine wave replication task. This co-activity was greater for extension than for flexion.

- DeLuca and Mambrito (1987) showed that co-active muscle behavior is present in conditions of task uncertainty or force reversal. This co-activity may shift proportionately at points of force reversal.

- The force-EMG relation can be of a linear or non-linear form depending on the fiber composition and the recruitment and rate coding characteristics of the muscle.

- Empirical models have been developed which can accurately describe the nature of agonist and antagonist activity during isometric exertions.

3. EXPERIMENTAL METHOD

3.1 Overview

The purpose of this research was to study the relationship between muscle coordination and the performance of a simple manual tracking task. The experimental methods were a combination of the techniques used by Berkowitz (1990) to study preferred force levels during isometric tracking and Woldstad (1989) to study muscle co-contraction during isometric elbow exertions. This experiment studied the isometric exertions of subjects performing several tracking tasks using stationary wrist cuff equipped with a force sensing transducer. The experimental task was the digital equivalent of a pursuit tracking task with a laterally translating, sinusoidal forcing function. Force measurement and electromyography (EMG) were used to investigate the effects of C/R ratio and target velocity on muscle co-activation and tracking error during an isometric, pursuit tracking task. EMG modeling techniques, based on the work of Redfern (1988) and Woldstad (1989), were used to quantify the force contributions of the elbow flexor and extensor muscles to the observed tracking force.

3.2 Experimental Apparatus

The study used an experimental apparatus consisting of an EMG measurement system, a force measurement system, and a closed-loop, tracking display system (Figure 3.1). The acquisition and storage of force and EMG data was facilitated by a programmable, 16 channel AtoD converter connected to an IBM PS/2 microcomputer. VGA color displays were used by

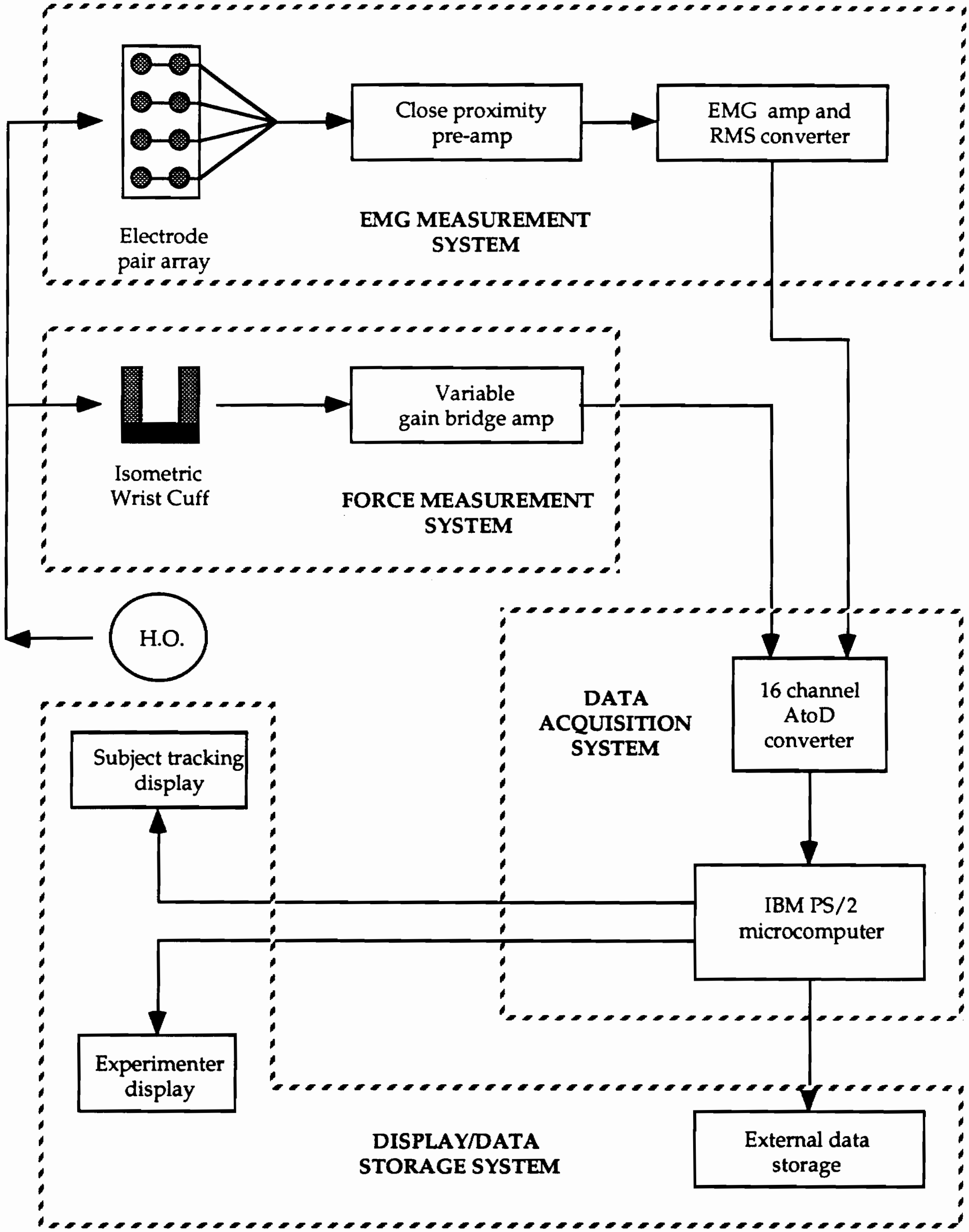


Figure 3.1 Experimental apparatus.

the subject to perform the pursuit tracking task and by the experimenter to monitor subject and equipment performance. A posture bench equipped with an adjustable arm support was used to control subject posture throughout the experiment.

3.2.1 EMG Measurement System.

Muscle activity was measured using a bipolar, surface EMG system with 8 mm diameter, Ag-AgCl electrodes. Multiple electrode pairs were used on each muscle to reduce the spatial bias (noise) induced by electrode placement (Redfern, 1988; Woldstad, 1989). Redfern (1988) showed that a significant reduction in EMG signal variability can be achieved using the multiple electrode pair technique. The electrode pairs were arranged in four sets running parallel to the muscle fiber. Each electrode pair was spaced 1.5 cm apart (Basmajian and DeLuca, 1985) and set into flexible rubber templates which were attached to the skin using double sided, adhesive tape. A four channel pre-amplifier (Analog Devices AD624A, 200x) was then attached directly onto each electrode pair array via a connector, also mounted to the template. The close proximity of the pre-amp to the electrode pairs reduced the amount of electrical noise induced into the system by external factors in an attempt to maximize the signal-to-noise ratio of the EMG signal. Each pre-amplifier was calibrated prior to each subject's run by sampling its output at zero and 1.0 mV (RMS) . The pre-amplified EMG signals were sent to a variable gain amplifier (Burr-Brown INA102KP, 1-1000x) and subsequently to a root-mean squared (RMS) estimator (Analog Devices AD536A). The RMS estimator used a small time constant of 25 msec which will allowed for some reduction in the noise of the EMG (smoothing) while retaining the critical information contained in the signal (Redfern, 1988). The computer sampled the EMG output at 100 Hz using the 16 channel analog-to-digital converter.

One concern was that the close proximity of the muscles to each other and the increased surface area covered by the electrode pair arrays would increase the chances of cross-talk between the EMG pairs located on the muscle systems. This cross-talk would make it impossible to isolate the individual activity of each muscle. Test data taken prior to subject testing revealed that there was little correlation, $\underline{r} = 0.037$, between the biceps and triceps EMGs during the performance of high force, high velocity, isometric flexion and extension exertions while in the experimental apparatus. This type of exertion reduces correlated muscle activity in the biceps and triceps from co-contraction (legitimate correlation) and can better isolate the presence of cross-talk.

3.2.2 Isometric Force Measurement System

An adjustable wrist cuff was used to perform the pursuit tracking task. Pilot data revealed contributions from the pronators and supinators of the forearm while performing the tracking task with a standard isometric joystick. However, these unwanted force contributions were greatly reduced when a wrist cuff was used in place of the joystick. Therefore, using the wrist cuff isolated the force contributions of biceps and triceps while performing the tracking task. The wrist cuff was 10 cm high, 7 cm long, and had an adjustable range of 6 cm in width. The wrist cuff unit was attached to a 300 lb load cell (Lebow Products model 3397) which could detect and measure the lateral force applied at the cuff. Before each subject run, the force measurement system was calibrated by sampling the load cell output with no load and with a 100 N load applied at the wrist cuff. The voltage signal produced by the load cell was sent to a custom made, single-channel amplifier with fixed gain and stored in the microcomputer. The computer sampled the force output at 100 Hz using the 16 channel analog-to-digital converter.

3.2.3 Data Acquisition System.

The central element of the experimental apparatus was a IBM PS/2 model 50, microcomputer equipped with two IBM model 8513 VGA color displays. The displays were approximately 23 cm wide and 17 cm high. The computer housed a programmable 16 channel analog-to-digital (AtoD) converter (MetroByte model DASH-16F). This computer generated the pursuit tracking tasks, displayed subject performance, and collected EMG, force, and tracking error data simultaneously. The AtoD board was programmed to collect 9 channels of data (8 EMG, and 1 force) at 100 Hz. The AtoD board operated in bi-polar mode with the equipment's hardware and software gains adjusted to use as much of the -2047 to +2048 integer range as possible. The excitation voltage for the load cell was +/- 5.0 volts. With the adjustability in the system, a very high resolution (AtoD units/force) could be achieved, maximizing the sensitivity of the force and EMG measurements.

Computer software generated the closed-loop tracking task by presenting a computerized target that moved across the screen according to a preset, lateral, sinusoid forcing function. The subject tracked the moving target with a cursor by applying lateral force to the isometric wrist cuff in an attempt to minimize the distance between them. The input of force from the wrist cuff changed the position of the cursor according to the preset control/response (C/R) ratio. The computer continually updated the positions of the target and the cursor on the screen. One area of concern in the closed-loop tracking task was the control lag between the isometric force input and the cursor response on the screen. Excessive lag could alter the subject's motor performance (Woldstad, 1989). The control lag of the system was constrained by the VGA monitor refresh rate. This rate is approximately 60 Hz., resulting in a control lag of approximately 17 msec.

3.2.4 Posture Support Bench

A custom posture support system was used to allow exact positioning of the right upper limb during the tracking tasks. Specifically, the posture to be maintained for each trial was:

- 90° shoulder abduction
- 0° shoulder flexion in the frontal plane
- 90° included elbow angle
- forearm in semi-supinated (thumb up) position,

following the methods used by Berkowitz (1990) and Woldstad (1989). The forearm was maintained in a semi-supinated (thumb up) position. The limb support structure allowed for lateral, fore/aft, and height adjustment in order to locate the wrist cuff optimally for the right wrist of each subject. The wrist cuff was adjusted against the forearm so that there was no play during flexion and extension. The forearm was placed in the cuff such that the distal head of the radial bone was just ahead of the cuff. The subjects were instructed to use only their biceps and triceps muscles to perform the task. This minimized the contributions of other muscles in performing the isometric tracking task.

3.3 Subjects

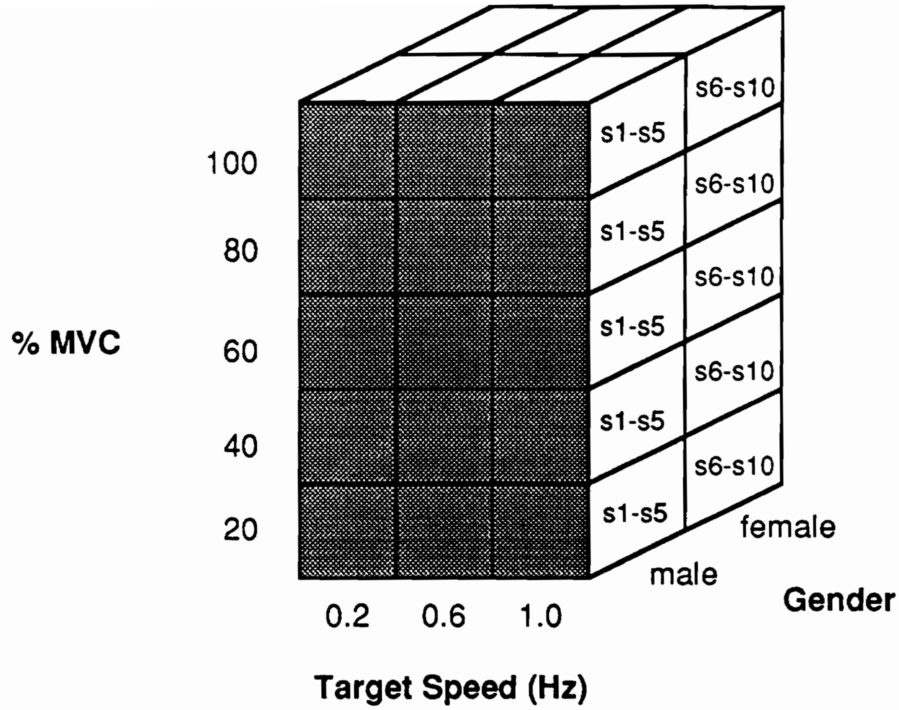
The subjects used for the experiment were 10 students (five male and five female) with no history of musculoskeletal disorder. Both males and females were used to provide a more accurate representation of the population involved in manual control performance. The subjects were recruited from the graduate student population in the Industrial and Systems Engineering Department at Virginia Tech. The study used only right-handed subjects due to the constraints of the posture support system described above. Table 3.1 shows the relevant statistics on the population group studied here. The experiment required one, two hour session for each subject. Pay was \$5.00 per hour for a total of \$10.00 per subject for the entire experiment.

3.4 Experimental Design

A mixed factors design was used with one between-subjects variable (gender) and two within-subjects variables (tracking force level and target frequency). The within-subjects parameters included five levels of required force (20%, 40%, 60%, 80%, 100% of maximum extension and flexion force) and three levels of target forcing function frequency (0.2, 0.6, 1.0 Hz). Figure 3.2 illustrates the experimental design. The changes in the level of required tracking force reflect changes in the C/R ratio for the task since the amplitude of the forcing function on the screen remained the same for all trials.

Table 3.1 Subject statistics.

Measure		Males	Females
Age (years)			
	Mean	26.80	25.40
	Std.	3.90	1.52
	Minimum	23	24
	Maximum	31	28
Height (cm)			
	Mean	185.00	166.00
	Std.	6.44	4.53
	Minimum	177	160
	Maximum	193	171
Weight (kg)			
	Mean	89.32	59.44
	Std.	9.84	6.65
	Minimum	77.40	49.00
	Maximum	104.70	67.40



Trial # \ S #	1		2		3		4		5		6		7		8		9		10	
1	* 1	1 2 3	2	2 1 3	3	3 1 2	4	2 3 1	5	1 2 3	4	2 1 3	5	3 1 2	1	2 1 3	2	3 1 2	3	1 2 3
4	2	3 1 2	3	3 1 2	4	3 1 2	5	2 1 3	1	3 1 2	3	1 2 3	4	2 3 1	5	3 2 1	1	1 2 3	2	3 2 1
7	5	3 1 2	1	1 2 3	2	1 2 3	3	2 1 3	4	1 2 3	5	2 1 3	1	1 2 3	2	2 3 1	3	1 2 3	4	1 3 2
10	3	3 2 1	4	1 3 2	5	3 2 1	1	1 2 3	2	2 1 3	2	3 1 2	3	2 3 1	4	1 3 2	5	2 1 3	1	3 1 2
13	4	2 3 1	5	1 3 2	1	3 2 1	2	3 2 1	3	3 2 1	1	2 1 3	2	1 3 2	3	2 3 1	4	2 3 1	5	2 1 3

* force level (balanced) → | 1 | 1 2 3 | ← frequency conditions (random)

Figure 3.2 Experimental design.

The order of conditions for each subject was partially balanced. The five levels of required tracking force followed a Latin square balancing scheme and the frequency levels were randomized within each force level (Figure 3.2). Force was used as the balancing variable because of the effects of subject fatigue on force production capability and EMG measurement. EMG amplitude will increase with fatigue while force capability will decrease (Chaffin and Andersson, 1991). By balancing the order of presentation for the force conditions, the effects of differential transfer and fatigue due to trial order was also minimized, reducing the effect of nuisance variables on the dependent measures.

The dependent measure used to evaluate tracking performance was the average absolute tracking error (N) as a proportion of the required force for that trial (PE, Equation 2.3 from Berkowitz, 1990). This measure of tracking error accounts for changes in the C/R ratio and removes the effects of within subject strength variability.

The dependent measures used to evaluate muscle co-contraction were absolute level of antagonist force (AAF) and the ratio of antagonist to agonist force during the trial (co-contraction ratio, CR). The antagonist muscle was defined as the muscle that works to oppose the direction of exertion. For example, in flexion the biceps is the primary mover while the tricep acts as the antagonist muscle. For extension, the situation is reversed.

From the EMG, the actual level of maximum muscle force (MMF) required to track was also determined for each tracking condition. The MMF for flexion was the maximum value of biceps force during the flexion phase of the track while the MMF for extension was the maximum force of the triceps during the extension phase of the track. The values for MMF were normalized with respect to the MMF observed at the 100% MVC, 0.2 Hz condition to get a measure of %MMF needed to track at each condition.

3.5 Experimental Procedures

Subjects were asked to perform both static, maximum strength trials and the tracking trials in a two hour experimental session.

3.5.1 Strength Measurement

In order to specify the individualized C/R ratios for the tracking task, each subject performed both maximum flexion and extension strength trials. Standardized maximum strength testing procedures were used (Chaffin, 1975). These procedures use a six second trial where the subjects gradually rise to their maximum force (within two seconds), and hold that maximum for three seconds. One second is added onto the five second period in order to insure no effects due to end of trial anticipation. The maximum force was averaged over seconds two to five. For a strength trial to be acceptable, the subject had to maintain a +/- 15% confidence interval about the mean value between seconds two to five of the trial (Chaffin, 1975).

After giving the "ready, set, go" command, the experimenter monitored the force output on the CRT. If there were any deviations from the criteria above, the trial was discarded, the subject rested, and the trial was repeated. Three maximum flexion and extension strength trials were averaged to get an accurate measure of flexion and extension maximum voluntary contractions (MVC) for each subject. The data collected for the maximum strength trials included only force output.

Table 3.2 shows the maximum flexion and extension strengths for males and females as well as for each subject. These data are consistent with that reported by Chaffin and Andersson (1991). However, the strengths observed here were higher than reported by Berkowitz (1990) who tested subjects under very similar conditions. The reason for this disparity is unknown.

A t-test for sample means showed significant differences in flexion and extension strength for the two groups studied here ($\alpha = 0.05$, $n = 5$). However, due to the small sample size, this result should not be generalized as a difference between male and female strength capability.

3.5.2 The Tracking Task

The main experimental trials consisted of a one dimensional, zero order (position), pursuit tracking task with zero lag using a wrist cuff. The task was similar to that used by Berkowitz (1990) and Burke and Gibbs (1965). Subjects tracked a hollow green circle (10 pixels in diameter) on a black background with a yellow, cross-shaped cursor (10 pixels in each arm). The objective of the task was to keep the cross within the circle at all times. The target translated laterally according to a sine function with a preset frequency for the trial condition. The peak to peak amplitude of the forcing function was 10.8 cm on the screen.

The control/response ratio for the control was determined by the amplitude of the forcing function and the required force output (%MVC) for the trial. The amplitude of the sine wave forcing function was always the

Table 3.2 Maximum voluntary contraction strengths (MVC, N).

Subject	Gender	Flexion MVC (N)	Extension MVC (N)
1	M	190.56	165.55
2	M	347.18	245.18
3	M	305.96	202.8
4	M	260.45	175.05
5	M	252.35	173.63
6	F	84.38	78.89
7	F	148.42	134.92
8	F	167.64	139.02
9	F	130.3	110.7
10	F	197.39	130.23
Male			
	Mean	271.30	192.44
	Std.	59.07	32.67
	Minimum	190.56	165.55
	Maximum	347.18	245.18
Female			
	Mean	145.63	118.75
	Std.	42.30	24.79
	Minimum	84.38	78.89
	Maximum	197.39	139.02

same. Modifications to the C/R ratio involved changes in the force required to reach the peak amplitudes for that trial. Since the task involved flexion and extension exertions, the gain setting in the flexion direction of the track was determined by the percentage of maximum flexion strength ($\%MVC_f$) and the gain in the extension direction was determined by the maximum extension strength ($\%MVC_e$). There was a change in the C/R setting everytime the controlled element crossed the zero point in the center of the screen (moving from flexion to extension tracking). The purpose of changing the C/R ratio during the task was to ensure that the subjects were required to track at the same relative level of difficulty at every tracking condition. Therefore, each subject was required to track at a percentage of his/her strength capability. This technique removed the effects of control gain and subject strength capability from the tracking and muscle data, allowing direct analysis of subject tracking behavior.

As the trials were partially balanced, the subjects did not know which condition was to be performed prior to the trial. The software generated preparation cues ("prepare to move cursor" and "move cursor at signal") on the screen as the trial was about to begin. At an auditory tone the target moved and the subject began to track. Data was collected for ten seconds after a five second practice tracking period had been performed. These time values were chosen to optimize the trade-off between fatigue and preprogramming of the task. Over time, subjects will become fatigued (Kroemer, Kroemer, and Kroemer-Elbert, 1986) and they will also start to track in an open-loop fashion (Frost, 1972). As this research is attempting to observe normal muscle activity during closed-loop tracking, the time factor was controlled.

The data collected for the tracking trials included the forcing function trajectory, the force output (tracking trajectory), and the eight channels of integrated EMG. The force and iEMG data were collected at the computer in analog to digital (AtoD) units.

3.6 Experimental Protocol

Each subject participated for a two hour session. One session was used due to the difficulties in obtaining reproducible results from multiple applications of the EMG measurement system (Basmajian and DeLuca, 1985). Using EMG, small changes in electrode placement may lead to very different results due to cross-talk from synergistic muscles (Redfern, 1988). Also, the number and length of trials was sufficiently small to avoid data artifacts from muscle fatigue.

Upon arriving at the Industrial Ergonomics Laboratory, each subject was asked to read a set of written instructions and to fill out an informed consent form (Appendix A). Once informed consent was obtained, the experimenter reviewed the procedures and answered any questions that the subject might have. The experimenter then prepared the biceps and tricep muscles of the right arm for application of the surface electrodes. This preparation included removing any hair that may interfere with the EMG attachment and rubbing the skin with a cloth soaked in alcohol to remove any dead skin cells which would increase the skin tissue resistance. The EMG cavities were also filled with an electrode gel to reduce this resistance. Once the EMGs were in place, the subject was seated in the experimental chair. The chair was adjusted to fit the anthropometric requirements of the subject.

Approximately 15 to 20 minutes were taken to allow saturation of the electrode gel into the skin before taking any EMG measurements. This was to avoid artificial increase in the EMG signal amplitude due to reduction in the electrical resistance of the skin over the course of the experiment.

The subject performed the three maximum flexion and extension strength trials. There was a three minute rest period in-between each trial (Chaffin, 1975). These trials were averaged to obtain the maximum values used in the calculation of the C/R ratios for the following tracking trials. The 15 experimental tracking combinations were then performed, each with the specific Force and Frequency parameters. A three minute rest was given in-between each of these trials as well.

After the trial session, the subject was debriefed and paid. The session took approximately two hours.

4. DATA ANALYSIS AND RESULTS

4.1 Overview

ANOVA was used to determine the effects of Force and Frequency on the dependent measures: proportional tracking error (PE), absolute antagonist force (AAF), and antagonist/agonist co-contraction ratio (CR). The p-values reported here are corrected for the maximum heterogeneity of co-variance among repeated measures using the Greenhouse-Geisser (G-G) correction factor. This correction is conservative and will negatively bias the F-test. Orthogonal polynomial contrasts were used to determine the functional nature of the dependent measures across the Force and Frequency tracking parameters. Subsequently, least squares models were fit to the tracking error and antagonism data using the tracking parameters as model inputs. If these models were quadratic, the value of the optimum was determined by partial differentiation of the model with respect to the required Force parameter. Appendix B contains a summary description of the programs developed or used to generate, compile, and analyze the EMG/tracking study data.

In order to better understand the study's results, a brief description of the tracking task and its characteristics is needed. Figure 4.1 provides an example of a single tracking period in a typical sinusoidal forcing function. It should be noted that each period of the forcing function has a flexion phase and an extension phase. The phases have opposite signs with respect to force output (flexion is positive force, extension is negative force). The peaks of these phases (denoted by the arrows) are the maximum required tracking

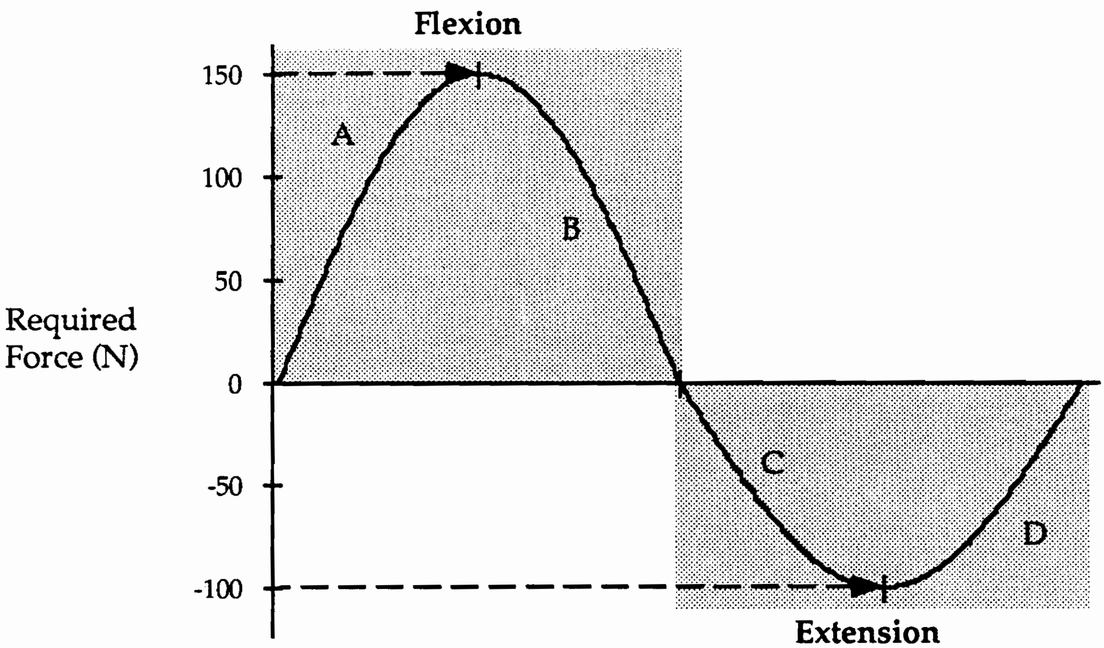


Figure 4.1 Characteristics of the track forcing function.

forces for this sample tracking task. During the flexion phase of the track, the biceps are the primary agonist muscles while the triceps serve as the antagonist group. Conversely, the triceps are the primary agonist for the extension phase of the track while the biceps act as the antagonist muscle group.

Each phase of the track requires two different types of physical effort: exertion and release. For example, during the flexion phase part 'A' represents flexion exertion (increasing effort) while part 'B' represents flexion release (decreasing effort). Parts 'C' and 'D' represent extension exertion and release respectively.

4.2 Tracking Error

Tracking error was calculated from the actual force output (track) and the target forcing function of the trial. The error measure used was Proportional Tracking Error (PE) expressed as the average absolute force error between the forcing function and the tracking response, divided by the corresponding maximum flexion or extension values of the forcing function required for that trial:

$$PE_f = \frac{\sum \frac{\text{Abs(forcing function(N) - actual track(N))}}{\text{Required flexion force for trial: }\%MVC_f}}{\text{Number of observations}} \tag{4.1}$$

$$PE_e = \frac{\sum \frac{\text{Abs(forcing function(N) - actual track(N))}}{\text{Required extension force for trial: }\%MVC_e}}{\text{Number of observations}} \tag{4.2}$$

Therefore,

$$PE = \frac{(PE_f + PE_e)}{2} \qquad (4.3),$$

represents the overall error associated with a given tracking condition. This measure was found by Berkowitz(1990) to be the most meaningful measure of tracking error because it is sensitive to alterations in the force required to track the target (C/R ratio).

4.2.1 Overall Track

Tracking error was analyzed in a mixed-factor ANOVA against the independent variables: Force, and Frequency. Gender was included as a between-subjects blocking variable. The ANOVA summary table is shown in Table 4.1. Significant main effects for PE were found for target speed (Frequency, $\underline{p} < 0.0001$). The Force effect was not significant with the G-G correction factor ($\underline{p} < 0.0817$) but was significant without the correction for heterogeneity of variance ($\underline{p} < 0.0441$).

Orthogonal polynomial contrast were used to determine the nature of the main effect of Frequency and the suspect effect of Force. The contrasts revealed a significant first order fit for the Frequency effect ($\underline{p} < 0.0001$) as well as a significant second order fit for Force level ($\underline{p} < 0.0101$). These results are consistent with those found by Berkowitz (1990).

Polynomial regression was used to model PE as a function of the Force and Frequency cell means. The model was formulated by using the results of

Table 4.1 Overall track ANOVA summary table for Proportional tracking Error (PE).

ANOVA model for PE						Type III SS
Source	df	SS	MS	F-Value	P-Value	G-G
Gender	1	0.024	0.024	0.917	0.3663	
S(Gender)	8	0.213	0.027			
Frequency	2	0.406	0.203	61.097	0.0001	0.0001
Frequency * Gender	2	0.003	0.002	0.454	0.6429	0.5636
Frequency * S(Gender)	16	0.053	0.003			
Force	4	0.026	0.006	2.767	0.0441	0.0817
Force * Gender	4	0.004	0.001	0.465	0.7607	0.6644
Force * S(Gender)	32	0.074	0.002			
Frequency * Force	8	0.012	0.002	1.029	0.424	0.409
Frequency * Force * Gender	8	0.007	0.001	0.6	0.7747	0.6733
Frequency * Force * S(Gender)	64	0.096	0.002			

the ANOVA and polynomial contrast analyses reported above. The regression model was:

$$PE = B_1 + B_2\text{Frequency} + B_3\text{Force} + B_4\text{Force}^2 \qquad (4.4).$$

The model was significant at $p < 0.0001$ and accurately predicts tracking error ($R^2 = 0.97$). Figure 4.2 shows a plot of the Frequency by Force interaction means and the model prediction values. This plot clearly shows the increasing linear trend in tracking error with increasing Frequency (faster tracking) as well as the increasing error at higher and lower required tracking Force levels. Partial differential analysis of Equation 4.4 with respect to Force ($\partial PE/\partial Force$) indicates that overall minimum tracking error occurs at a Force level of approximately 61% MVC. This is consistent with the findings of Berkowitz (1990) which revealed an optimal tracking at approximately 66% of extension MVC.

4.2.2 Part Task Analysis

Each part ('A - 'D') of the track was analyzed separately to determine if different tracking error relationships existed for flexion and extension exertion and release portions of the track. Appendix C shows the expanded ANOVA summary table for PE by phase (flexion or extension) and by part (exertion or release) of the tracking task. PE was found to be greater for extension than flexion ($p < 0.003$). No differences were found between the exertion and release parts of each phase for PE ($p < 0.449$). Table 4.2

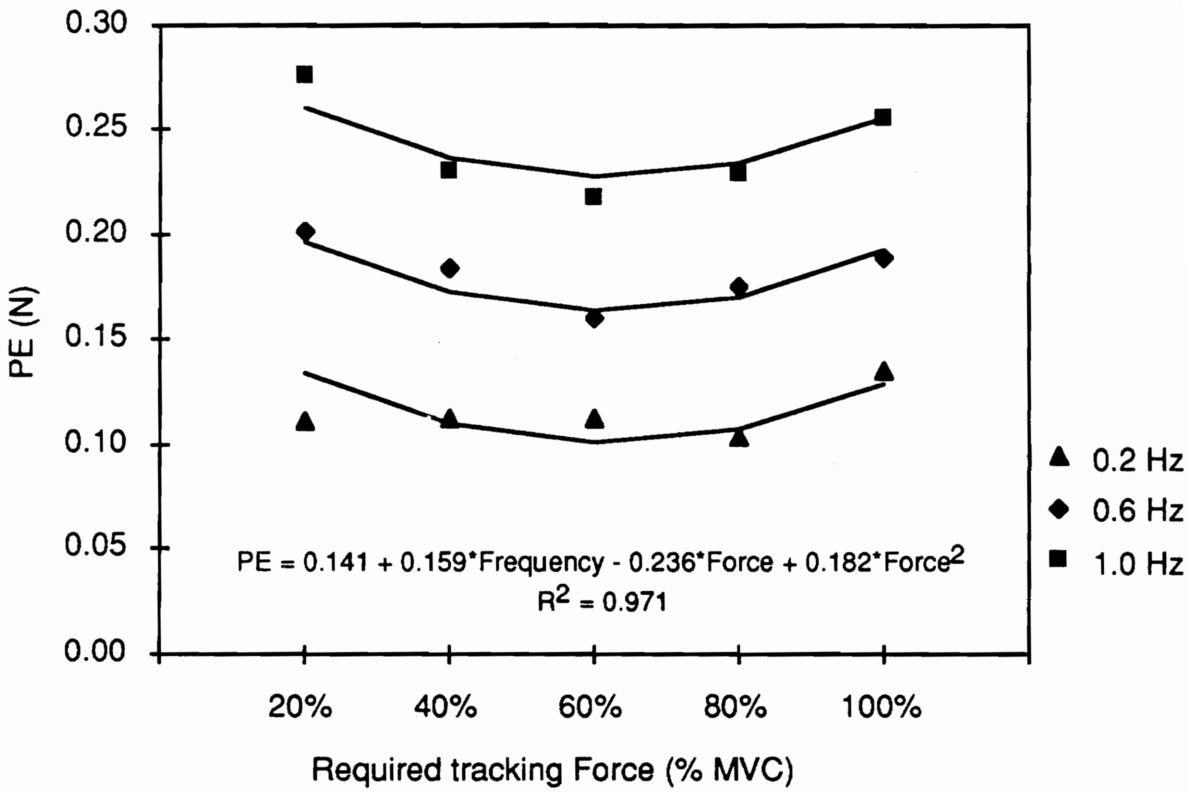


Figure 4.2 Two way interaction of Frequency and Force for proportional error and the polynomial regression model fit (solid lines).

Table 4.2 Breakdown of Proportional tracking Error (PE) for each part of the tracking task.

PE G-G corrected p-values	Overall Track	Flexion Phase exertion ('A')	 release ('B')	Extension Phase exertion ('C')	 release ('D')
ANOVA Effect					
Frequency	$p < 0.0001$	$p < 0.0001$	$p < 0.0001$	$p < 0.0001$	$p < 0.0001$
Force	NS††	NS††	NS	NS	NS††
Frequency x Force	NS *	NS	NS	NS	NS
Contrast for Frequency					
Linear	$p < 0.0001$	$p < 0.0001$	$p < 0.0001$	$p < 0.0001$	$p < 0.0001$
Quadratic	NS	NS	NS	NS	NS
Contrast for Force					
Linear	NS	NS	NS	NS	NS
Quadratic	$p < 0.0101$	$p < 0.0126$	NS	NS	$p < 0.0106$
Regression Model					
R^2	0.971	0.970	0.952	0.912	0.943
Optimum value	61.46%	66.52%	61.04%	65.27%	55.85%

* NS indicates not significant at $\alpha = 0.05$

††significant at the α=0.05 level without G-G correction

summarizes separate ANOVA findings for each part of the track along with the results from the overall tracking error analysis reported earlier. The results for each part are consistent with the overall tracking results. Again, the Force main effect teeters about significance with and without the conservative G-G correction. The quadratic contrast for the force effect also drops from significance during parts 'B' (flexion release) and 'C' (extension exertion) of the track.

4.3 Agonist/Antagonist Co-contraction

For each muscle, the four channels of iEMG data were corrected for resting level and then averaged to obtain a single measure of average activity for that muscle during the tracking task. For each subject and each trial, the integrated and averaged iEMG data were used in a regression model to determine the muscle force contributions from each muscle during the performance of the tracking task (Woldstad, 1989; Woldstad et al., 1988). The assumption of equivalent muscle groups allowed modeling of the entire flexor or extensor group using only the biceps and triceps respectively (see Section 2.4.1). A linear model was used to describe each muscle's force-iEMG relationship. The model equation took the form (Section 2.6.2.1):

$$NF(t) = B_1{*}iEMG_B(t) + B_2{*}iEMG_T(t) \qquad B_1>0;\ B_2<0 \qquad (4.5),$$

where NF is the net force observed at the wrist, $iEMG_B$ and $iEMG_T$ are the average, integrated electromyograms from the biceps and triceps at time t, and B_1 and B_2 are coefficients found using a least squares regression of actual tracking force at the wrist cuff against the integrated EMG (Woldstad, 1989). Figure 4.3 shows the estimated muscle force for the biceps and triceps groups. Figure 4.4 shows the modeled tracking force estimated from the least squares regression model (Equation. 4.5) and the actual tracking force. Since the biceps and triceps contribute oppositely to the observed force, the two-degree-of-freedom model accurately represents the force activity of the agonist/antagonist muscle pair. The coefficients (B_1 and B_2) have opposite signs in the model. The data was modeled separately for each trial condition resulting $\underline{R}^2$ values ranging from 0.63 to 0.93.

The coefficients B_1 and B_2 represent the linear components of each muscle group contributing to the total force. With this information and the value of the integrated iEMG for each trial condition it was possible to calculate the muscle force produced by each of the muscle groups involved in the performance of the tracking task (Figure 4.3). Measures of co-contraction as a ratio of antagonist muscle force to agonist muscle force (CR) or the absolute level of antagonist muscle force (AAF) were then determined for each trial.

4.3.1 Absolute Antagonist Force

With Absolute Antagonist Force (AAF) as the repeated measure, the same mixed-factor ANOVA was used to analyze co-active muscle behavior during the tracking task. Table 4.3 shows the ANOVA summary for AAF. The main effect of gender ($\underline{p} < 0.0003$) was found to be the only significant effect in the model with significantly higher antagonist force levels occurring within the male subject group. The Frequency by Force interaction was

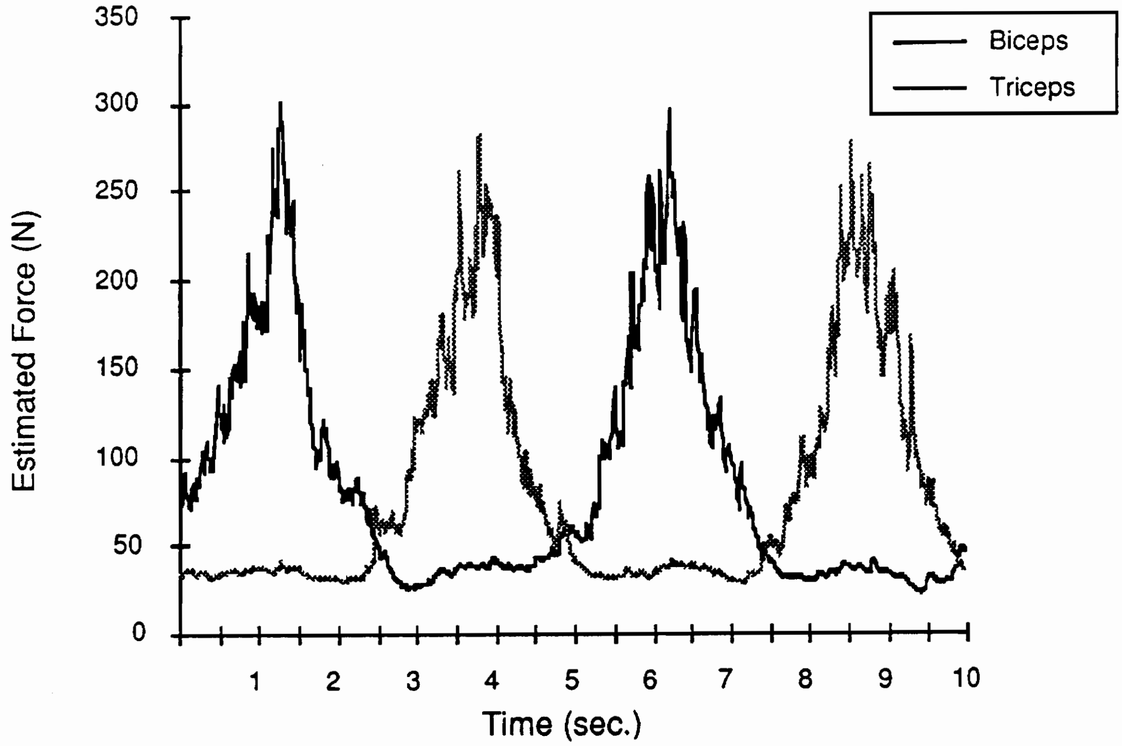


Figure 4.3 Data plot from a ten second tracking exertion (100% MVC, 0.2 Hz) showing estimated biceps and triceps muscle force contributing to the tracking exertion.

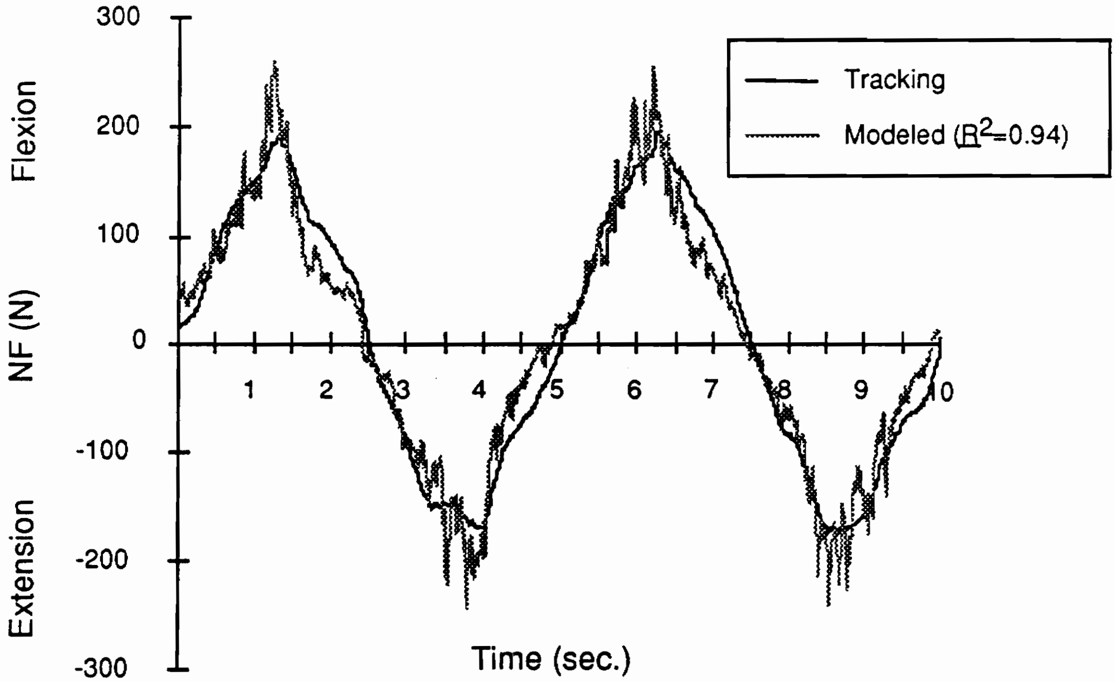


Figure 4.4 Data plot from a ten second sample sinusoidal tracking exertion (100% MVC, 0.2 Hz) showing isometric tracking force and modeled force.

Table 4.3 ANOVA summary table for AAF measure of co-activity.

ANOVA model for AAF					Type III SS	
Source	df	SS	MS	F-Value	P-Value	G-G
Gender	1	29179.38	29180.00	35.968	0.0003	
S(Gender)	8	6490.06	811.26			
Frequency	2	42.64	21.32	0.423	0.6621	0.6389
Frequency * Gender	2	3.84	1.92	0.038	0.9627	0.9491
Frequency * S(Gender)	16	806.27	50.39			
Force	4	422.44	105.61	0.842	0.5087	0.4417
Force * Gender	4	253.52	63.38	0.506	0.7319	0.5986
Force * S(Gender)	32	4011.91	125.37			
Frequency * Force	8	525.70	65.71	2.984	0.0066	0.0526
Frequency * Force * Gender	8	136.26	17.03	0.774	0.6273	0.5181
Frequency * Force * S(Gender)	64	1409.25	22.02			

significant at $\alpha=0.05$ without the G-G correction factor. Changes in Force ($p < 0.4417$) and Frequency ($p < 0.6389$) alone did not significantly effect the level of AAF during tracking. Analysis for each part of the tracking exertion revealed similar results except that Frequency was shown to be significant for the flexion release part ('B') of the task ($p < 0.0414$). The two-way interaction of Force and Frequency was also shown to be significant in parts 'A' ($p < 0.0391$) and 'B' ($p < 0.0152$) of the track. Appendix C shows the ANOVA summary table for AAF including the effects of phase and part of the tracking trial. AAF was found to be greater in the release parts ('B' and 'D') than in the exertion parts ('A' and 'C') of the tracking task ($p < 0.007$). AAF was also found to be significantly greater during extension than for flexion ($p < 0.001$). Figure 4.5 shows the interaction of Phase of tracking and Force level. As previously mentioned, the Force effect shown in Figure 4.5 is not significant.

4.3.2 Co-contraction Ratio

The ANOVA was also performed to determine the effects of the tracking parameters on the ratio of antagonist to agonist muscle activity. For each phase of the track, the Co-contraction Ratio (CR) was defined as the mean ratio of antagonist to agonist muscle force observed within that phase of the track. This has been one of the more popular measures of co-contraction found in the literature.

Table 4.4 shows the ANOVA summary table for CR. Significant main effects for CR were found for Force ($p < 0.0001$) and Frequency ($p < 0.0001$). The two-way interaction of Frequency and Force was also significant ($p < 0.0128$). CR increased with increasing target speed (Frequency) and decreased

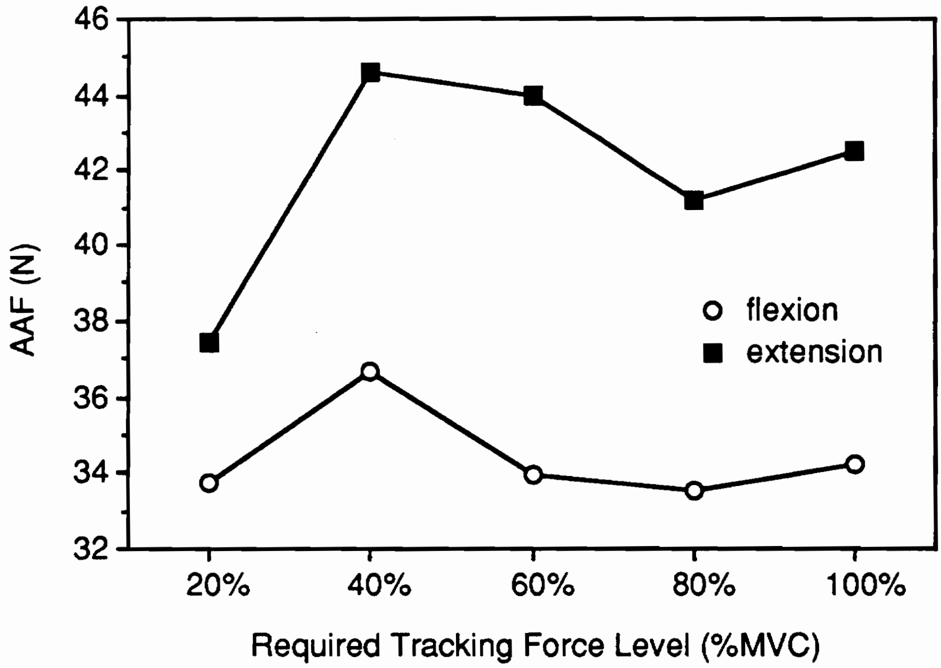


Figure 4.5 Two-way interaction of Phase and Force level for absolute antagonist force (AAF).

Table 4.4 ANOVA summary table for Co-contraction ratio (CR).

ANOVA model for CR					Type III SS	
Source	df	SS	MS	F-Value	P-Value	G-G
Gender	1	0.073	0.073	2.478	0.1541	
S(Gender)	8	0.237	0.03			
Frequency	2	0.267	0.134	31.871	0.0001	0.0001
Frequency * Gender	2	0.001	0.0005	0.113	0.8938	0.825
Frequency * S(Gender)	16	0.067	0.004			
Force	4	2.368	0.592	111.103	0.0001	0.0001
Force * Gender	4	0.006	0.001	0.263	0.8995	0.7895
Force * S(Gender)	32	0.171	0.005			
Frequency * Force	8	0.039	0.005	4.009	0.0007	0.0128
Frequency * Force * Gender	8	0.006	0.001	0.589	0.7834	0.6553
Frequency * Force * S(Gender)	64	0.078	0.001			

with increasing required tracking Force (Figure 4.6) A Newman-Keuls post-hoc test for the interaction effect showed that is was present only at very low levels of required tracking force.

Orthogonal polynomial contrasts reveal significant first order effects for both Frequency ($p < 0.0001$) and Force ($p < 0.0001$) as well as significant second order effects for Force ($p < 0.0004$) for CR. The second order effect is seen at high levels of tracking force where the co-contraction ratio asymptotes (Figure 4.6).

The analysis of CR for each part of the track reveals similar results to the analysis for the overall track. However, the Frequency by Force interaction was not significant for part 'B' ($p < 0.10$), 'C', ($p < 0.1812$) and 'D' ($p < 0.1050$) of the track. Also, CR *decreased* with increasing Frequency for the exertion parts of the task while it *increased* with increasing Frequency for the release parts of the task (Figure 4.7, $p < 0.0001$). Further analysis of the individual contributions of the agonists and antagonist muscles during flexion and release indicated that the change in CR over task part was due to both changes in the level of agonist activity as well as the level of antagonist activity. Figure 4.8 shows the two-way interaction of Part and Frequency for AAF ($p < 0.0001$) and the level of Agonist muscle activity ($p < 0.0001$). Agonist muscle force decreased rapidly with increasing target Frequency during release while antagonist activity increased slightly. Agonist activity remained almost constant for exertion at all Frequency levels while AAF decreased with increasing Frequency. In general, CR is greater during release (decreasing force) than in exertion (increasing force) at $p < 0.0001$. The co-

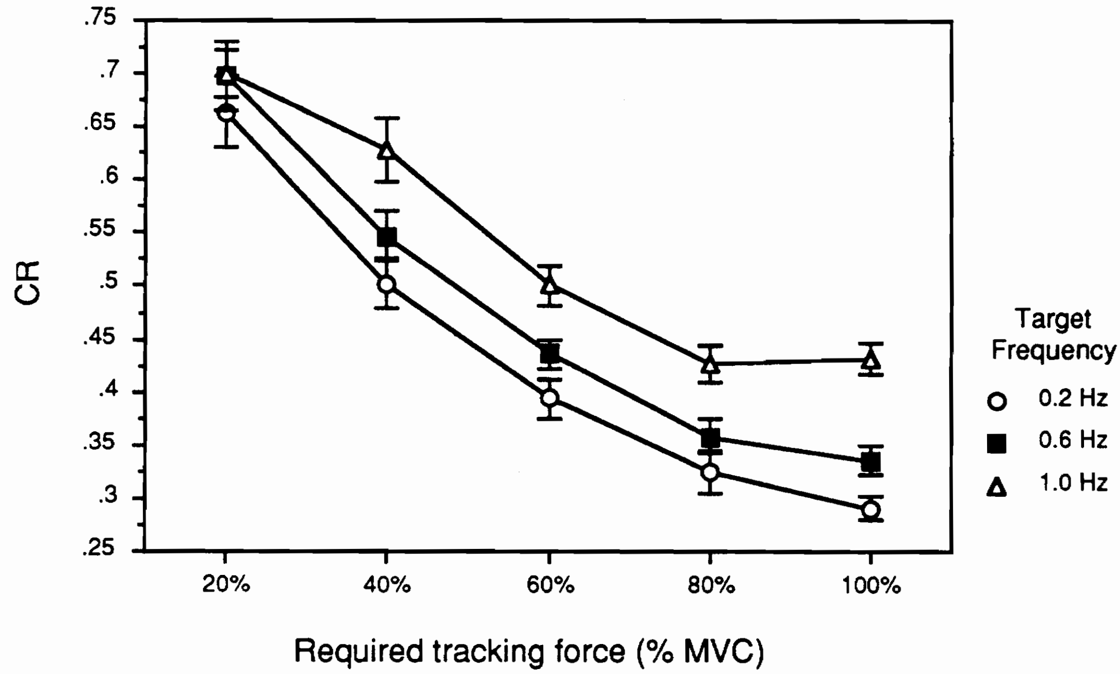


Figure 4.6 Two-way interaction of Force and Frequency for whole track Co-contraction ratio.

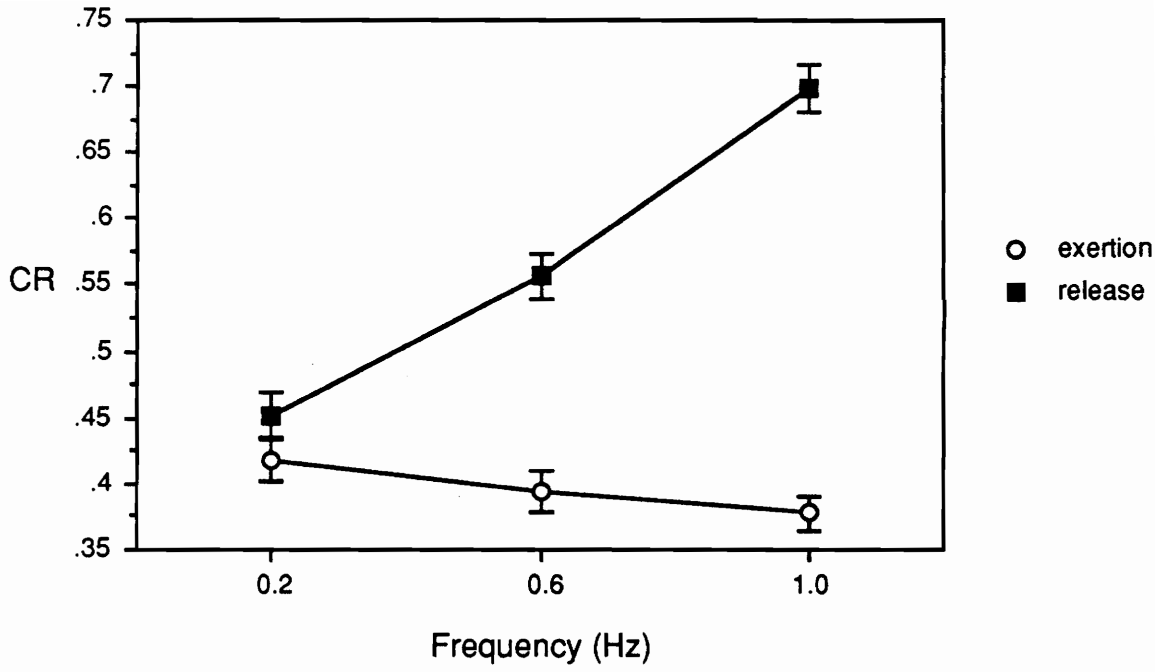


Figure 4.7 Co-contraction ratio as a function of Frequency for the different parts of the tracking phase: exertion and release.

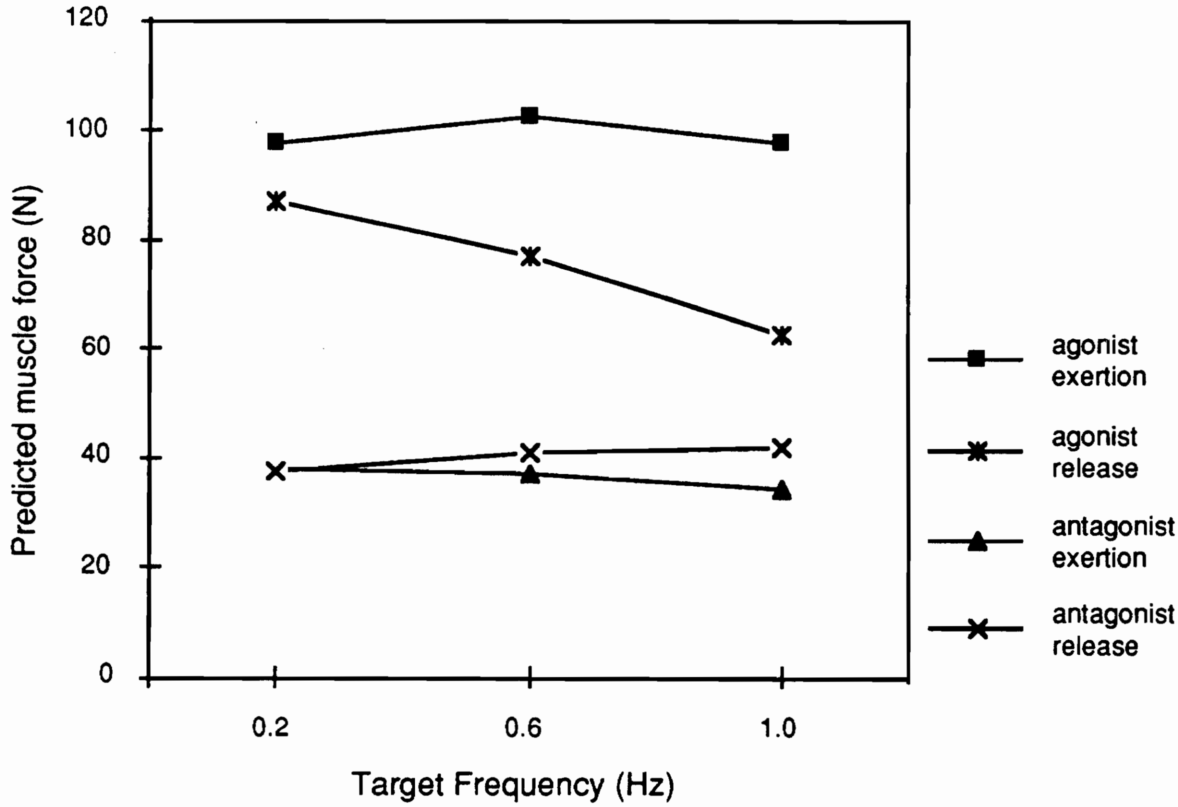


Figure 4.8 Two-way interaction of Part and Frequency for Agonist force and Antagonist muscle force (AAF) during exertion and release.

contraction ratio was also greater in extension than in flexion ($p < 0.003$). Appendix C shows the ANOVA analysis table including each phase and part of the tracking trial as effects.

4.4 Tracking Error and Actual Muscle Force

Another measure of interest for this study was the actual muscle force needed to perform the tracking task and its relationship to tracking error. The set levels of the Force for the tracking trials were a percentage of the *net* flexion and extension force obtained during the flexion and extension maximum trials. Therefore, the optimal force levels found for tracking were force values which underestimate the actual levels of biceps and triceps muscle activity because of the significant levels of muscle co-activation. A question arises as to the *actual* levels of biceps and triceps muscle force needed to perform the tracking tasks.

For each subject trial, the maximum biceps and triceps muscle force (MMF) was determined for their respective phases of the tracking task. These values were calculated as a percentage of the maximum biceps and triceps muscle force observed at the 100% MVC, 0.2 Hz condition. For example, the required biceps tracking force during flexion for each trial was expressed as a percentage of the maximum biceps muscle force observed during the flexion phase of the 100% MVC, 0.2 Hz tracking condition. The triceps maximum was found during the extension phase of the track. The values were averaged across frequency conditions to get a mean level of percent maximum muscle force (%MMF) activity required at each tracking force level.

The %MMF values were included in a quadratic regression model similar to that used to analyze proportional tracking error (Equation 4.4). Since the %MMF data were averaged across frequency conditions, there was no frequency term in the model. Table 4.5 compares the %MMF required to perform the tracking tasks to the net %MVC used in setting the control/response ratio for the tasks. The table also shows the level of significance for the difference between the %MMF and %MVC levels. The MMF values are consistently higher than the net force values, with the exception of the highest force level. This is due to the difficulty that the subjects had in reaching the 100% MVC force level in tracking. Both the flexion phase MMF ($\underline{R}^2 = 0.97$) and extension phase MMF ($\underline{R}^2 = 0.95$) regressions accurately predicted proportional tracking error and were significant at $\underline{p} < 0.0001$. Partial differential analysis with respect to the actual %MMF used for each trial revealed that minimum tracking error occurs at 66% MMF biceps during the flexion phase and 65% MMF triceps during the extension phase of the track. These values are greater than the predicted net optimal force of 61% MVC presented earlier. However, the statistical significance of this difference is unknown due to the difficulty in generating the appropriate statistic for the test. In particular, it is very hard to determine the correct standard error term for a comparison of predicted values generated from different regression models.

Table 4.5 Net force (%MVC) compared with actual percent of maximum muscle force (%MMF) values found for performance of the tracking task. Percent MMF values are averaged across Frequency conditions.

Required %MVC Net Force	Estimated Biceps %MMF (N)	Estimated Triceps %MMF (N)
20%	31.0% ($p<0.0001$)	34.5% ($p<0.0001$)
40%	49.4% ($p<0.0001$)	58.3% ($p<0.0001$)
60%	63.8% ($p<0.0618$)	74.2% ($p<0.0001$)
80%	81.9% ($p<0.2033$)	86.2% ($p<0.0129$)
100%	92.7% ($p<0.0003$)	91.8% ($p<0.0001$)

5. DISCUSSION

5.1 Overview

The purpose of this research was to examine co-active muscle coordination as it relates to the performance of an isometric tracking task. The results have implications in three areas: tracking performance, co-active muscle performance, co-active muscle performance as it relates to tracking performance.

5.2 Tracking Performance

5.2.1 Error and Force

The proportional tracking error (PE) measure revealed an optimal zone of tracking performance at median levels of required tracking force. The quadratic model fit showed that the best tracking performance occurred at approximately 61% of both maximum flexion and extension maximum voluntary contractions (MVC). This is consistent with the work of Berkowitz (1990) who also found a quadratic model for PE as a function of required tracking force. Berkowitz (1990) found a zone of optimal control at 66% of extension MVC.

When PE was fit against the biceps and triceps percent of maximum muscle force (%MMF), it also exhibited quadratic characteristics. Partial differential analysis with respect to the force term revealed that optimal tracking performance was found at approximately 66% MMF for biceps and 65% MMF for triceps. These optimal values are greater than the net optimal value of 61% MVC showing that net force analyses underestimate the actual muscle force contributions because of the presence of co-activation. As mentioned in Section 4.4, the statistical significance of this difference is difficult to measure due to the difficulty in generating the appropriate statistic. Optimal tracking performance occurs at approximately the same level of activity for both the biceps and triceps. This is contrary to the results of Berkowitz (1990) who found optimal tracking at approximately 41% net flexion activity and 58% net extension activity. Again, the presence of co-activation underestimates the actual muscle forces required to perform the tracking task.

Berkowitz (1990) hypothesized that increased tracking error at high required force levels (high C/R ratio) may be explained by the size principle of motor unit recruitment (Astrand and Rodahl, 1986). It may also be explained by the known relationships of force variability (Schmidt et al., 1979). The size principle of motor recruitment suggests that motor units are contracted in order of size with smaller (fine) motor units activating at low force levels and larger (coarse) motor units firing at higher force levels. The coarse motor units contribute to output force in larger quantities and therefore promote highly variable control. The size principle is consistent with the theories on impulse variability which states that increases in output force are accompanied by nearly proportional increases in force production error (Equation 2.7). Difficulty in tracking may be due to the inability to have fine control at high levels of force production.

Berkowitz (1990) also hypothesized that the increased tracking error at low force levels may be attributed to low level motor noise present in the neuromuscular system at all times. This motor noise proportionately increases as force is reduced thus creating more variable force output at low force tracking. This theory is analogous to decreasing the signal to noise ratio by decreasing the signal strength. People may have difficulty tracking at low force levels because the neuromuscular control signals get lost in the low level motor noise. As the control signal gets stronger, it is able to get above the noise to effectively control force production.

5.2.2 Error and Frequency

Tracking error was also shown to linearly increase with Frequency. Again, this result is consistent with the work of Berkowitz (1990). The impulse variability model (Schmidt, 1979 Equation 2.6) predicts this result. The model suggests that error in the level of force production will decrease with increasing movement time (slower tracking is more accurate). However, the increased tracking error found at higher tracking speeds is in conflict with the theories timing variability (Schmidt 1988, Equation 2.11). This theory suggests that timing error in force production increases as movement duration decreases (faster tracking is more accurate). The results of the tracking study show that tracking performance is better at *slower* tracking speeds. Although the two sources of variability (force and timing of force) may work opposite of each other, the results found here suggest that force variability is more fundamental to tracking performance error than timing variability.

As described in the Section 1.0, the task studied here was the simplest form of closed-loop manual control. Periodic sine wave tracking over long periods of time may eventually become an open-loop task requiring little or no visual feedback. The influence of force production error may become less pronounced as the tracking task becomes more open-loop. Under open-loop conditions, timing error (represented by tracking lag) may be the fundamental source of performance error.

5.2.3 Error and Part of Track

Proportional tracking error was found to be higher for the extension phase of the track. This may be due to the fact that pure extension effort is not a well learned behavior. Higher error would be expected in behaviors that are not well learned. Since, flexion is much more natural than extension, lower tracking error would be expected.

5.3 Co-active Muscle Performance

Two measures were used to quantify muscle co-activation for the tracking task: absolute antagonist activation (AAF) and the ratio of antagonist to agonist muscle activation (CR). In general, the results show that overall antagonist activity (AAF) remains constant over the varied tracking Force and target Frequency conditions examined here. This is consistent with the work of Woldstad (1989) who found largely constant levels of co-contraction for dynamic movements. However, AAF was shown to be greater in extension than in flexion.

Co-contraction ratio was greater for release (decreasing force) than for exertion (increasing force) for the track. Further, increases in co-contraction during the release part of the track were greater with increasing tracking Frequency (target speed) while they decreased with increasing Frequency in exertion (Figure 4.7). This means that higher rates of increasing force produced *less* co-contraction while higher rates of decreasing force produced *more* co-contraction. Changes in CR for exertion and release were due to systematic changes is agonist and antagonist activity across Frequency conditions (Figure 4.8). The increase in co-contraction for release was due to an increase in antagonist activity and a decrease in agonist activity as Frequency increased. However for exertion, the agonist activity remained nearly constant while the antagonist dropped with increasing Frequency. The hypothesis that co-contraction is greater at higher rates of force production (Smith, 1987) seems to be dependent on the direction of force production (exertion or release). DeLuca and Mambrito (1987) showed that co-activation would be present in tasks where uncertainty exists (tracking) or under conditions of force reversal such as in changing tracking direction from exertion to release. They also suggest that this co-activity does not simply turn on and off but changes in proportionality. This may explain the proportional shifts in co-contraction at the points of force reversal in the tracking task.

One theory to explain this change in co-activity for exertion and release is that high velocity force productions exhibit the properties of reciprocal inhibition of antagonist muscles because they are similar to ballistic movements. However, since increased control is needed at high rates of relaxation (more uncertainty), greater co-contraction is needed to stabilize the joint.

Greater co-contraction in release was also found in the work of Redfern (1988) and Basmajian and DeLuca (1985). These studies hypothesized that the increased co-contraction is needed when the rate of decreasing muscle force is greater than the relaxation rate of the primary mover. In this case, co-contraction is needed to help achieve faster rates of force release (or deceleration). Since no significant increases in tracking error (PE) were found between the exertion and release aspects of the task it is hypothesized that co-contraction played a significant part in facilitating tracking performance during the release parts of the track.

As found here, Redfern (1988) and Woldstad et al. (1988) have also shown greater muscle co-activation in the extension phase than in the flexion phase of isometric tasks. Although the reason for this finding is still unknown, it may be due the inability of subjects to perform the awkward task of high force extension. Whereas the flexor group of the elbow is often used to lift, carry, and place objects, humans are not as used to performing high precision elbow extension tasks. This leads to greater movement uncertainty and therefore more co-contraction of the muscles involved (DeLuca and Mambrito, 1987). This increased co-contraction was accompanied by higher tracking error (PE) in the extension phase.

These results regarding muscle co-activation confirm hypothesis **H_1** which stated that **significant levels of muscular co-contraction are present in the performance of the isometric tracking task**. The results also confirm H_3 which stated that **co-contraction is higher for isometric extension than in isometric flexion of the elbow**.

5.3 Co-contraction as it Relates to Tracking Performance

The previous sections have discussed the relevant findings for tracking error performance and muscle co-activation separately. The central question of this research is whether or not a characteristic co-activation pattern exists which explains the optimal tracking performance at median levels of tracking force found by Berkowitz (1990). These results confirm that a quadratic relation exists between tracking error (PE) and required tracking Force (Figure 4.2). This quadratic relation reveals that optimum tracking performance is found at near medial levels of net force activation. However, no complimentary relation was found for co-activation of the biceps and triceps muscles measured either in absolute (AAF, Figure 4.5) or relative (CR, Figure 4.6) terms. Near constant levels of antagonist activity over all tracking conditions led to a general decline in the co-contraction ratio as the required tracking force increased. This is contradictory to the hypothesis **H_2** which stated that **co-contraction was a second order function of required tracking force revealing a optimal level of co-activity resulting in optimal tracking performance**. Although not significant, the characteristic function of absolute antagonist force (AAF) and tracking force level (Figure 4.5) shows a potential optimal level of antagonist activity for tracking. However, the effect was not significant in this study. It is possible that an increased sample size, data smoothing, or alternate muscle model equations would reduce the variability in the data, yielding a significant result. These issues are covered in Section 5.4.

5.4 Difficulties with the Methods and Analysis

As mentioned previously, there are areas for improvement for this study that may lead to a reduction in data variability and an improvement in the strength of interpretation. These areas include:

- *Data Smoothing*: filtering the averaged iEMG signal prior to performing the Force-EMG model fits would reduce the variability in the data and lead to higher $\underline{R}^2$ coefficients.

- *Alternate Muscle Models*: although the linear model of Force-iEMG worked well for each trial's regression fit, the results were not largely generalizable between trial conditions. Use of quadratic, exponential, or logarithmic models may increase the generalizability of the results indicating a more accurate description of the Force to iEMG modeling component of this study.

- *EMG sensitivity*: The use of electomyograms at low levels of muscle activity result in poor data resolution and, therefore, poor interpretability due to the effects of internal and external noise. The muscle models for the 20% MVC force conditions were generally poorer in explaining the net force variability than the high force condition models. Here again, muscle component force estimates are constrained in accuracy by the fit of the muscle model for each trial. There is no good solution to the problem of low level EMG sensitivity, but it must be kept in mind as results are interpreted.

- *Modeling Time Delays*: there is usually a lag between the detection of electromyographic activity and the onset of force production. Computation of the muscle force measures reported here did not take this lag into account. The relative lag is more exaggerated at high tracking speeds which leads to difficulty in defining the phases of force production and, therefore, the quantification of co-activity. This effect is much less at the lower tracking frequency conditions. However, correction of each trial fit for this lag would result in greater muscle model accuracy and better measures of co-contraction.

One final area of concern in this research was the instability of the modeled isometric Force-iEMG relationship for each subject. In theory, the Force-iEMG relation should not change for a subject regardless of the force or speed of contraction. However, the models did change across tracking conditions which made it necessary to model each condition separately. Subsequent determination of the force components were constrained by the accuracy of each trial fit – some fits were better than others. In other words, the relationship between net output force and muscle iEMG was changing between each trial condition. There are several possible reasons for the lack of generality in this relationship:

1.) assumption of a linear Force-iEMG relationship was not adequate and use of higher order models was necessary. It should be kept in mind that there is wide disagreement on the exact form of the isometric Force-EMG relationship (Section 2.6).

2.) change in the resistance of the electrical connection at the skin surface over the duration of the trial run. This effect may have been reduced by balancing the order of presentation across force levels.

3.) fatigue induced increases in the iEMG signal potentials resulted in changes in the Force-iEMG model. However, the effects of fatigue on the overall results was minimal as the experimental design was balanced over Force conditions and the trial time was kept small.

4.) the isometric Force-EMG relationship is not constant but is a dynamic relationship which varies with the conditions of the exertion.

Further research into these areas is needed to insure an understanding of the nature of the Force-iEMG relation.

5.5 Future Research

The findings regarding performance error and co-activation of muscles during tracking generates future research questions. There are many areas of research that could bring about a better understanding of the neuromuscular control of goal oriented exertions. There is also much work to be done concerning tracking performance under varying control and control/response conditions. In summary, future areas for research might include:

- Investigation into isometric tracking performance with higher order control systems and with more random forcing functions.

- Investigation of open versus closed loop tracking behavior by including analysis of performance error due to lag (timing variability).

- Analysis of individual motor unit activation in the investigation of the neuromotor noise explanation of increased proportional tracking error at low required force levels.

- Analysis of individual motor unit activation in the investigation of the motor unit recruitment explanation for increased tracking error at higher levels of required tracking force.

- Further analysis into the exact nature of the Force-EMG relationship. More specifically, does this relationship really have a dynamic nature; changing with varying performance conditions? Do higher order models exist which have more generality among varying exertion conditions.

- Investigation into smoothing techniques to increase the ability to model the individual muscle force contributions from EMG data.

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APPENDICES

APPENDIX A INSTRUCTIONS AND INFORMED CONSENT

Isometric Tracking Study
Instructions for Experiment

Thank you for agreeing to participate in this study. This study is being conducted by the Industrial Ergonomics Laboratory of the Human Factors Engineering Center at Virginia Tech. The experiment is being run by Chris Rockwell under the supervision of Dr. Jeff Woldstad, a professor in Industrial and Systems Engineering.

The purpose of this study is to investigate muscle activity during the performance of a manual control task using a wrist cuff. The wrist cuff is activated by the force applied to it rather than by wrist cuff movement. When force is applied to the wrist cuff, the position of the computerized object will move depending on the magnitude of the force applied: more force leads to move object movement. At no time during the study will the wrist cuff move.

The task you will be performing is a tracking task. The tracking involves following a moving target on a computer screen with an object which you will control using the wrist cuff. The target is a green circle which will be moving horizontally, back and forth, across the screen. The object you are controlling is a yellow cross called the cursor. Your job is to try and keep the yellow cross inside the green circle at all times. You can do this by altering the amount of force applied to the wrist cuff.

We are researching how accurately you can track the target under differing target movement and control response levels. We are particularly interested in how your muscle activity changes with the different tracking conditions. During some trials the target will move faster than others or the control will seem more sensitive to force input. The most difficult trials will require almost all of your biceps and triceps force while the easiest trials will require very little force. Please attempt to track the target as accurately as possible for all experimental trials. Your biceps and triceps muscle activity will be recorded using a technique called electromyography (EMG) which allows measurement of the electrical activity in your muscles by way of highly sensitive electrodes placed on the surface of your skin.

The experiment will take approximately one hour. You will be paid $5.00 per hour. After reading these instructions and filling out an informed consent form, you will be ready for the experiment. First, your upper arm must be prepared for the placement of the EMG electrodes. This includes a brisk rubbing of the skin, a cleaning with alcohol, and possibly the shaving of excess arm hair. The EMGs will be placed on the skin using double sided adhesive tape. Once the EMGs are in place you will be seated in the experimental chair and held in position with Velcro straps.

Before the tracking trials are performed, you will be asked to perform six maximum strength trials with the isometric wrist cuff. It is important that you give your maximum safe effort during the these trials. The strength trials will be followed by a set of 15 tracking trials as described above. The risk of muscle strain associated with the procedures used is approximately 1 in 1000 people. There may be some discomfort and hair loss upon the removal of the EMG electrodes as they are attached with a double sided adhesive tape. Also, some redness may remain on the skin for a day or two after the EMG electrodes are removed. If, for any reason, you feel that you are experiencing pain or abnormal discomfort during the session, please inform the experimenter immediately. Please feel free to stop the experiment if you need any clarification about the study, or if you need additional rest.

The data from the experiment should be analyzed by May, 1992. The results will be made available to you should you desire to review them. The research team members are:

Chris Rockwell, Graduate Assistant, ISE
Dr. Jeffrey Woldstad, Professor, ISE

If you have any questions or concerns about the way you have been treated during this experiment, please contact Dr. E. R. Stout at (703) 231-9359.

Thank you again for your participation. If at any time today you have questions about the experiment please feel free to ask the experimenter. We hope you enjoy your experience in this research effort.

Isometric Tracking Study: Informed Consent Form

This form constitutes informed consent by you to participate in this study. Please read it carefully, as well as the attached sheet, and then sign it below.

Your Rights as a Subject are:

1.) It is your right as a subject to withdraw from the study at any time and for any reason.

2.) Any of the research team members will answer any questions that you may have, and you should not sign this consent form until you understand fully all of the terms involved.

3.) You have a right to see your data and withdraw it from the study if you so desire. Please inform the experimenter immediately of this decision, as the data will be handled anonymously and not possible to track once the session is over.

4.) You have the right to be informed of any risks or discomforts in this research. There is minimal risk associated with this experiment. You may experience some muscle fatigue and soreness after the completion of the experiment. You may also experience some discomfort or redness upon removal of the EMG electrodes attached by adhesive tape. However, the fatigue and redness should be short-lived and pose no further complication or discomfort to you.

5.) Should any further questions arise, please contact one of the team members. If you have any concerns about the way the experiment is being conducted or the way you are being treated, you may contact Dr. Stout's office at 231-9359

Your participation is greatly appreciated and we hope that you will find the study a pleasant and interesting experience. Your signature below indicates that you have read this document and the description of the experiment attached to it in its entirety, that your questions have been answered, and that you consent to participate in the study described.

Signature: ______________________________ Date ____________

Address: ______________________________

____

APPENDIX B DATA MANIPULATION PROGRAMS

Table B.1 Table showing a description of the data analysis programs used generate, manipulate and analyze the EMG and tracking data.

Program	Function	Data Form
EMGTRK.pas	• collect force and iEMG data from subject run	Raw Data
PREREG.pas	• convert load cell data to force • correct for resting iEMG and average iEMG data from each muscle • format data for regression analysis	Pre-regression form
SYSTAT statistics package	• obtain regression coefficients for muscle activity: Force=iEMGb + iEMGt	Regression by statistics package
POSTREG.pas	• use regression coefficients to convert average iEMG data to force data • format data for final compilation	Post regression form
TRACKING ANALYSIS.pas	• generate all dependent measures from post regression form data	Compiled subject data
SuperANOVA statistics package	• perform ANOVA on dependent measures • perform orthogonal contrasts to determine nature of data • regress data to find model parameters	Analyzed data

APPENDIX C EXTENDED ANOVA TABLES

Table C.1 ANOVA summary table for proportional tracking error (PE) by phase and part of the tracking task.

Breakdown ANOVA for PE						Type III SS
Source	df	SS	MS	F-Value	P-Value	G-G
Gen	1	0.098	0.098	0.922	0.365	
S(Gen)	8	0.852	0.107			
Phase	1	0.218	0.218	17.059	0.003	0.003
Phase*Gen	1	0.015	0.015	1.171	0.311	0.311
Phase*S(Gen)	8	0.102	0.013			
Part	1	0.005	0.005	0.633	0.449	0.449
Part*Gen	1	0.017	0.017	2.082	0.187	0.187
Part*S(Gen)	8	0.066	0.008			
Freq	2	1.622	0.811	61.065	0.000	0.000
Freq*Gen	2	0.012	0.006	0.445	0.648	0.568
Freq*S(Gen)	16	0.213	0.013			
Force	4	0.102	0.026	2.763	0.044	0.082
Force*Gen	4	0.017	0.004	0.466	0.760	0.664
Force*S(Gen)	32	0.296	0.009			
Phase*Part	1	0.008	0.008	5.696	0.044	0.044
Phase*Part*Gen	1	0.000	0.000	0.306	0.595	0.595
Phase*Part*S(Gen)	8	0.011	0.001			
Phase*Freq	2	0.011	0.006	2.599	0.105	0.124
Phase*Freq*Gen	2	0.011	0.006	2.623	0.104	0.123
Phase*Freq*S(Gen)	16	0.034	0.002			
Part*Freq	2	0.008	0.004	3.462	0.056	0.082
Part*Freq*Gen	2	0.005	0.002	2.008	0.167	0.185
Part*Freq*S(Gen)	16	0.020	0.001			
Phase*Force	4	0.006	0.001	0.444	0.776	0.714
Phase*Force*Gen	4	0.006	0.002	0.482	0.749	0.689
Phase*Force*S(Gen)	32	0.100	0.003			
Part*Force	4	0.018	0.005	1.426	0.248	0.263
Part*Force*Gen	4	0.009	0.002	0.697	0.600	0.551
Part*Force*S(Gen)	32	0.102	0.003			
Freq*Force	8	0.049	0.006	1.032	0.422	0.408
Freq*Force*Gen	8	0.029	0.004	0.598	0.776	0.675
Freq*Force*S(Gen)	64	0.384	0.006			

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Table C.1 cont.

Breakdown ANOVA for PE (cont.)						Type III SS
Source	df	SS	MS	F-Value	P-Value	G-G
Phase*Part*Freq	2	0.001	0.001	0.372	0.695	0.690
Phase*Part*Freq*Gen	2	0.001	0.000	0.268	0.768	0.763
Phase*Part*Freq*S(Gen)	16	0.027	0.002			
Phase*Part*Force	4	0.009	0.002	0.977	0.434	0.415
Phase*Part*Force*Gen	4	0.018	0.005	1.946	0.127	0.156
Phase*Part*Force*S(Gen)	32	0.075	0.002			
Phase*Freq*Force	8	0.010	0.001	0.369	0.933	0.828
Phase*Freq*Force*Gen	8	0.070	0.009	2.685	0.013	0.049
Phase*Freq*Force*S(Gen)	64	0.209	0.003			
Part*Freq*Force	8	0.017	0.002	0.893	0.528	0.464
Part*Freq*Force*Gen	8	0.011	0.001	0.568	0.800	0.652
Part*Freq*Force*S(Gen)	64	0.149	0.002			
Phase*Part*Freq*Force	8	0.013	0.002	0.610	0.766	0.640
Phase*Part*Freq*Force*Gen	8	0.018	0.002	0.867	0.549	0.485
Phase*Part*Freq*Force*S(Gen)	64	0.168	0.003			

Table C.2 ANOVA summary table for co-contraction ratio (CR) by phase and part of the tracking task.

Breakdown ANOVA for CR — Type III SS

Source	df	SS	MS	F-Val	P-Val	G-G
Gen	1	0.294	0.294	2.487	0.154	
S(Gen)	8	0.946	0.118			
Phase	1	0.978	0.978	17.860	0.003	0.003
Phase*Gen	1	0.133	0.133	2.436	0.157	0.157
Phase*S(Gen)	8	0.438	0.055			
Part	1	4.477	4.477	121.311	0.000	0.000
Part*Gen	1	0.004	0.004	0.103	0.757	0.757
Part*S(Gen)	8	0.295	0.037			
Freq	2	1.070	0.535	31.994	0.000	0.000
Freq*Gen	2	0.004	0.002	0.117	0.890	0.821
Freq*S(Gen)	16	0.268	0.017			
Force	4	9.472	2.368	110.853	0.000	0.000
Force*Gen	4	0.022	0.006	0.262	0.900	0.790
Force*S(Gen)	32	0.684	0.021			
Phase*Part	1	0.155	0.155	13.352	0.007	0.007
Phase*Part*Gen	1	0.005	0.005	0.468	0.513	0.513
Phase*Part*S(Gen)	8	0.093	0.012			
Phase*Freq	2	0.004	0.002	0.850	0.446	0.421
Phase*Freq*Gen	2	0.000	0.000	0.005	0.995	0.985
Phase*Freq*S(Gen)	16	0.037	0.002			
Part*Freq	2	2.046	1.023	130.263	0.000	0.000
Part*Freq*Gen	2	0.001	0.001	0.087	0.917	0.788
Part*Freq*S(Gen)	16	0.126	0.008			
Phase*Force	4	0.087	0.022	5.376	0.002	0.007
Phase*Force*Gen	4	0.013	0.003	0.817	0.524	0.491
Phase*Force*S(Gen)	32	0.129	0.004			
Part*Force	4	0.020	0.005	2.841	0.040	0.078
Part*Force*Gen	4	0.006	0.002	0.897	0.477	0.438
Part*Force*S(Gen)	32	0.057	0.002			
Freq*Force	8	0.156	0.020	3.982	0.001	0.013
Freq*Force*Gen	8	0.023	0.003	0.583	0.788	0.660
Freq*Force*S(Gen)	64	0.314	0.005			

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Table C.2 cont.

Breakdown ANOVA for CR (cont.)					Type III SS	
Source	df	SS	MS	F-Value	P-Value	G-G
Phase*Part*Freq	2	0.029	0.014	13.858	0.000	0.004
Phase*Part*Freq*Gen	2	0.005	0.003	2.622	0.104	0.139
Phase*Part*Freq*S(Gen)	16	0.017	0.001			
Phase*Part*Force	4	0.012	0.003	2.385	0.072	0.098
Phase*Part*Force*Gen	4	0.001	0.000	0.255	0.905	0.849
Phase*Part*Force*S(Gen)	32	0.042	0.001			
Phase*Freq*Force	8	0.048	0.006	1.551	0.158	0.215
Phase*Freq*Force*Gen	8	0.017	0.002	0.546	0.818	0.694
Phase*Freq*Force*S(Gen)	64	0.246	0.004			
Part*Freq*Force	8	0.012	0.002	0.969	0.468	0.428
Part*Freq*Force*Gen	8	0.005	0.001	0.361	0.937	0.798
Part*Freq*Force*S(Gen)	64	0.103	0.002			
Phase*Part*Freq*Force	8	0.009	0.001	1.021	0.430	0.406
Phase*Part*Freq*Force*Gen	8	0.004	0.001	0.443	0.891	0.750
Phase*Part*Freq*Force*S(Gen)	64	0.073	0.001			

Table C.3 ANOVA summary table for absolute antagonist force (AAF) by phase and part of the tracking task.

Breakdown ANOVA for AAF						Type III SS
Source	df	SS	MS	F-Value	P-Value	G-G
Gen	1	1.2E5	1.2E5	35.968	0.000	
S(Gen)	8	2.6E4	3245.03			
Phase	1	8498.30	8498.30	10.791	0.011	0.011
Phase*Gen	1	3476.42	3476.42	4.414	0.069	0.069
Phase*S(Gen)	8	6300.34	787.54			
Part	1	2005.68	2005.68	12.962	0.007	0.007
Part*Gen	1	459.19	459.19	2.968	0.123	0.123
Part*S(Gen)	8	1237.92	154.74			
Freq	2	170.58	85.29	0.423	0.662	0.639
Freq*Gen	2	15.36	7.68	0.038	0.963	0.949
Freq*S(Gen)	16	3225.13	201.57			
Force	4	1689.73	422.43	0.842	0.509	0.442
Force*Gen	4	1014.09	253.52	0.506	0.732	0.599
Force*S(Gen)	32	1.6E4	501.48			
Phase*Part	1	18.27	18.27	0.146	0.712	0.712
Phase*Part*Gen	1	47.66	47.66	0.381	0.554	0.554
Phase*Part*S(Gen)	8	1001.06	125.13			
Phase*Freq	2	63.29	31.65	0.814	0.461	0.432
Phase*Freq*Gen	2	25.02	12.51	0.322	0.730	0.667
Phase*Freq*S(Gen)	16	622.35	38.90			
Part*Freq	2	1509.45	754.72	27.022	0.000	0.000
Part*Freq*Gen	2	156.86	78.43	2.808	0.090	0.115
Part*Freq*S(Gen)	16	446.87	27.93			
Phase*Force	4	663.47	165.87	3.953	0.010	0.028
Phase*Force*Gen	4	76.11	19.03	0.454	0.769	0.686
Phase*Force*S(Gen)	32	1342.56	41.96			
Part*Force	4	129.41	32.35	3.232	0.025	0.064
Part*Force*Gen	4	149.53	37.38	3.735	0.013	0.045
Part*Force*S(Gen)	32	320.30	10.01			
Freq*Force	8	2102.74	262.84	2.984	0.007	0.053
Freq*Force*Gen	8	545.05	68.13	0.774	0.627	0.518
Freq*Force*S(Gen)	64	5637.09	88.08			

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Table C.3 cont.

Breakdown ANOVA for AAF (cont.)						Type III SS
Source	df	SS	MS	F-Value	P-Value	G-G
Phase*Part*Freq	2	13.41	6.70	0.278	0.761	0.627
Phase*Part*Freq*Gen	2	0.80	0.40	0.016	0.984	0.913
Phase*Part*Freq*S(Gen)	16	386.27	24.14			
Phase*Part*Force	4	113.99	28.50	1.939	0.128	0.188
Phase*Part*Force*Gen	4	52.30	13.08	0.890	0.481	0.410
Phase*Part*Force*S(Gen)	32	470.40	14.70			
Phase*Freq*Force	8	307.36	38.42	1.421	0.205	0.262
Phase*Freq*Force*Gen	8	85.33	10.67	0.395	0.920	0.751
Phase*Freq*Force*S(Gen)	64	1730.03	27.03			
Part*Freq*Force	8	125.21	15.65	3.411	0.003	0.024
Part*Freq*Force*Gen	8	22.26	2.78	0.607	0.769	0.647
Part*Freq*Force*S(Gen)	64	293.63	4.59			
Phase*Part*Freq*Force	8	23.61	2.95	0.424	0.903	0.786
Phase*Part*Freq*Force*Gen	8	19.52	2.44	0.350	0.942	0.838
Phase*Part*Freq*Force*S(Gen)	64	446.06	6.97			

VITA

Christopher John Rockwell was born on December 30, 1966 in Columbus, Ohio. He received his B.S. in Industrial and Systems Engineering at The Ohio State University in June of 1990. He went on to pursue his graduate work in the Human Factors Engineering Laboratory at Virginia Polytechnic Institute and State University, Blacksburg, Virginia. He is an active member of both the Institute of Industrial Engineers and the Human Factors Society. He will be working for the Hewlett-Packard Company in Fort Collins, Colorado on the design and improvement of computer based information systems.

Christopher John Rockwell

Christopher John Rockwell